

A BIOMECHANICAL COMPARISON BETWEEN TWO SHOTPUTTING TECHNIQUES
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Shotputting in competition at or above the intercollegiate level has traditionally been limited to two techniques: variations of the standard "O'Brien" glide and the rotational method. Recently, a third technique, described as the revised glide, has been used successfully in inter-collegiate competition.

The purpose of this study was to quantify selected kinematic and temporal variables describing the standard glide (SG) and the revised glide (RG) techniques. Four female collegiate-level shotputters were filmed at 100 frames per second during two test sessions one week apart. Subjects performed 10 trials for each of the two techniques. The three best and three worst trials for each technique based upon the measured distance of the put were digitized and the data smoothed using a cubic spline routine to reduce filming and digitizing errors. The mean absolute difference of 2.1% between calculated and measured distances was considered acceptable.

Mean values for the three best and three worst puts were computed for each technique. The mean differences for selected temporal and kinematic variables were statistically evaluated ($\alpha = 0.05$) using a within-subjects procedure. The mean values for the best and worst trials were compared within each technique. In addition, comparisons were made between the best trial values for each technique as well as between the worst trial values.

Summary of the Between Condition Results. The means of the three best trials and the three worst trials describing each technique were compared to evaluate possible differences between techniques. Two subjects exhibited significantly greater best mean trial measured and corrected distances for the RG technique. All of the best trial distances for the RG were either statistically equal to or greater than the SG distances.

The number of between techniques significant differences for all subjects were similar (49 to 71% for the best trials and 34 to 63% for the worst trials). There was no apparent relationship between the distance thrown and the number of significant differences. The mean percentage of significant differences for worst trial comparisons (48.8%) was slightly less than for the best trials (58.0%). The phase duration variables exhibited the greatest percentage of significant differences (87.5 and 81.3% for the best and worst trials, respectively). This was expected based upon the obvious differences between the two performance techniques. The implement average velocity variables were shown to have the second highest percentage of significant comparisons (81.3 and 43.8%). This was also expected based upon the differences in phase durations and velocities at the beginning of the push-off phase between the two techniques.

The best trial mean implement height at release was significantly different for only one subject, despite significant differences in distance exhibited between techniques by two subjects. Only one subject exhibited a greater mean vertical implement release velocity and release angle for the RG.

Significantly greater implement horizontal velocities were observed for all subjects at the start of the push-off phase for the SG, however, no significant

differences were exhibited by any of the subjects for the horizontal release velocities. In two of the four cases, the duration of the push-off phase was significantly greater for the SG. Of the remaining two subjects, one exhibited a significantly shorter duration for the SG, while the other exhibited no significant difference.

Three of the four subjects generated a greater horizontal velocity of the system CM at the beginning of the push-off phase for the SG. None of these subjects exhibited a significant difference between horizontal velocities of the system CM at release. This is reasonable since the horizontal velocity of the system CM must be near zero at release to prevent movement past the toeboard. This would require a greater impulse in the direction opposite the throw. The greater horizontal velocity observed for the system CM at the start of the push-off phase for the SG could have been the result of a "rocking" motion of the trunk toward the front of the circle during the transitional phase and continuing into the push-off phase. This type of motion could affect the athlete's use and generation of forces during the phase resulting in the development of a lesser impulse for both the shot and the system CM.

Summary of the Within Condition Results. The means of the three best trials and the three worst trials for each technique were compared to ascertain what differences distinguished successful from unsuccessful performances. The percentages of significant differences were considerably less than the between technique values, with a mean of 20.0 % within the SG technique and 14.8% within the RG technique. These results were expected and probably due to the fairly rigid trial acceptance criteria, in that the measured distance had to be at least 90% of the maximum pre-test distance. The actual number of meaningful differences was even smaller since the mean results are inflated by the significant differences attributed to distance measures which were the basis for the trial groupings. Excluding these significant differences, only 15.2 and 9.7% of the comparisons were significantly different for the SG and RG, respectively.

The basic factors affecting the distance of a put are the horizontal distance in front of the toeboard achieved prior to release, the height of release, the speed of release, and the angle of release. Very few trends for any of these parameters were observed across subjects.

Conclusion. The objective of shotputting is to maximize the measured distance of the put. All subject performances using the RG technique were significantly greater than or equal to the SG performances. Within the constraints of the study, and accounting for release distance, it was concluded that the RG technique produced better overall results. These results were accomplished by imparting a greater impulse to the shot in the horizontal direction during the push-off phase. If the RG can be accomplished so that the horizontal velocity at the start of the push-off phase is increased without compromising either body position or a change in velocity during the phase, the potential exists for an even greater horizontal release velocity of the implement. If this is accomplished without compromising the vertical release velocity, the horizontal distance of the put will be increased even more.

MAXIMAL MUSCLE POWER OUTPUT IN BICYCLING AS A FUNCTION OF RIDER POSITION,
RATE OF PEDALLING AND DEFINITION OF MUSCLE LENGTH

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INTRODUCTION In competitive cycling, especially in sprint cycling, the power output of the rider is arguably the most important indicator for performance. Design parameters of a bicycle-rider system which significantly influence the power output are the crank length, the positions of the seat and handle bar and the gear ratio. The purpose of this study was to compare the maximal value of the average power output of the lower limb muscles of a bicycle-rider system for two different definitions of the lengths of muscles, and to study the dependence of this power on the design parameters mentioned above.

METHODS The mechanical model of the bicycle-rider system is shown in Figure 1. It consists of five rigid bodies located in a plane (pelvis, thigh, shank, crank, and seat tube), and four smooth pin joints (hip, knee, pedal shaft, and crank shaft). The musculoskeletal model is shown in Figure 2. Individual muscles crossing the knee and/or hip-joint were grouped into seven functional units. The muscles crossing the ankle joint were not considered in this study. r , α , h , and $\dot{\theta}$, denote the crank length, pelvic inclination, seat height, and rate of crank rotation, respectively.

Each muscle group was assumed to behave like a single muscle with a point-like origin and insertion and a continuous line of action. In addition, the muscles were assumed to behave according to an adapted form of Hill's equation

$$(F+a)(v+b) = (gF_0+a)b \quad (1)$$

with thermodynamic constants $a=F_0/4$ and $b=v_0/4$. F_0 is the maximal active isometric force at optimal length, and v_0 is the maximal velocity of

muscle shortening. 'g' represents the maximal force a muscle can exert as a function of its instantaneous length and normalized to F_0 . It is expressed as

$$g = -6.25 (l/l_0)^2 + 12.5 (l/l_0) - 5.25 \quad (2)$$

l and l_0 denote the instantaneous length and optimal length of the muscle, respectively. l_0 was defined as the arithmetic average of the longest and shortest muscle length within anatomically reasonable joint configurations of the lower limb. l was defined in two ways, 1) as the line integral of the muscle line of action between attachment points, and 2) as the average muscle fiber length.

The average power of the lower limb was defined as the sum of the work done by the seven muscle groups in one full crank rotation divided by the time for one full crank cycle. It was assumed that a muscle group was either fully active or completely silent for the phases of shortening and lengthening, respectively.

RESULTS AND DISCUSSION For a fixed crank length of $r=17$ cm, the maximal value $\bar{P}_M$ of the average power $\bar{P}$ of one lower limb, and the optimal values of the design parameters for the above two definitions of muscle lengths were 1) $\bar{P}_M=636$ W, $\alpha=2^\circ$, $h=55$ cm, and $\dot{\theta}=177$ rpm, and 2) $\bar{P}_M=509$ W, $\alpha=9^\circ$, $h=64$ cm, and $\dot{\theta}=147$ rpm, respectively. The value of $\bar{P}_M$ for definition 2) was 20% smaller than that for definition 1). This is the case because the muscle lengths during one crank revolution are farther away for longer periods of time from the region of near optimal muscle length for definition 2) than definition 1).

Figure 3 shows the average $\bar{P}$ normalized to $\bar{P}_M$ as a function of α , h , and $\dot{\theta}$. The value '0' (zero) on the horizontal axes of Figures 3a, 3b, and 3c designates the optimal value of the corresponding independent variable for each $\bar{P}$ with the two different definitions of muscle lengths. Figure 3a and 3b show that $\bar{P}$ was much more sensitive to changes in α and h for definition 2) than definition 1). This is due to the fact that the same amount of change in α and h means a much greater change of $1/l_0$ in eq. (2) for definition 2) than definition 1) in each position of the crank, and hence a greater change in force the muscles can exert. This is not true for changes in $\dot{\theta}$ (Figure 3c).

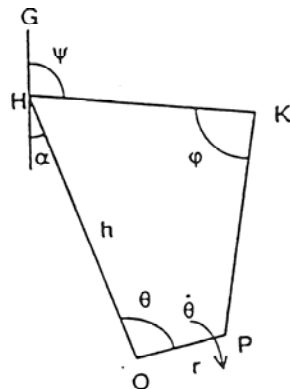


Figure 1: Mechanical model of the bike-rider system

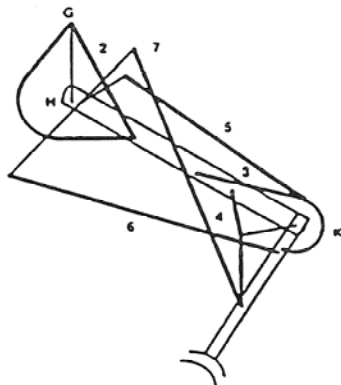


Figure 2: Musculoskeletal model of an average lower limb and pelvis of the rider.

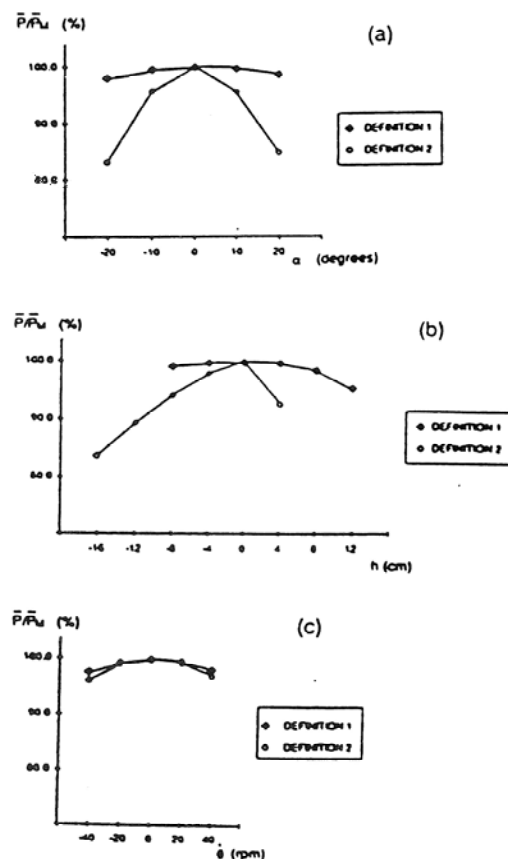


Figure 3: Change of $\bar{P}$ normalized to $\bar{P}_M$ as a function of one design variable ($\alpha, h, \dot{\theta}$) while the other design variables were kept at their optimal values. For these considerations, r was kept constant at 17 cm.

THE DYNAMIC MODEL OF SKELETAL MUSCLE:
I. RESPONSE OF NINE DIFFERENT MUSCLES

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INTRODUCTION

The dynamic model of skeletal muscles is controversial and not fully documented. Furthermore, since different muscles have different architectural and dynamic properties (twitch duration, pennation, weight, tendon length, etc.) It is conceivable that the dynamic model of various muscles is different, or muscle specific.

The objective of this study was to define the dynamic model of nine different muscles in the hind limb of the cat and to relate the resulting model with the muscles weight, pennation, twitch duration (1,2), and tendon length (3).

METHODS

Sinusoidal excitation of each muscle was accomplished with the newly developed electrical nerve stimulation system described in detail elsewhere (4,5). This valuable research tool allow the orderly recruitment of motor units according to the "size principle" (6) while simultaneously increasing the action potentials firing rate of the recruited motor units in a similar mode to the motor control during voluntary contraction. The motor units were recruited sinusoidally as well as the corresponding firing rate. Sinusoidal frequencies in the range of .4 Hz and up to 6 Hz were applied. Isometric muscle force was recorded with a Grass FT-10 transducer. The distal tendon of each muscle was released and was connected to the transducer with an aluminum chuck. The Fast Fourier Transform (FFT) of the force and sinusoidal stimulus input were derived at each stimulation frequency and the gain and phase were calculated and normalized with respect to their value at .4 Hz. The Bode plot for each muscle was constructed from the data and the "best fit" model identified using the method of least squares ($P < 0.05$). At the end of each of the six experiments, each muscle was subject to a train of 10 pulses at a rate of .5/sec. The twitch response recorded and the average duration required for the twitch to reach its peak calculated. Furthermore, the weight, length, pennation and tendon length of each of the nine muscles was recorded and averaged for the 4-6 preparations tested.

RESULTS

The "best fit" model for all nine muscles consisted of a second order linear low pass filter with double poles (both poles at the same frequency). The location of the poles, however, varied widely from muscle to muscle. Ranging from 1.55 Hz to 2.8 Hz. Similarly, the time delay associated with the model of each model varied widely, and ranged from 8 msec to 17 msec. The correlation coefficient for each model was relatively high, ranging from .912 to .981 ($P < 0.05$).

A table summarizing the results of the study, including the nine muscles, their poles, time delay (Td), twitch duration (Tw), mass, tendon length to muscle length ration and pennation was constructed.

CONCLUSIONS

The most important conclusions of the study is the fact that the dynamic response of various muscles varies widely and probably depends on various architectural and functional properties which are specific to each muscle. Large, slow twitch muscles, like the soleus and gastrocnemius reflect their properties in a low pole value as well as long time delay. The gastrocnemius muscles, although classified as fast twitch muscles, exhibited lower pole values than other fast twitch muscles (tibialis), reflecting their larger mass. The tibialis posterior, a fast twitch, small and long muscle exhibited the highest pole value reflecting its small size and fast force generation properties. The direct effect of tendon could not be clearly identified from the experimental protocol and may require a different study method. Similarly, pennation mode did not seem to be a clearly identifiable factor in the muscles dynamic model. Separation of the muscles into functional groups (i.e. ankle flexors and extensors) show that most extensors have a low pole value while flexors had much higher pole value. The possibility of the muscles functional role as a component of its dynamic model is considered.

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THE DYNAMIC MODEL OF SKELETAL MUSCLE:
II. DYNAMICS OF THE TENDON AND JOINT

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INTRODUCTION

The dynamic response of a skeletal muscle is the manifestation of its twitch characteristics, mass, fiber structure and tendon length among other factors. Furthermore, the viscoelastic properties of the joint tissue such as ligaments, articular surface and tendons of inactive muscle may also contribute to the formation of its force transmission dynamic properties.

The objectives of this study are to identify the contributions of the tendons viscoelastic stiffness and the joints soft tissue characteristics to the muscles dynamic response model.

METHODS

Four adult cats were used. The M. gastrocnemius muscle nerve was exposed and instrumented with an electrode system. The stimulations scheme which allows orderly recruitment of motor units according to their size simultaneously with firing rate increase is described in detail elsewhere (1,2). Sinusoidal motor unit recruitment and firing rate increase were applied to obtain sinusoidal force oscillations. Initially, the ankle joint was fixed isometrically at 90° flexion, and joint force was measured 1.5 inches below the joint. Sinusoidal force traces in the range of .4 Hz to 6 Hz were recorded. Next, the muscle tendon was severed at its insertion and connected to a Grass FT-10 force transducer. The sinusoidal force traces described above were recorded. Finally, the tendon was shortened by 1 inch and connected to the force transducer before repeating this test. The individual data points were used to construct the Bode plots for each condition including the model gain and phase. The data from the joint, whole muscle and muscle with shortened tendon were compared using a two tailed T-test ($P < 0.05$) to identify any significant differences.

RESULTS

The dynamic model of the M. gastrocnemius muscle including its tendon consisted of a linear second order system with double poles at 2 Hz and a time delay of 11 seconds. The T-test failed to demonstrate any statistical significance between the muscle model and that of the muscle without its tendon. This points out that the viscoelastic stiffness of the tendon is probably very high and not influenced by the range of forces up to 75% of the tetanic force employed in this study.

The dynamic model of the muscle acting through the joint was fitted with a linear third order system with poles near 1.5 Hz. The T-test comparing the whole muscle (including tendon) and the muscle model when acting through the ankle joint demonstrated significant differences ($P < 0.05$) for the

frequency range from 1 Hz and up to 6 Hz. The phase of the muscle model acting through the joint was not shown to be significantly different from the whole muscle model.

CONCLUSIONS

The major conclusions of this study demonstrate that the tendon length is not a major contributor to the muscles dynamic model under isometric conditions and when subjected to force oscillations in the range of 25%-75% of the maximal tetanic force. This may seem as a contradiction to the tendon studies of Proske and Morgan (3) who demonstrated that tendon viscoelastic stiffness increases as tendon length decreases. One should keep in mind, however, that the range of forces required to elicit tendon viscoelastic properties may be higher than the mid-range of muscle force generation capabilities employed in this isometric study. It is conceivable that the tendons viscoelastic properties are much more pronounced in strong contractions as well as in eccentric contractions. For the conditions employed in this study, the tendon acted as a stiff force transmission component.

The linear third order model obtained for the muscle acting through the joint is the direct manifestation of the additional viscoelastic properties of the various soft tissue associated with the joint, ligaments, articular cartilage, tendons of inactive agonist and antagonist muscles, interstitial fluids trapped between various membranes, etc.

Investigators should employ muscle models cautiously when modelling human limb motion. The muscles dynamic model may be subjected to significant modifications when its action is applied through the joint vs its action at its tendon. It is also conceivable that the joint dynamic further modified when considering the limb mass and inertia in non-isometric conditions.

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MUSCLE STIMULATION POWERED ORTHOSIS:
A PRACTICAL WALKING SYSTEM FOR PARAPLEGICS

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INTRODUCTION

Spinal cord injury devoids the lower motor units from receiving centrally mediated voluntary motor control and deem the patient as paraplegic, being unable to stand-up and walk. As a wheel chair bound patient, such individuals are unable to perform many daily functions with further complications due to decreased cardio-pulmonary condition, increased spasticity, osteoporosis and increased episodes of urinary and bowel problems. Attempts to restore simple walking functions to such patients require analysis of joint and muscle biomechanics and electrophysiology associated with locomotion.

Conceptually, the following functions are required for the rehabilitation of such patients:

- A. Antigravity support to the hip, knee and ankle joint to allow upright posture.
- B. Control of balance and stability for safety (prevention of a fall).
- C. Performance of the swing phase of one leg.
- D. Simultaneous push-off of the contralateral leg.

This study describes the development of a muscle stimulation powered body orthosis capable of providing the above functions along with performance evaluation of six paraplegics with injuries to the C-6 to T-10 spinal segments.

METHODS

The orthosis

An orthotic system was designed, developed and evaluated on six patients. The orthosis consists of an ankle-foot plastic splint, a lockable knee joint and a free hip joint. Lateral aluminum uprights connect the ankle-foot splint with the knee and hip, as well as extend upward to the mid-thoracic level. The two hip joints were connected with Bowden cables such that hip flexion of one leg will produce hip extension of the lateral leg. When hip flexion or extension are not attempted, the hip joints provide stable anti-gravity support with only 5° hip motion to allow the patient to balance himself by shifting his upper trunk and shoulders. A spring loaded lock was installed in the posterior aspect of the knee joints such that when the patient rises from a chair, the knee joints will lock when fully extended.

The orthosis provides the patients with full anti-gravity support to the ankle, knee and hip. It can, with proper individual alignment support up to 250 lbs. dead weight in upright posture without external interference.

The orthosis developed can, therefore, provide anti-gravity support, balance and stability, and reciprocal contralateral push-off upon hip flexion.

MUSCLE STIMULATION

The power to generate hip flexion and contralateral hip extension was derived from surface electrical stimulation of the quadriceps and hamstrings muscles. Gelled, conductive rubber electrodes were placed 2 inches above and below the respective muscles and wired to a four channel electrical stimulator. Activation of muscular contraction was provided by pressing a switch attached to the handle bars of a rolling walker. Pressing the left switch contracted the left quadriceps simultaneously with the right hamstrings. Since the R. Femoris crosses also the hip, with the knee fixed, all its force is dedicated to hip flexion. Similarly, two branches of the hamstrings cross the hip joint, and their stimulated force generates hip extension.

The electrical muscle stimulation provided the necessary biomechanical functions of swing phase in one leg and contralateral push-off in the other leg.

RESULTS

SYSTEM PERFORMANCE

1. Upright posture from sitting - five out of the six patients can rise from sitting by pressing with arms on the handle bars of the rolling walker. The sixth patient has weak arms (a C-6 level)
2. Sitting from upright posture - all six patients can perform this function.
3. Balance and stability - all six patients can maintain at least 15 seconds of balanced upright posture with arms raised to their sides or over their head. All patients can maintain satisfactory balance during locomotion.
4. Distance travelled - this varied among the patients. And ranged from .27 miles and up to 1.24 miles of continued walking.
5. Top locomotion velocity - ranged from 31.1 ft/min and up to 68.68 ft/min.
6. Cadence (steps per minute) at leisurely rate - 41 steps/min to 50 steps/min.
7. Range of motion about the hip with stimulation - ranged from 32° to 49.6° .
8. Average increase in thigh circumference after six week (3 sessions per week) of stimulation (8 inches above lateral condyle) - 5% or 1 inch, due to reversal of atrophy of disuse of muscles.

CONCLUSIONS

The combination of a biomechanically designed mechanical with electrical stimulation of the thigh muscles provides a practical solution for the rehabilitation of simple gait functions in paraplegics.

NEUROMUSCULAR ADAPTATIONS AND HORMONE BALANCE IN STRENGTH ATHLETES, PHYSICALLY ACTIVE MALES AND FEMALES DURING INTENSIVE STRENGTH TRAINING

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INTRODUCTION. The early increases in muscle strength during intensive strength training especially among initially untrained subjects may be accounted for largely by neural factors with a gradually increasing contribution of muscle hypertrophy as the training proceeds (Moritani and DeVries 1979). In strength athletes the magnitudes and the time courses of neuromuscular adaptations during their training may differ from those of reported for previously untrained subjects (Häkkinen et al. 1985, 1987). The sexual differences are very apparent in the magnitude of muscle mass and strength but less information is available on specific neuromuscular adaptations and their time courses during strength training in females. The purpose of the present study was to investigate training-induced neuromuscular adaptations and hormone balance in strength athletes, physically active males and females during intensive strength training.

METHODS. 7 male strength athletes (MSA), 9 physically active males (MPA) and 10 physically active females (FPA) volunteered as subjects for the study. The subjects were tested repeatedly at 3,5 week intervals before, during and after the 10,5 week experimental period. Bilateral isometric leg extension force-time curve and maximal force were measured on a electromechanical dynamometer. Electromyographic activity was recorded telemetrically by surface electrodes from the superficial leg extensor muscles and integrated to obtain IEMG-time curve of the contraction (Häkkinen et al. 1985). The cross-sectional area of quadriceps femoris muscle was measured by means of an ultrasonic apparatus. Blood samples were drawn for the determination of serum concentrations of total and free testosterone, cortisol, luteinizing hormone (LH) and sex hormone-binding globulin (SHBG). The subjects participated in supervised heavy resistance strength training (with the loads between 70 and 100 % of the maximum) 3,5 times a week (every second day).

RESULTS. Significant changes occurred primarily in the higher force portions of the force-time curves with increases of 7.7 % ($p < 0.05$), 13.4 % ($p < 0.01$) and 19.6 % ($p < 0.05$) in maximal force in MSA, MPA and FPA, respectively. Significant ($p < 0.05-0.01$) increases took place in the voluntary neural activation of the trained muscles in MPA and FPA but not in MSA. Enlargements of 7.8 % ($p < 0.05$), 13.6 % ($p < 0.01$) and 11.3

* ($p < 0.01$) occurred in muscle cross-sectional area in MSA, MPA and FPA, respectively. MSA demonstrated greater ($p < 0.01$) maximal force per muscle cross-sectional area than MPA and FPA both before and after the training period. Minor changes were observed in the concentrations of serum hormones and SHBG examined. The individual mean serum levels of testosterone in FPA correlated ($r = .76$; $p < 0.05$) with the individual changes in maximal force.

DISCUSSION. The present findings that significant changes occurred primarily in the higher and maximal force portions of the force-time curves support well the concept of the specificity of heavy resistance strength training. The results further indicate that this specificity may be rather independent on the pretrained status of the subjects as well as on the sex. The findings both in male and female subjects further indicate the important roles of neural and hypertrophic adaptations for strength development (see also Moritani and DeVries 1979, Häkkinen et al. 1985). Strength development during the present short term strength training among the males and females was rather similar in magnitude but the females demonstrated greater interindividual variation especially during the later weeks of the training period. The adaptations in the neuromuscular system become limited in magnitude in strength athletes who already have an intensive training background of several years. Although this limitation may be apparent especially with regard to the nervous system, the present athletes could still demonstrate greater maximal force per cross-sectional area than the other two groups also after the experimental training period. The present results additionally suggest that strength training may not have major effects on serum levels of endogenous anabolic and catabolic hormones in strength athletes, adult males or females. Great interindividual differences especially in females in serum testosterone levels may indicate differences in the trainability.

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A CONSTITUTIVE EQUATION FOR THE HUMAN DURA MATER ESSENTIALI

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The quasi-linear theory which describes the viscoelastic properties of the general biotic soft-tissue was proposed by Feng in 1972. The primary assumption upon which this theory is based is that the following relationship was founded when a step-strain acts on the specimen.

$$\sigma(t, \epsilon) = G(t) \cdot T^e(\epsilon) \dots (1)$$

where: $\sigma(t, \epsilon)$ denotes the stress response

$T^e(\epsilon)$ denotes the 'elastic response'

$G(t)$ denotes the normalized relaxation function

When a continuous variable strain acts on the specimen, the stress response is

$$\sigma(t) = \int_0^t G(t-\tau) \frac{\partial T^e[\epsilon(\tau)]}{\partial \tau} d\tau \dots (2)$$

Many experimental results of some biotic soft-tissue show that the normalized relaxation function is dependent on not only the relaxation history, but also the strain level. To constant strain level, the normalized relaxation function is still the linear function of $\ln t$. So, we applied the generalized form $G(t, \epsilon)$ of the normalized relaxation function $G(t)$ which was proposed by Feng. When a step-strain acts on the specimen, we can similarly conclude:

$$\sigma(t, \epsilon) = G(t, \epsilon) \cdot T^e(\epsilon) \dots (3)$$

When a continuous variable strain acts on the specimen, the stress response is:

(1) Only considering the change of the elastic response resulted by $\delta \epsilon(\tau)$, the stress response is calculated, that is,

$$\sigma(t, \epsilon) = \int_0^t G[t-\tau, \epsilon(t-\tau)] \frac{\partial T^e[\epsilon(\tau)]}{\partial \tau} d\tau \dots (4)$$

(2) Considering not only the change of the elastic response but also the change of the normalized relaxation function, which is resulted by $\delta \epsilon(\tau)$, the stress response is calculated, that is,

$$\sigma(t, \epsilon) = \int_0^t G[t-\tau, \epsilon(t-\tau)] \frac{\partial T^e[\epsilon(\tau)]}{\partial \tau} d\tau + \int_0^t T^e[\epsilon(t-\tau)] \frac{\partial G[t-\tau, \epsilon(t-\tau)]}{\partial \tau} d\tau \dots (5-1)$$

If $t=0$, $\epsilon(0)=0$, we obtain the expression of elastic response

$$\sigma(t, \epsilon) = 2 \int_0^t G[t-\tau, \epsilon(t-\tau)] \frac{\partial T^e[\epsilon(\tau)]}{\partial \tau} d\tau - T^e[\epsilon(t)] \dots (5-2)$$

We call the formula (4) the A type expression of the generalized quasi-linear theory, in short, generalized A type, and the formula (5) the B type expression, in short, generalized B type. Sometime generalized A and B type can describe more accurately viscoelastic properties of some biotic soft-tissue.

In order to verify the accuracy of generalized A and B type expression, we completed a series of the relaxation test on the Autograph testing machine in tension. The fresh dura mater specimens were obtained from 20 grown-up subjects (10 men, 10 women). During the test, we keep mechanical properties of the human dura mater no changing. The experiment results show that to constant strain level, the normalized relaxation function is still linear function of $\ln t$. So, we suppose

$$G(t, \epsilon) = (a_0 + a_1 \epsilon) + (b_0 + b_1 \epsilon) \ln t \dots (6)$$

In order to determine the parameters in the formula (6), a simple linear regression analysis was performed. The values of these parameters which calculated from the experiment datum is respectively

$$\begin{array}{ll} a_0=0.6476 & a_1=2.4001 \\ b_0=-0.05300 & b_1=0.3675 \end{array}$$

$$(10^{-3}\text{min} \leq t \leq 10\text{min})$$

We suppose the elastic response

$$\tau^e(\epsilon) = C\epsilon + d\epsilon^2 \dots (7)$$

where: $C=11.23\text{MPa}$ $d=632.8\text{MPa}$

The value of the parameter C and d was calculated from the experiment datum at a strain rate of 0.1sec^{-1} .

The stress-strain curves at variable strain rates can be calculated from generalized A and B type expression. When the strain rate has increased by 10^3 times, but the stress has increased no more than twice. Obviously, generalized A and B type expression are not rate sensitive. This point is just consistant with the viscoelastic properties of biotic soft-tissue. At a strain rate of 0.001sec^{-1} and 0.05sec^{-1} respectively, the deviation of the value of any point on the testing stress-strain curve compared to value of the corresponding point on generalized B type curve were within 10 per cent, and compared to corresponding value of generalized A type curve and Feng's quasi-linear theory curve were within 25 and 20 per cent respectively. Obviously generalized B type expression of Feng's quasi-linear theory can describe more accurately viscoelastic properties of human dura mater. But then viscoelastic properties of medial collateral ligament of the rabbit knee were described on the use of generalized A type expression, its result is more accurate than that on the use of the Feng's quasi-linear theory expression.

In summary, the Feng's quasi-linear theory is a basic theory which conveniently describes viscoelastic properties of such biotic soft-tissue that the changes of the strain level have no effect on normalized relaxation function or the effects should be neglected. But then to some biotic soft-tissue, the changes of the strain level have effect more obvious on normalized relaxation function, thus generalized A and B type expression can describe more accurately the viscoelastic properties of these biotic soft-tissue. We should pay much attention to this point.

PROPRIOCEPTIVE CONTROL OF MUSCLE CONTRACTION UNDER
CHANGING LOAD - MODEL INVESTIGATIONS

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The particularity of impulse flow transformation in autogenous control systems and mechanisms of motoneuron (MN) activity organization have been studied on the neuron (N) network model including electron analogues of MN, afferent N, exciting and inhibiting interneurons (IN), and analogues of muscle and its receptors as well. Characteristics of change potential on the outputs of model elements were obtained after having solved the differential equations describing the mechanisms of their functioning. Studying the muscle spindle and Golgi tendon organ (TO) properties we consider the muscle model characteristics as properties of muscle fibres of one motor unit. It allows us to standardize influence and to exclude uncertainty connected with interactions of the structural relations of different muscle fibres with receptors. In experiments with TO the different patterns of activity depending on muscle contraction were singled out. Changing the muscle length it is possible to create such conditions for TO when in response to each muscle twitch one discharge in afferent 1b will be recorded. During muscle stretch from this position we can watch the proportional increase of discharges in the 1b fiber in response to every contraction. Passive muscle can be stretched under the load lifting it during active contraction. In response to a single contraction the same number of pulses in afferent 1b is recorded at the larger length of muscle passively stretched by the load than if the muscle contracting isometrically. Limiting the stretching of the passive muscle by a stop taking the weight of load connected to the muscle we make it possible to shorten actively. Under these conditions the number of discharges in afferent 1b does not depend on the weight of the load, being only defined by the initial muscle length. Increasing stimulation up to tetanic contraction we obtain the growing of proportionally developed tension in a muscle to the impulse flow intensity in afferent 1b. Simulating the work of a spindle in a muscle we have reproduced the location of sensitive endings of Ia and II groups in different parts of intrafusal fibres. Studying receptor potential formation we have found that the different parts of spindle receptive regions have different transmission coefficients (K) varying with the change of sinusoidal stretch frequency. The relation of the receptor potential amplitude to the spindle stretching one is taken as K. Taking the input amplitude for 1, K in the primary endings where a receptor potential proportional to the second derivative of changing length is increasing from 0.6 (stimulation about 0.4Hz) up to 4.0 at the stimulation more 20Hz, while in the secondary endings with a receptor potential proportional to spindle length K (equal to 1 at the input less than 0.4Hz) is decreasing with the increase of stimulation and practically becoming 0 at the frequency over 10Hz. There were found non-linear changes of K in parts of a spindle capsule being very sensitive to the velocity of input signal alteration. The 1-st derivative from the capsule length has a maximum K of 2 within the frequency of 1Hz becoming less at the change of stretch frequency both with less and greater values. The force coming

through the intrafusal muscle fibre to the spindle central formations has minimum K of 0.6 at the input frequency of 1Hz. K is increased both with decreasing of stimulation frequency coming up to 2.5 (less 0.4Hz) and increasing of stimulation frequency approaching to 3.7 at 5Hz and more. The decrease as well as increase of spindle length change correlation of elastic and viscous properties of intrafusal fibres that result in change of K in different parts of spindle receptive zones. The passive spindle can't be the regulator of MN activity as the servo-mechanism. The gamma-control indeterminates the response of spindle receptor formations concerning the muscle length changing their adaptive properties. The most adaptive time has been shown at gamma-stimulation small frequencies while the growing of pulse frequency leading to the fast increase of intrafusal fibre tension, is decreasing the adaptive time. It allowed us to suggest that alpha-gamma-coactivation function was that the maintenance of sensitivity in the spindle receptor apparatus at the high level tuning its rigidity to the rigidity of contracting muscle provides the adequacy of reflex responses intended for the compensation of external perturbations intervened in the sequential movement performance. Thus muscle proprioceptive afferentation can't simply describe the state of motor system but obviously provides its dynamic stability promoting fulfilment of coordinate movements according to homeostasis in the locked structure of control. The closing of proprioceptive loops leads to the transition from isometric contraction to isotonic one at the constant descending level of MN excitation connected with the decreasing of TO activity, following the increasing activity of Renshaw cell (RC) related to MN discharge, and evidently being a compensator of the decreasing activity of IN in 1b afferent chain. Simultaneously the muscle contraction frequency and number of impulses (per 1 discharge of TO) generated by IN are increased. IN is offered as an element with variable K controlled by afferent flow intensity. K is increased with the decrease of TO activity. The stopping of recurrent inhibition via RC enlarges the IN activity in 1b chain. Thus the circuit of recurrent inhibition along the motor axon collateral and the chain of annular inhibition through 1b afferents work synchronously and restrict MN discharges during switching off one of the chains. IN as RC influences more strongly on MN non-involved in autogenic inhibition that may serve as the mechanism of motor-contrast promoting the selective control of MN activity in the pool. The excitation of MN along 1a fibers is accompanied by higher rhythmic activity of MN providing possible shortening of muscle fibres and decrease of 1a flow or facilitation of their work if the excitation is spreading to MN contracting other muscle fibres. Stronger activity of spindles non-connected with the motor unit in the closed system regulation provides the load uniform distribution on the whole muscle contrary to motor-contrast. The carried out work main result consists in the fact that the system of numerous feed-backs via peripheral sensory ways providing many-sided influence on MN has a stable character in the structure of peripheral annular control independent a kind of autogenic influence. It may be considered as mechanism of homeostatic regulation where afferentation has not an original controlling significance at the spinal level but promotes the programme movements in non-stationary environment.

A STUDY OF STABILITY OF STANDING SHOOTING

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Shooting performance is closely related to shooting stability in the events of fixed target. In Free Rifle 3X40 shots competitions, scores of standing shots is usually the determinant in shooters' final success. In order to improve the standing shots technique, we study shooter's stability of standing shots by sports biomechanics.

Subjects: Shooters including nine men of national team(6 International Master Sportsmen, 3 Master Sportsmen), seven shooters of Beijing Shooting School(2 First-class Sportsmen, 3 Second-class Sportsmen, 2 Third-class Sportsmen) and seven normal men were measured at our laboratory. Ninety-two males(41 Master Sportsmen, 51 First-class to Third-class Sportsmen) were tested at the sports field.

Experimental instruments contain a Selspot II system, a Kistler force platform, a 1PL 16 mm high-speed camera, GP-2000 film analysis system and a WJGT-30 graphic analysis system etc.

Methods: While shooters were accurately aiming to fire in the standing shots, data of kinetics and kinematics were synchro-measured by the Selspot II system connected with a Kistler force platform. Sampling frequency of system was 50 Hz, sampling time was 12 seconds. Subjects natural stability were measured by a platform. Photographs and pictures were taken and standing position of the feet were also measured in sports field. Measurement data of the Selspot-Kistler system were transformed and filtered with a PDP 11/23 computer of system, then final data and graphics could be obtained.

Results: By using the Selspot-Kistler system, the displacement and the velocity of the muzzle and the center of rifle's gravity in

three dimensional directions, angles formed the longitudinal rifle and three coordinate planes while shooters were shooting, the center of pressure while shooters were natural standing and shooting were collected. Composite centers of shooter's gravity and rifle's angles of stability of shooter and rifle system by photograph, areas of shooter's standing position and other by measuring were also obtained. Through in four seconds shooting stable phase, data, graphics and tables of the muzzle and rifle's angular displacement and above other data were analysed, compared. The following preliminary results indicated that:

a. Multi-indexes measuring and analysing of the Selspot-Kistler system could evaluate shooter's stability in standing shooting. It was found that rifles were displaced along the direction of target and there were some problems in shooters' techniques. Testing results could reflected shooters' performance in some way, therefore it could predict how shooters shall perform in the near future competition.

b. Score of standing shooting were related to displacement of the muzzle, which were more closely related to rifle's angular displacement. In three kinds rocks of rifle, most of shooters showed a positive angular rock of rifle.

c. In four seconds shooting stable phase, both elite and preliminary shooters had the largest variation of displacement of their muzzle in the Y direction and the variations of X and Z displacement were close. There was the largest angular variation between rifles and the XOZ plane.

d. There was significantly difference of stability of rifle muzzle between elite shooters and preliminary shooters ($p < 0.05$), when shooting in standing stance.

A MUSCLE MODEL RELATING HIP ABDUCTOR TORQUE AND SURFACE EMG

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INTRODUCTION

A method for clinical interpretation of computer averaged EMG profiles [1] can be based on the idea that muscle activation is responsible for muscle force, and that muscle force is a clinically relevant variable. This concept is applied to the hip abductors. Abductor torque is predicted from surface EMG using a muscle model, and the model can assist in the interpretation of EMG profiles.

THE HIP ABDUCTOR MODEL

A hip abductor model was developed [2] which relates surface EMG to external abductor torque during level walking using the M. Gluteus medius as a muscle equivalent for the abductor group, assuming isometric contraction. The model can be implemented as a recursive digital filter, with transfer function

$$H(z) = T(z)/E(z)$$

$$= K [(z+1)(z+1)] / [z(1+t_1') + (1-t_1')] [z(1+t_2') + (1-t_2')]$$

where $T(k)$ is the data series which represents the abductor torque signal, and $E(k)$ is the series describing the "intensity" of the surface EMG of the M. Gluteus medius, expressed in the full wave rectified and low-pass filtered EMG (3rd order Butterworth filter, - 3dB at 25 Hz +/- 5 %). These series are obtained by sampling and A/D conversion of the continuous time signals at 200 Hz. A calibration recording of EMG data $E(k)$ and torque data $T(k)$ during dynamic isometric contraction [3] is required to identify the parameters of the model.

HIP ABDUCTOR ACTIVATION AND TORQUE DEVELOPMENT

In order to investigate the effect of varying step frequency on the observed muscle activation patterns, the activation of the M. Gluteus medius was studied in eight normal subjects (age 22 - 30 years, one female, seven male) at step frequencies of 60, 90 120 steps/minute with a constant stride length of 1.2 metres. Results are summarized in figure 1, where the ensemble averages of the recorded activation patterns are shown. In the graph 0 and 100 % gait cycle correspond to the moment of heel-on, with toe-off around 60 %.

From these activation patterns the time course of the abductor torque is estimated using the muscle model with $t_1' = 17$ and $t_2' = 17$, which are representative values found in three subjects. These values correspond to time constants of approx. 40 msec for a proposed continuous-time second order muscle model [4]. Because a representative value for the parameter K cannot be defined for these averaged patterns, only the relative time course of the abductor torque for the various step frequencies can be determined. Results are indicated in figure 2.

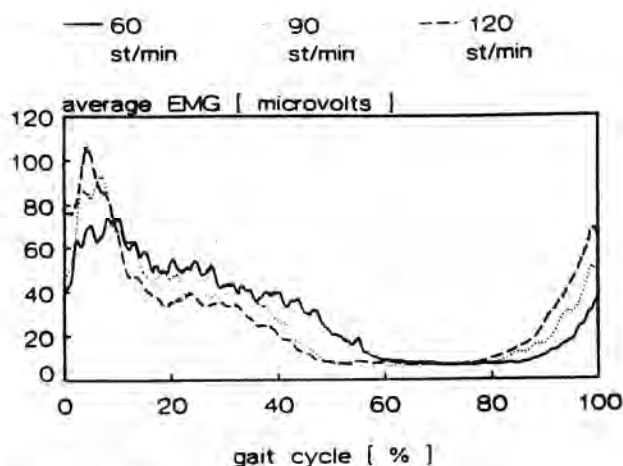


Fig. 1: Activation of M. Gluteus medius depending on step frequency.

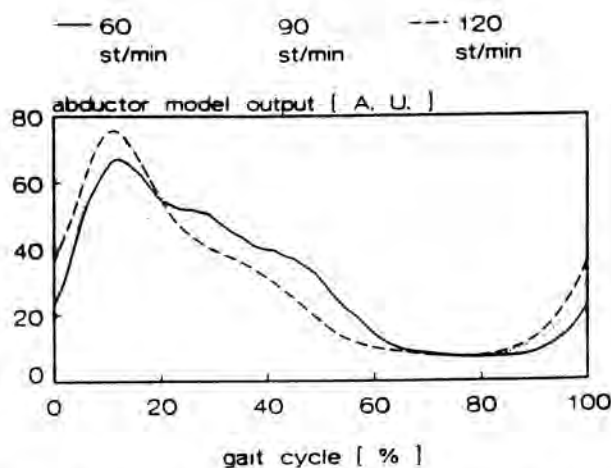


Fig. 2: Estimated time course of abductor torque, depending on step frequency.

CONCLUSIONS

The muscle model quantitatively accounts for the observed activation patterns to occur progressively early with increasing step frequency, where the generated torque remains relatively independent of this frequency. The maximum of the torque occurs at 10 - 15 % gait cycle which coincides with the end of the double support phase. This is biomechanically satisfying.

Also, the results indicate that effects of muscle dynamics have to be taken into account when clinical interpretation of averaged EMG profiles is attempted via their relation with muscle force. A quasi-static linear relation between intensity of EMG and muscle force can only be applied at step frequencies far lower than normal.

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Effect of Foot Orthotics on the Kinematics of the Knee Joint

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Introduction

Patellofemoral Pain Syndrome (PFPS) is the leading cause of chronic knee pain in adolescents and is associated with excessive subtalar pronation (1). Clinicians have employed the use of foot orthotics as an adjunct to the conservative management of PFPS in order to reduce the pronation at the subtalar joint and in turn possibly decrease the transverse and frontal motion occurring at the knee. This study employed the use of an optoelectronic system to examine the effects of foot orthotics on the three-dimensional (3D) kinematics of the knee.

Review and Theory

Clinicians have reported success in alleviating lower extremity problems with the use of foot orthotics (2)(3). Two-dimensional (2D) data is not sufficient to describe the motion of a joint which possesses 6 degrees of freedom. One investigation used a triplanar electrogoniometer but reported no significant effects of foot orthotics on the knee joint (4). A need for calculating the representative or average values for several gait cycles has been reported if subtle differences in lower extremity function are to be detected (5). A standardization of the speed of locomotion, age and gender of subject are also necessary as these variables have been shown to affect gait parameters (6)(7).

Methodology

Seven female adolescents (see Table 1 for subject description) were assessed and diagnosed with PFPS by the same physician. Ten degrees of forefoot varus and/or subtalar varus as assessed by the same physiotherapist was required for inclusion of the study. Subjects did not display a leg length discrepancy greater than 1 cm or possess any known pathological or neurological disorder which could affect their gait pattern. The same physiotherapist fitted each subject with a soft orthotic constructed from a flat neoprene insole and posted with medial rearfoot and forefoot wedges.

Data was collected by a 4 camera optoelectronic system (WATSMART) at 75 Hz under 2 conditions whose presentations were randomized: running shoe and flat neoprene insole, running shoe and orthotic. All subjects wore the same running shoe (Reebok CL006). A foot switch was applied to the heel of the shoe to record consecutive heel strikes. Five 3 cm infrared light emitting diodes (IREDs) were placed on the thigh, 5 on the lower leg and 6 on the foot. Two 20 s trials of walking at 3.2 km/hr and running at 6.4 km/hr on a treadmill were recorded. A test-retest was performed with 5 subjects to define the reproducibility of the measures across time. Cycles were averaged across the 2 trials to allow inclusion of at least 30 gait cycles. Using a sagittal view of the 2D data the gait cycle was divided into phases of contact, midstance, and propulsion as defined in Table 2.

Three-dimensional body segment positional data were obtained from multiple sets of 2D data using a rigid body tracking method. The reference position was defined when the subject stood relaxed with the feet parallel and directed forward. Mathematical transformations were employed to rotate and translate the segment around 3 axes to make its locally defined IRED coordinates coincide with the coordinates observed by the cameras. Software determined the orientation of the proximal end of the distal segment relative to a vertically aligned proximal segment. Analysis of the rotations in each plane was performed by a repeated design with a factorial layout incorporating the effect of orthotics, mode of ambulation (walking and running), and phase of the gait cycle.

Results

In the sagittal plane, the orthotics did not produce any significant difference in any of the parameters. As significant effects and interactions were found in the frontal and transverse plane, post-hoc methods (Tukey's Test) were used to further analyse the motion in these planes. Mean values, standard deviations and statistical analysis are shown in Table 3. Figure 1 demonstrates some of the effects of foot orthotics in one individual. In walking, significant differences were found in the transverse and frontal planes during the contact phase and in the frontal plane during the midstance phase. No significant differences were found in the propulsion phase. In running, significant differences were only found in the propulsion phase in the frontal and transverse planes.

Discussion

Foot orthotics significantly affect the kinematics of the knee in the frontal and transverse planes. The effect of the orthotics differs in walking and running. This phenomena may be due to the change in the duration and mechanics of each phase from walking to running or due to the orthotic compressing at various phases of the gait cycle. It is possible that previous investigators did not detect any significant effect from the use of foot orthotics at the knee joint as it is not only necessary to use precision

equipment, but also to average over many cycles to detect subtle differences. Differences are not always found over the total stance period but may be found when the stance period is divided into subphases of contact, midstance and propulsion. One explanation for the success of foot orthotics in alleviating PFPS may be related to the alteration of motion at the tibiofemoral joint which in turn affects the patellofemoral contact and joint reaction force.

Conclusions

Foot orthotics alter the motion of the knee in the transverse and frontal planes, but the effect differs in walking and running. These findings demonstrate the need: (a) for 3D kinematics for analysing joint motion, (b) for concurrent kinetic analysis using foot orthotics, (c) to further divide the stance period into subphases, and (d) for further research to compare the biomechanical results achieved with orthotics to the clinical effectiveness of foot orthotics in the patient with PFPS.

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Table 1 - Subject Description

Height(cm) - 160.6(9.8)
Mass(kg) - 52.9(11.8)
Age(years) - 13.9(1.1)
Forefoot Varus(degrees) - 12.0(3.5)
Subtalar Varus(degrees) - 6.9(5.1)

Table 2 - Range of Motion (ROM) Definitions

Stance - Max ROM from heel strike to toe-off
Contact - Max ROM from heel strike to foot flat
Midstance - Max ROM from foot flat to heel off
Propulsion - Max ROM from heel off to toe-off

Table 3 - Mean Values and Statistical Analyses for Kinematic Variables of the Knee Without and With Orthotics (in degrees), $p^* < .05$, standard deviation in parentheses

	Frontal Plane		Transverse Plane		Sagittal Plane	
	Without	With	Without	With	Without	With
Walking						
Stance	8.0 (3.2)	7.5 (2.8)	12.7 (5.5)	12.3 (5.6)	25.4 (5.5)	26.2 (6.2)
Contact	3.5 (1.8)	2.9 * (1.4)	4.9 (2.4)	3.1 * (1.0)	12.3 (3.9)	11.7 (4.0)
Midstance	3.5 (1.2)	3.2 * (1.0)	5.4 (2.9)	5.1 (3.3)	10.7 (6.2)	11.6 (7.1)
Propulsion	4.7 (2.1)	4.9 (2.0)	5.2 (2.6)	4.4 (3.0)	20.6 (5.4)	20.0 (6.7)
Running						
Stance	9.6 (4.8)	8.7 (4.4)	7.9 (5.1)	6.9 (2.3)	31.8 (2.7)	32.4 (4.6)
Contact	4.4 (3.6)	5.0 (3.0)	5.6 (3.4)	5.1 (2.5)	25.9 (6.3)	25.7 (6.4)
Midstance	3.7 (2.0)	4.2 (1.7)	3.2 (1.3)	3.4 (1.2)	9.4 (3.8)	9.9 (3.3)
Propulsion	5.5 (2.6)	4.5 * (3.1)	4.3 (2.1)	3.7 * (2.4)	18.8 (4.5)	17.9 (4.6)

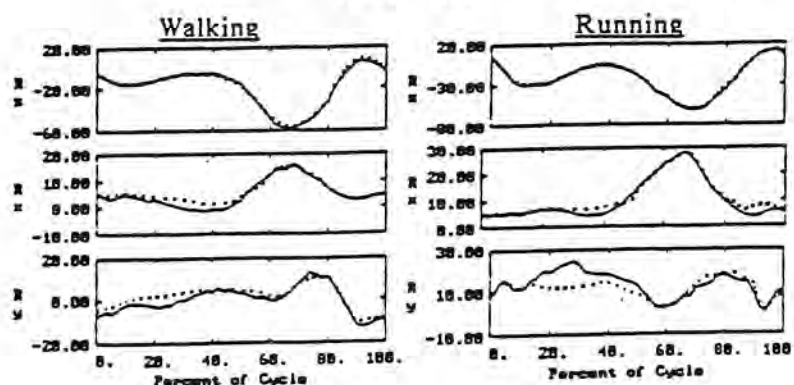


Figure 1 - Rotations of the Left Lower Leg in Respect to the Femur

without orthotics

with orthotics

Z : + flexion, - extension
X : + abduction, - adduction
Y : + external rotation
- internal rotation

MODELS INCORPORATING HEIGHT, DISTANCE AND LANDING TECHNIQUE TO PREDICT IMPACT FORCES

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Jumping is a component of many movement activities. All jumping activities include a landing phase during which the performer may be exposed to high and potentially injurious forces. Valiant and Cavanaugh (1985) reported landing forces during basketball rebounds of up to 6.0 bodyweights while Panzer (1987) reported forces during double back somersault landings as great as 15.0 bodyweights. The purpose of the study was to evaluate the effects of height (H), distance (D), and technique (T) on impact forces during landings.

METHODOLOGY

Three semi-skilled male volunteer subjects performed three trials for each of three Hs (40, 60, 100 cm), Ds (40, 70, 100 cm) and landing Ts (stiff, slightly flexed, fully flexed knees) for a total of 81 trials. High speed film records of the right lower extremity (100 fps) were obtained during landing onto a force platform (500 Hz). All ground reaction force (GRF) data were normalized to body mass for comparative purposes. Film data were interactively smoothed and interpolated to temporally match the GRF records.

Temporal and kinetic data describing the maximum impact force values and lower extremity joint position and velocity data were used in the analysis. Between and within-subject 3-way analyses of variance (ANOVAs) were computed for first (F1) and second (F2) maximum vertical forces. Three regression models (mechanical, biomechanical, refined biomechanical) were computed to predict F1 and F2 for all subjects. The mechanical regression model incorporated the independent variables of H and D while the biomechanical model added the independent variable of T. The refined biomechanical model substituted a set of selected kinematic variables (ankle, knee and hip joint position and velocity at contact, F1 and F2) for T.

RESULTS AND DISCUSSION

Results of the group ANOVAs indicated significant ($p < 0.05$) main effects for H, D and T and H and T for F1 and F2, respectively. The within-subject ANOVA model for S1 most closely matched the group model exhibiting significant H and T main effects for both F1 and F2. Significant main effects were observed for S2 (H and T) and S3 (H, D and T) but these effects were confounded by interactions for both F1 and F2.

Results of the three regression models (mechanical, biomechanical, refined biomechanical) for F1 and F2 for all subjects are given in Table 1. The mechanical model for F1 accounted for 56.7, 82.5 and 38.1% of the variance for the three subjects, respectively. H was the only significant variable to enter all models with D contributing no significant improvement. The addition of T into the model improved the prediction for S1 (24.3%) but had no significant effect for S2 and S3. The refined biomechanical model did not improve S1's model, provided a nominal improvement (0.9%) for S2 and greatly improved the predictions for S3 (42.8%).

H was the dominant mechanical variable for all F2 subject models although the total explained variance was considerably less than for the F1 models accounting for an average of 19.0% of the variance compared to 59.1% (Table 1). The biomechanical model results for all subjects included T as the first variable to enter accounting for an average of 39.1% of the variance. The refined biomechanical model was a slightly better predictor of F2 than the mechanical model for S1 (37.7 versus 30.6%) but inferior to the biomechanical model (72.0%). The refined biomechanical model was the best F2 predictor for S2 and S3 with 81.3 and 88.0% explained variance, respectively.

The collapsed ANOVA statistical analysis was not an accurate representation of individual subject performances. All three subjects exhibited unique ANOVA results different from the group model. The within-subject ANOVA analyses produced significant interactions indicating the complex nature of the performance strategies used by the subjects. Individual subject regression models computed on F1 and F2 also produced unique models for all subjects. The biomechanical model was the best predictor of both F1 and F2 for S1. The refined biomechanical model was the best predictor of F1 for S2 and S3 with H and ankle joint kinematic descriptors being the most prevalent independent variables. The refined biomechanical model was also the best predictor of F2 for S2 and S3. H and knee joint related variables were most prevalent in the prediction equations for all subjects. The observed use of individual performance strategies by subjects suggests that group data analysis results can lead to inappropriate conclusions. The investigation of movement phenomena using a within-subject design limits one's ability to generalize results. However, if the experimental manipulation produces multiple individual performance strategies or a continuum of strategies then group generalizability is not only inappropriate but misleading. Therefore, it is important to determine the individual strategies and attempt to identify common components which may be generalizable across individuals.

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Table 1. Prediction of First and Second Maximum Force from Mechanical, Biomechanical and Refined Biomechanical Models

	Subject 1				Subject 2				Subject 3			
	F1	F2	F1	F2	F1	F2	F1	F2	F1	F2	F1	F2
<u>Mechanical</u>	H	56.7*	H	30.6*	H	82.5*	H	18.1*	H	38.1*	H	8.2*
	D	57.6	D	6.2	D	82.9	D	19.7	D	40.5	D	10.1
<u>Biomechanical</u>	H	56.7*	T	32.7*	H	82.5*	T	35.5*	H	38.1*	T	49.0*
	T	81.0*	H	72.0*	D	82.9	H	58.0*	D	40.5	H	57.1*
	D	81.7	D	72.8	T	83.3	D	60.4*	T	40.5	D	59.1
<u>Refined Biomechanical</u>	H	56.7*	H	30.1*	H	82.5*	H	36.6*	H	36.4*	D	51.5*
			AF2V	37.7*	ACF1V	83.4*	KF2P	68.9*	ACP	62.8*	H	67.0*
							ACF2V	76.2*	ACV	73.8*	KF2P	73.6*
							KCV	78.1*	AF1P	77.8*	AF2P	84.8*
							KF1P	79.9*	HF1V	80.9*	KF2V	86.6*
							AF2V	81.3*			KF1P	88.0*

Note: numeric value is percent explained variance; * = $p < 0.05$; H = height; D = distance; T = technique; A = ankle; K = knee; H = hip; C = contact; F1 = 1st max force; F2 = 2nd max force; CFI = avg from C to F1; CF2 = avg from C to F2; P = position; V = velocity.

BIOMECHANICAL CORRELATES OF FAST BOWLING AND BACK INJURIES IN CRICKET: A PROSPECTIVE STUDY

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The increased participation of children in organised sport and fitness activities has resulted in an increase in the number of injuries (Michell, 1983). In fact an epidemic of back injuries has been reported for fast bowlers in cricket. A biomechanical evaluation of the fast bowling motion was therefore undertaken in an endeavor to identify the relationship between bowling technique, the forces associated with each delivery and back injuries in cricket.

Eighty-two high performance young male fast bowlers (mean age 16.8 years) were filmed both laterally (200Hz) and from above (100Hz) immediately prior to the season while bowling such that their front foot impacted a force platform during the delivery stride. The resulting kinetic data were synchronised with kinematic data obtained from film. The players then completed a log book over the ensuing season, that detailed their training and playing programs. All cricket related injuries over this season were assessed by a sports physician, who used computerised tomography to assist in the diagnosis of spinal injuries. At the completion of this season the players were grouped according to their injury status (Group 1-bony injury to the vertebrae); Group 2- soft tissue injuries to the back that caused the player to miss at least one match and Group 3-no injuries). A one-way analysis of variance was used to identify if any biomechanical variables significantly differentiated between the three groups and a Scheffe post-hoc comparison was used to determine which groups were significantly different.

Eleven percent of the players sustained a stress fracture (1 at L4; 7 at L5; 1 at an incompletely sacralised S1) while 27% sustained soft tissue injuries to the back. Four of the 9 players, who sustained stress fractures were found to be genetically predisposed to pars defects. However, the other 5 players developed pars fractures probably as the result of the repetitive stress of bowling. The incidence of stress fractures from this study is, however, less than the 50% reported by Payne et al. (1987) for a group of elite bowlers evaluated over 5 years.

No significant differences were recorded between the groups for peak vertical ground reaction force (GRF) or peak horizontal GRF at front foot impact (FFI). Although the mean peak values of 5.4BW (vertical) and - 2.7BW (horizontal) show that large GRFs must be absorbed by the body at each FFI during the delivery stride. Nineteen of a sample of 32 players (59%), who bowled in excess of the mean number of matches for the group, suffered a stress fracture or soft tissue injury to the back compared to the 38% injury frequency for the total group. Twenty-seven bowlers, of a sample of 45 players who bowled in excess of 60 deliveries

In a single spell, reported back pain the following day. Bowling too often and/or bowling too many deliveries may therefore be a contributing cause of injury.

No significant difference was recorded between the groups for shoulder alignment at back foot impact (BFI). However, when two groups of bowlers were selected from the total number based on shoulder alignment at BFI, 9 bowlers were classified as side-on ($<190^\circ$) and 15 were considered to be excessively front-on ($>240^\circ$). Of the side-on bowlers only 1 sustained a back injury while 8 of the 15 excessively front-on bowlers sustained back injuries. These results support the data of Elliott and Foster (1984), where the front-on bowling action was identified as a possible cause of back injuries. A significant difference was recorded between Injury Group 1 and Group 3 for shoulder rotation between BFI and the position of maximum rotation, with Group 1 rotating more than Group 3. Bowlers who excessively rotated to re-align the shoulders between BFI and FFI were more likely to sustain back injuries.

The release height of the ball significantly differentiated between Group 1 and the other two groups. The angle at the knee and hip joints of the front limb were larger for the injured group. It is possible that bowlers who flexed more at the front knee and hip joints to absorb the force at impact were less likely to be injured than those bowlers who delivered the ball over a more extended front limb. No significant differences were recorded between the groups on any other biomechanical variables.

In conclusion over-use injuries such as stress fractures of the lower back are probably caused from a combination of factors. Bowling with specific characteristics for extended periods has therefore been linked to back injuries. However, the physiological state of the bowler is another factor that must be considered. Good technique and appropriate practise/match loads are certainly two aspects of bowling that must be monitored if sporting injuries are to be kept to a minimum.

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A Research for the Innovatory Movement of
Straight Tkatchev in Horizontal Bar

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It is the life of gymnastics to bring constantly forth new ideas in the movement. In this paper innovational movement designed by authors, that is, the Straight Tkatchev Vault (STV) is researched into.

Method

In our research the STV is assumed to be composed by four phases, that is, swing with whipping movement of legs (SWML), release of grip, vault and regrasp. Based on the mechanical principles, the optimal conditions of swing down in the SWML, the release and the regrasp are deduced, and a computer program calculating the optimal values is designed. Zhao Zhiqiang, a member of Jiangsu gymnastic team, is chosen as the subject of the study. Since he began to learn Tkatchev in 1984, we had helped him constantly to diagnose and improve the technique through film analysis and computer simulating based on biomechanical method. As the result of our research and guidance, he succeeded in finishing the STV in 1987.

Results and Discussion

The SWML is a phase of movement in which gymnasts could get more kinetic energy through the movement of legs. So our discussion will first concentrate on the gymnasts could take more effect of gravity moment. Based on the mechanical principles, the relation between the angle of crooked hip joint and the gravity moment is given. Using extreme method, we obtained the optimal condition of the swing down. It means that the sum of the angle of upbody swing down and the angle of crooked hip joint must be kept 90° , which is called "cover" SWML. This technique is used in Zhao Zhiqiang's training. As a result, his initial velocity of release increased from 3.5m/s to 4.14m/s, which laid the foundation of his finishing the new movement.

The release grip is a key technique in the STV. A good release grip is the guarantee of the body flying with optimal condition. According to our research, the "No Gravity State", that is 41.8° on the bar, is the optimal position to release. The gymnast was asked to do so in our experimental training. It caused his initial angular velocity to increase from 2.11rad/s to 2.58rad/s .

Condition of initial vault includes velocity and angle of initial vault. It decides the flight trace of the body center of gravity(COG) after releasing. The optimal condition of initial vault is the important guarantee for finishing the STV and regrasping at optimal position. The results of our research and computer simulating showed that Zhao's optimal velocity and angle of initial vault is 4.18m/s and 56° respectively in the STV.

Condition of regrasp includes the radius of regrasp(the length of the tie between the COG and horizontal bar) and the angle of regrasp(the angle included between the tie and the level line). The trace of the COG is a parabola in vault and a circle in giant swing. The optimal condition of regrasp asks to regrasp at the moment when the two curvilinear motions are coincidence and the same direction. The results of our research and computer simulating showed that Zhao's optimal angle and radius of regrasp is 32.2° and 0.91m respectively. Through the experimental training the gymnast has skillfully performed the STV now and his regrasp condition has come close to above optimal values.

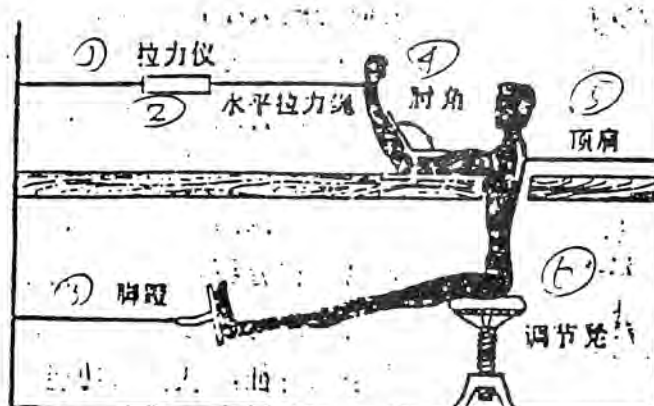
Conclusions

1. It is possible that gymnasts bring to success the new movement of the STV. The technique of the "cover" SWML is useful to fulfilling the STV.
2. The values of the optimal conditions of release and regrasp in the movement for Zhao's STV are obtained, such as the angle of release 41.8° ; the velocity and angle of initial vault 4.18m/s and 56° ; the angle and radius of regrasp 32.2° and 0.91m .

AN EXPERIMENTAL STUDY ON THE DYNAMIC AND STATIC HORIZONTAL FORCE OF ELBOW JOINT

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The present artical makes dynamic and static test and analysis of elbow's horizontal pulling force of elbow bending and stretching at different angle positions, from the general need of research on the movement techniques of human body, so as to provide basis for research on functional anatomy, biodynamics and sport skills. Its objects are 93 sportsmen of baseball and throwing, jogging and jumping events. Apparatus used for test is Model YL-82A piezoelectrical pulling force meter developed by Dalian Polytechnology University (testing range, 0-300 kgf; linearity $< 1\%$; resolving power, 0.5 kgf; inherent frequency, $< 9\text{KHz}$; Weight, 50g). It displays the value of static horizontal force through piezoelectric digital meter and scanscrables curves of dynamic horizontal force through light-ray recorder. Tested were 19 groups of data of dynamic and static forces of elbow bending and stretching at five angles, 30°, 60°, 90°, 120° and 150°. At the same time, scan-scrabbling were made of the electromyograms of humeral biceps and triceps by means of Model 272 multi-purpose remote measuring meter made by the Japaness Sanrong Tester Co. Ltd.. Test was conducted on the "multi-purpose force measuring bed". (See Fig.)



- | | |
|---------------------|----------------------------------|
| 1. pullmeter | 2. rope of horizontal pull force |
| 3. pedal | 4. elbow angle |
| 5. shoulder jack-up | 6. adjustable stool |

Results are as follows,

Comparison was made among bending and stretching dynamic and static forces of elbow joints of different sportsmen with different sexes and, the trends of change in elbow bending and stretching dynamic and static horizontal forces at different angle positions were obtained; Individual difference was distincted; Analysis was made between horizontal force values and relative conditions of the entire indicators of human body shape; Comparison was made in dynamic and static horizontal force indicators between group of athletes of throwing event and group of athletes of jogging and jumping events; Observations were made to characteristics of electromyograms with elbow joint at different angle positions and in different force values obtained with different ways.

With elbow bended at 130°, its static horizontal force is at its maximum. The angle 130° may be used as angle position for judging bending elbow static capacity. With elbow bended at 90°, its dynamic horizontal force is at its maximum. This angle may be used as the angle position for judging dynamic ability.

With elbow stretched at 150°, its static horizontal force is at its maximum. The angle may be used as the angle position for judging the static ability of the stretching elbow.

With elbow stretching at 120°-150°, its dynamic horizontal force is at its maximum. The maximum value of the stretching elbow's dynamic horizontal force of high-level sportsman is concentrated on 120° angle position, so it is proper to use 120° angle position when the dynamic ability of the stretching elbow is judged.

Understanding the trends of change in the dynamic and static horizontal forces of elbow bending and stretching at different angle positions, may be helpful for the sportsman to select the optimum angle for exerting his force, to heighten physical ability and to realize the optimum sport techniques.

BIOMECHANICS OF THE REARFOOT DURING WALKING IN HIGH AND LOW FAT FEMALES.

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The purpose of this study was to examine differences between high and low fat groups of females on selected anthropometric, foot angle, and rearfoot movement variables during the stance phase of walking. Twenty-six females between the ages of 25 and 50 years ($M = 36.7 \pm 6.7$ yrs.) volunteered as subjects for this investigation. The subjects were separated into a low fat group for subjects with less than 26% body fat ($n = 13$) and a high fat group for subjects with greater than 26% body fat ($n = 13$). The mean percentage of body fat for the high and low fat groups was 30.22% and 22.33%, respectively. The anthropometric evaluation included the following measurements: weight, height, sum of three skinfolds, Q angle, ankle joint range of motion, arch index, and wrist, thigh, and calf circumferences that were used to calculate limb mass and limb volume. In addition, foot angles, angle of gait, and stride width were calculated from foot prints obtained when subjects walked along a walkway with inked adhesive felt attached to their shoes' outsoles. After the subjects were habituated to treadmill walking, high speed photography (100 Hz) was performed during the last 15 s of a 10 min treadmill walk at $1.34 \text{ m}\cdot\text{s}^{-1}$. Film records were digitized and the data smoothed using a fourth order low-pass digital filter with a cutoff frequency of 10 Hz. The calculated rearfoot kinematic parameters included touchdown angle, maximum pronation angle, total rearfoot movement, maximum pronation velocity, and time to maximum pronation. Three trials per subject were analyzed and later averaged to yield representative values for each subject.

Two-way ANOVAs (group x leg) repeated across one factor (legs) were used to compare rearfoot movement variables. There were no significant differences in overall main effects; however, there was a significant ($p < .05$) group by leg interaction. More specifically, a simple effects procedure revealed that maximum pronation velocity between the right ($-78.02 \pm 9.5 \text{ deg}\cdot\text{s}^{-1}$) and left ($-51.3 \pm 9.6 \text{ deg}\cdot\text{s}^{-1}$) legs of the high fat group were significantly different; however, this asymmetry was not apparent in the low fat group. Although there were no significant main effects, the high

fat group showed trends toward greater total rearfoot movement ($M_{HF} = -7.32 \pm 1.17$ deg; $M_{LF} = -5.98 \pm 1.19$ deg), greater maximum pronation velocity ($M_{HF} = -78.02 \pm 9.5$ deg·s⁻¹; $M_{LF} = -52.35 \pm 9.67$ deg·s⁻¹), and a shorter time to maximum pronation ($M_{HF} = .224 \pm .02$ s; $M_{LF} = .286 \pm .03$ s). In addition, the high fat group exhibited tendencies toward greater ($p > .05$) foot angles for both the right ($M_{HF} = 5.15 \pm 1.28$ deg; $M_{LF} = 3.26 \pm 1.06$ deg) and left ($M_{HF} = 3.14 \pm 1.11$ deg; $M_{LF} = 1.90 \pm 1.21$ deg) feet and a greater angle of gait ($M_{HF} = 1.23 \pm .14$ deg; $M_{LF} = 0.68 \pm .14$ deg). Interestingly, both groups showed greater foot angles for the right foot, suggesting that each foot may have a separate function during gait. Correlations between selected anthropometric and rearfoot movement variables revealed significant ($p < .01$) but moderate correlations between limb mass and foot angle ($r = .50$) and between % body fat and total rearfoot movement ($r = .53$). The results of this study revealed asymmetry in rearfoot movement among high fat females that was not apparent in a similar group of low fat females. Moreover, the tendency of the high fat group to exhibit greater rearfoot movement and larger foot angle values than the subjects in the low fat group may have implications for injury prevention and footwear design.

ACOUSTIC EMISSION DURING COMPRESSION, FROM NORMAL AND DISEASED BONE

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INTRODUCTION

The study of acoustic emission utilizes sonic signals emitted by a material when excited by physical stimuli. During mechanical loading energy is released as a result of dynamic processes such as microcracks and dislocations. The released elastic energy generates acoustic waves within the material. Recording of the acoustic signals emitted by materials under loading has been used for the examination of structural disorders in fiber reinforced composites. In the present study this technique was used for examining trabecular bone tissue from normal subjects and from patients with metabolic bone disorders. It was anticipated that this examination would provide an insight into the mechanisms responsible for the mechanical behaviour of trabecular bone under load.

MATERIAL AND METHODS

Trabecular bone specimens were taken from 31 femoral heads which were excised from osteoporotic patients during replacement of the femoral head following subcapital fractures and from 6 femoral heads of osteoarthritic patients during total hip replacement. These were compared to specimens taken from 13 femoral heads which were obtained at autopsy from subjects under the age of 50, free of known bone disease. The cylindrical specimens were taken from the trabecular region under the cortical layer of the femoral head so that their axis of symmetry coincided with that of the femoral neck. Each cylinder was of 10 mm height and 10 mm diameter. The specimens were kept at a temperature of -18° C during the period of measurements and thawed before loading at room temperature.

Each specimen was compressed along its axis at a constant speed of 1 mm/min, using a J.J. Lloyd M30K Material Testing Machine, until a maximum was obtained on the load-deflection curve. The acoustic signals emitted from the specimen during compression were recorded by a transducer attached to an aluminium clipper embracing the specimen. The signals were analyzed by an Acoustic Emission Testing System (AET 5000).

On each load-deflection curve the yield point and the maximum load point were identified. The ultimate compressive stress (UCS) and the elastic modulus in the pre-yield zone (EM) were calculated. The energy absorbed by the specimen up to the point of maximum load (AE) was calculated as the area under the load-deflection curve. The average rates of acoustic events up to the yield point (RY) and beyond it (RB) were calculated as the ratio between the number of events and the respective time interval. The average peak amplitude (AP) of the acoustic signals was also recorded. The values of these parameters in the

normal, osteoporotic and osteoarthritic subjects were compared and the significance of the differences was assessed by Mann-Whitney U Test.

RESULTS

Table 1 presents the average value and standard deviation of each parameter in the three groups of subjects.

	TABLE 1		
	Normals n=13	Osteoporotics n=31	Osteoarthritics n=6
UCS (MN/m ²)	11.5 ± 3.9	7.6 ± 4.5	9.7 ± 1.6
EM (MN/m ²)	157.1 ± 38.4	116.5 ± 50.0	116.1 ± 17.2
AE (Jx10 ⁻³)	0.867 ± 0.576	0.363 ± 0.299	0.677 ± 0.170
RY (sec ⁻¹)	1.6 ± 1.2	2.9 ± 2.1	3.9 ± 1.0
RB (sec ⁻¹)	4.2 ± 2.1	7.5 ± 4.8	8.1 ± 3.0
PA (dB)	44.4 ± 1.2	47.0 ± 3.4	46.3 ± 3.2

Table 2 presents, for each parameter, the ratios between the average value in the diseased and the normal bones.

	Osteoporotics/Normals	Osteoarthritics/Normals
UCS	0.66	0.84
EM	0.74	0.74
AE	0.42	0.78
RY	1.81	2.44
RB	1.79	1.93
AP	1.06	1.04

It was found that UCS and EM were significantly lower in both diseased groups while RY, RB and AP were significantly higher than in the controls. The AE was lower in both diseased groups, however between OA patients and normals the difference was not statistically significant.

DISCUSSION

The results of the mechanical behaviour under compression showed that trabecular bone tissue of both OP and OA patients was weaker, less stiff and less ductile than of normal subjects. The higher rate of acoustic events both before and beyond yield in both OP and OA patients may reflect the greater extent of trabecular failure during the loading process. The higher peak amplitude of signals in the diseased bones may be attributed to either the higher energy of the acoustic events or to a lower attenuation of the signals in the diseased bones.

THE MEASUREMENT AND ANALYSIS OF BODY SHAKING STATES

(Abstracts)

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When in vertical position, the body keeps shaking, which is in the category of the random vibration system. Varying from person to person, the shaking exerts a great influence to events which need good balance, especially those in gymnastics, shooting, and figure skating. The measurement and analysis of shaking states are, without doubt, of great significance for the selection and training of athletes. This paper introduces a new method for measuring body shaking coefficients and studies the data obtained in the experiments.

The Method

The shaking movement was measured with an X-Y axis crystal rate gyroscope installed on the helmet. The data obtained were processed, displayed, and printed by the computer. After health inquiries and vestibular examination, 30 college students in an age range from 18 to 20 were chosen as the subjects in the normal group, while 20 shooting athletes were taken as the subjects in the contrast group. Thus, the shaking coefficients were obtained. In the light of theories on random vibration and spectral analysis, those data were counted and statistically analyzed, and through the FFT (the fast Fourier transform) the properties of the frequency spectrum were observed.

The Result and the Discussion

The instrument we have made can measure body movements to get data concerning the angular displacement, angular velocity, and angular acceleration of body shaking, that is, shaking backward and forward, and shaking leftward and rightward. The curve showing the coefficients of body shaking is a random vibration curve. In our experiments, the sampling frequency was 10 Hz while the time history (T) was 60 seconds. The experiment are common in medical examination. Listed in table 1 are the statistical data about the 30 subjects in the normal group.

Table 1

	left-right rad		for-backward rad		left-right rad/s		for-backward rad/s		left-right rad/s/s		for-backward rad/s/s	
	R	L	F	B	R	L	F	B	R	L	F	B
1	.66±.16	.61±.22	.72±.16	.72±.22	.55±.16	.55±.16	.61±.16	.66±.22	.22±.05	.16±.05	.27±.05	.27±.05
2	.72±.16	.66±.22	.83±.22	.77±.27	.55±.22	.61±.16	.66±.16	.77±.27	.22±.05	.22±.05	.27±.05	.27±.05
3	1.05±.33	1.16±.44	1.27±.33	1.16±.33	.83±.22	.83±.27	1.11±.38	1.11±.38	.38±.11	.38±.11	.44±.11	.44±.11
4	1.38±.38	1.55±.50	1.72±.38	1.55±.44	1.11±.33	1.22±.38	1.50±.38	1.50±.44	.61±.16	.55±.16	.66±.11	.61±.16
5	.94±.27	1.16±.44	1.27±.33	1.11±.44	.83±.22	.88±.27	1.16±.33	1.05±.38	.44±.11	.38±.11	.44±.61	.44±.16
6	1.33±.38	1.38±.38	1.77±.55	1.41±.50	1.11±.27	1.22±.33	1.50±.44	1.38±.44	.61±.16	.55±.16	.72±.22	.66±.22

- 1) and 2) Romberg's experiment: eyes open, eyes closed.
 3) and 4) man's experiment: left foot forward, eyes open;
 left foot forward, eyes closed.
 5) and 6) man's experiment: right foot forward, eyes open;
 left foot forward, eyes closed.

Conclusion

1. By the way introduced in this paper we can easily measure the coefficients concerning body shaking states in order to study and evaluate the stableness of the body.
2. This method is of great significance for the selection of athletes who should be good in balance and stableness, especially those in shooting, gymnastics, and figure skating.
3. In the study of stableness and nimbleness, it is inevitable that the angular displacement, angular velocity, and angular acceleration should be under consideration.

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Best Instant for Take-off from Springboard

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We take the springboard for an equivalent harmonic oscillator with mass m_2 , supported by a spring with spring constant K , as shown in Fig. 1.

Measuring the period of the empty springboard

$$T_1 = 2\pi\sqrt{m_2/K}$$

and the period with a person of mass m_1 at standstill on the springboard

$$T_2 = 2\pi\sqrt{(m_1 + m_2)/K}$$

we can get parameters

$$K = 4\pi^2 m_1 / (T_2^2 - T_1^2),$$

$$m_2 = T_1^2 m_1 / (T_2^2 - T_1^2).$$

Let y_2 be the ordinate of the board, and the origin $y_2 = 0$ correspond to the equilibrium position. Let the mass of the diver be m_1 , and y_1 is the coordinate of diver of center of gravity, and let the origin $y_1 = 0$ correspond to diver at standstill on the stationary springboard.

From Newton's second law of motion we have the equation of motion

$$\begin{cases} m_1 \ddot{y}_1 = -m_1 g + N & (1) \end{cases}$$

$$\begin{cases} m_2 \ddot{y}_2 = -Ky_2 - N & (2) \end{cases}$$

By actual measurement and calculation, we obtained equivalent mass $m_2 = 130$ kg, $K = 13428.6$ N/m.

Suppose the equation (1) and equation (2) have an analytical solution.

The initial conditions are

$$\text{Before stamping, } y_1 = y_2, \quad (3)$$

$$\dot{y}_1 = \dot{y}_2. \quad (4)$$

The stamping ends at the instant

$$t = t_1 + \pi/2\omega_1,$$

Then, the diver leaves the springboard. And at that instant,

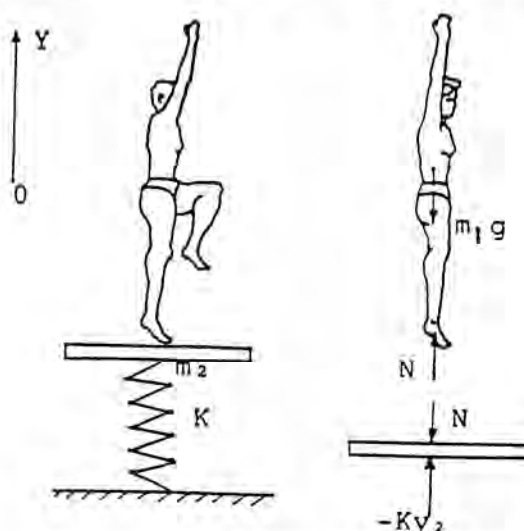


Fig. 1. The force diagram for Diver and springboard.

$$y_2(t) = g/K + A \cos \sqrt{K/(m_1+m_2)}(t_1 + \pi/2\omega_1) - \\ Lm_1\omega_1^2/[(m_1+m_2)\omega_1^2-K] \cos[\sqrt{K/(m_1+m_2)} \pi/2\omega_1] \quad (5)$$

$$v_2(t) = -A\sqrt{K/(m_1+m_2)} \sin \sqrt{K/(m_1+m_2)}(t_1 + \pi/2\omega_1) \\ - \frac{Lm_1\omega_1^2}{[(m_1+m_2)\omega_1^2-K]} \left[\omega_1 - \sqrt{\frac{K}{m_1+m_2}} \sin\left(\sqrt{\frac{K}{m_1+m_2}} * \frac{\pi}{2\omega_1}\right) \right] \quad (6)$$

$$y_1 = y_2 + L \quad (7)$$

$$v_1 = v_2 + L\omega_1 \quad (8)$$

We can calculate the energy of the diver and springboard, which are designated as E_1 and E_2 respectively:

$$\text{Energy of the diver} \quad E_1 = m_1 g y_1 + \frac{1}{2} m_1 v_1^2.$$

$$\text{Energy of the springboard} \quad E_2 = \frac{1}{2} m_2 v_2^2 + \frac{1}{2} K y_2^2.$$

$$\text{The total energy of the system is} \\ E = E_1 + E_2 \quad (9)$$

The E_1/m_1g is the greatest height H which the diver is able to reach.

$$H = E_1/m_1g \quad (10)$$

We assigned various values to A , L , m_1 , and used a computer to find the best instant t_1 for the beginning of stamping, optimizing energy E_1 obtained by diver, or equivalently optimizing the height reached by diver.

By calculation, the rule appears that the best position for stamping is a little bit above the lowest position when the springboard goes upward, i.e., $180^\circ - 212^\circ$ (the lowest position is 180°).

The Mechanics of Take-off from Springboard

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We take the springboard for an equivalent harmonic oscillator with mass m_2 , supported by a spring of which the spring constant is K ^[1].

Because in equation (1) and (2) have been $\dot{y}_1/dt$, $\dot{y}_2/dt$, and N three unknown numeric.⁽¹⁾ We did not strightly solving it. In order to calculate energy changes, we multiply equation (1), with dy_1 , equation (2) with dy_2 , and add up the two results:

$$d(\frac{1}{2}m_1\dot{y}_1^2 + \frac{1}{2}m_2\dot{y}_2^2 + \frac{1}{2}Ky_2^2 + m_1gy_1) = Nd(y_1 - y_2) \quad (3)$$

In view of Koenig's theorem, let

$$\begin{cases} m_0 = m_1 + m_2 & \text{be total mass,} \\ m' = m_1 m_2 / (m_1 + m_2) & \text{be reduced mass, and} \\ \dot{y}_0 = (m_1 \dot{y}_1 + m_2 \dot{y}_2) / (m_1 + m_2) & \text{be velocity of center of mass,} \end{cases} \quad (4)$$

then, equation (3) may be rewritten as

$$d[\frac{1}{2}m'(\dot{y}_1 - \dot{y}_2)^2 + \frac{1}{2}m_0\dot{y}_0^2 + \frac{1}{2}Ky_2^2 + m_1gy_1] = Nd(y_1 - y_2) \quad (5)$$

here, $\frac{1}{2}m'(\dot{y}_1 - \dot{y}_2)^2$ is relative kinetic energy of system relative to the center of mass. $Nd(y_1 - y_2)$ is work done by reaction force N at distance $d(y_1 - y_2)$.

Futhermore, adding equation (1) to (2), we get the equation of motion of the center of mass:

$$m_0 d\dot{y}_0/dt = -m_1g - Ky_2$$

Multiply each term with dy_0 , we get

$$d(\frac{1}{2}m_0\dot{y}_0^2 + m_1gy_1 + \frac{1}{2}Ky_2^2) = -Ky_2 d(y_0 - y_2) \quad (6)$$

Substitute equation (6) into equation (5), we get:

$$d\left[\frac{1}{2}m'(y_1-y_2)^2\right]-Ky_2d(y_0-y_2)=Nd(y_1-y_2) \quad (7)$$

When diver stamps from the beginning to the end on the springboard, then

$$d(y_1-y_2)>0.$$

The work of internal forces for diver

$$Nd(y_1-y_2)>0.$$

The mechanical energy increase in formula (7) comes from muscles. To be more concrete, let us suppose the stamping action is finished instantaneously. Though this suppose is not realistic, it enables us to analyse quantitatively immediately before stamping, the common velocity of the diver and the board is v_0 , and immediately after stamping the

velocity of the diver and that of the board are designated as v_1 and v_2

respectively. According to the law of conservation of momentum and the law of conservation of energy, we get:

$$v_1=v_0+\sqrt{(m_2/m_1)[2E/(m_1+m_2)]} \quad (8)$$

$$v_2=v_0-\sqrt{(m_1/m_2)[2E/(m_1+m_2)]} \quad (9)$$

Then, the changes of kinetic energy of the diver and of the springboard will be (E is energy of which released muscles of diver):

$$\frac{1}{2}m_1v_1^2-\frac{1}{2}m_1v_0^2=E\frac{m_2}{(m_1+m_2)}+v_0\sqrt{2E\frac{m_1m_2}{(m_1+m_2)}} \quad (10)$$

$$\text{and } \frac{1}{2}m_2v_2^2-\frac{1}{2}m_2v_0^2=E\frac{m_1}{(m_1+m_2)}-v_0\sqrt{2E\frac{m_1m_2}{(m_1+m_2)}} \quad (11)$$

Formula (10) points out: If $v_0=0$, so the kinetic energy increased is only

$E\frac{m_2}{(m_1+m_2)}$. If v_0 is minus, the kinetic energy increased is less than

$E\frac{m_2}{(m_1+m_2)}$. In order to get much energy, diver stamping should take

place when the springboard is slightly above the lowest position and is moving upward.

Reference:

- [1] Liu Tao and Wu Xiufang "Best Instant for Take-off from Springboard"
XII ISB June 26-30, 1989 UCLA USA.

Biomechanical Research for the Technique of
"Swing with Whipping Movement of Legs" in Horizontal Bar
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"Swing with Whipping Movement of Legs" (SWML) is a basic and key technique in sport on horizontal bar. In this paper, the rational movement of legs in the SWML is determined in theory, through filming the SWML performed by some elite gymnasts and studying the data of film analysis. The characteristics of SWML forward techniques in different movement on horizontal bar is investigated.

SWML could be divided into two phases, swing down and whipping movement of legs. Swing down is the phases in which the body swing down acceleratively from the position of handstand on the bar. How the body could get more gravitative moment is a key point of technique of swing down. In order to research the effect of crooking and stretching hip joint on the gravitative moment in the process of swing down, the two segments human model is used, which are up-body segment and legs segment (see Fig 1). The gravitative moment of the body at an arbitrary time of swing down is:

$$M = m_1 g r_1 \sin \alpha + m_2 g [L_1 \sin \alpha + r_2 \sin(\alpha + \theta)] \quad (1)$$

It's obvious that the gravitative moment M has relations not only

to the angle of upbody swing down

α , but also to the angle of crooked hip joint θ . With the equation (1), we have the derivation of gravitative moment M with respect to the angle of crooked hip joint :

$$\frac{\partial M}{\partial \theta} = m_2 g r_2 \cos(\alpha + \theta) \quad (2)$$

making $\frac{\partial M}{\partial \theta} = 0$, the extreme value of θ can be obtained

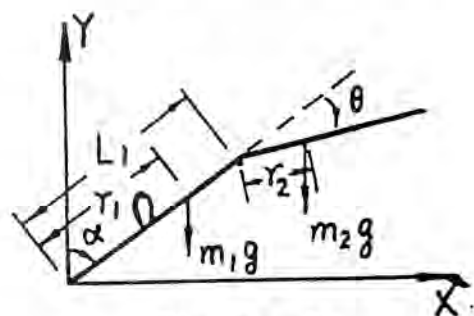


Figure 1

$$\alpha + \theta = 90^\circ$$

(3)

The second order derivation of gravitative moment M with respect to θ is negative, which showed that gravitative moment M has a maximum at the extreme point $\alpha + \theta = 90^\circ$.

The above analysis showed that:

1. the gravitative moment could be increased by crooking and stretching hip joint in the phase of swing down.
2. the range of hip's crooking is so limited that the sum of α and θ couldn't be over 90° .

The results of research showed that the technique of swing down in which the sum of α and θ is kept being 90° could be used for the movements of dismount and movements needing maximum initial velocity. For example, Tongfie's dismount of 720° tend which used the technique. But, in general "flight" movements, it is unnecessary to reach the maximum. For example, in Tkatchev vault, most gymnasts' maximum angle of crooked hip joint θ is only 37.9° , and the sum of α and θ is only 68.3° .

After swing down over the level of the bar, the body becomes position of arched body. This position combines the advantages of keeping the sum of α and θ being close to 90° which can delay the decrease of gravitative moment, and of stretching the muscle of crooking on regions of abdomen and legs which is useful for whipping the legs explosively. The data of practical measure showed that the movement of whipping the legs explosively could increase the sum of kinetic and potential energy of the body.

The data of elite gymnasts showed that in the movements of unvault "flight" and dismount whip time of legs is early. It begins at 50° under the bar generally. It is called early SWML. But in the movement of vault "flight" the whip time of legs is late. It begins at 60° under the bar generally. It's called late SWML.

GRIP STRENGTH IN MALE MANUAL WORKERS: INFLUENCE OF VIBRATION EXPOSURE AND SELF REPORTED SYMPTOMS.

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Introduction:

Complaints and disorders in the hand are common amongst manual workers (Hagberg in press; Silverstein et al, 1987) and grip strength has long been considered an objective measure of hand function following disease or injury (Lee et al, 1974). Little work, however, has considered grip strength as a response measure in relation to exposure amongst manual workers.

The purpose of this study is to determine whether there is a difference in grip strength in three occupational groups, one with low exposure to hand-arm vibration, one with high exposure and one with no exposure, and further, to consider the relationship between grip strength and age, grip strength and self reported symptoms in the different groups and whether grip strength could be used as a response measure in relation to exposure in manual workers.

Method:

The study base for this investigation consisted of sixty metal platers, sixty-nine car assemblers and forty-four office workers. All subjects were male.

Following informed consent, grip strength measurements were taken by a trained occupational health nurse in the medical centre adjacent to the plant. A questionnaire, administered by the same nurse, consisted of questions relating to age, exposure to hand held vibration tools, signs and symptoms and self assessment of the extent to which symptoms interfered with leisure activities.

The measuring instrument used was a strain gauge dynamometer with the adjustable hand grip span set at six centimetres and calibrated each day prior to the measurements. Each subject was tested individually following a warm up and they performed three maximal contractions for each hand alternately, beginning with the right hand. The highest of the three measurements was used but, if the highest two measurements differed by ten percent or more, a fourth measurement was considered.

Results.

Across the groups, there was a significant difference in grip strength, the assemblers being the strongest ($p < 0.01$) with a mean of 621 N for the right side and 591 N for the left. The platers lower grip strength measured (mean) 537 N on the right side and 513 N on the left; the office workers measured 529 N for the right side and 504 N for the left.

Surprisingly, the assemblers reporting wrist ache, daytime tingling and tremor were significantly stronger in left grip strength (mean 629N) than those who did not have the symptoms (mean 578N) ($p < 0.03$). The same tendency was noted for the right hand although it was not significant. Platers who reported nocturnal tingling demonstrated a significant weakness (513N right hand) compared to those without complaints (551N). There was a weak negative correlation between age and grip strength for the right hand among assemblers and

platers. There was no correlation between grip strength and exposure years to hand-arm vibration.

Discussion.

The stronger grip strength of assemblers may be a result of worker selection as truck assembling is a high status job in the automobile industry and is generally sought after by the stronger, young males. The work is heavy and requires a great deal of manual dexterity and as grip force is dependent on the use of small muscles of the hand (Kapandji, 1982), this training effect could account for the increased grip strength. The fact that the assemblers with symptoms were stronger than their counterparts without, may be because they have a greater grip strength initially, and therefore put excessive strain on ligaments and capsules.

The lower grip strength of the platers with nocturnal paraesthesia may be due to neuropathy. Nocturnal paraesthesia is an early sign of carpal tunnel syndrome which impairs motor function of the hand (Pfeffer and Gelberman, 1987).

The vibration exposure is low in assemblers compared to the platers (1.5ms²), however, the lack of relationship between exposure and grip strength may be due to the cross sectional design of the study for which the healthy worker effect was not controlled. The annual decrease of 2.5 - 5.5% in grip strength of lumberjacks is higher than the 1.5% normal decrease over the age of thirty-five (Parkila et al, 1986) and is interpreted as an effect of chronic vibration exposure.

There was little difference in the hand length measurement of the subjects and the constant six centimeter span used is optimum for these men (Petrovsky et al, 1980) and would not account for the differences in strength, however, different testers were at the two factories and there may have been a difference in encouragement given during testing.

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FORELIMB MOTION CYCLES OF THE EQUINE GALLOP STRIDE

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METHODS

Forelimb motion cycles were documented in 3 Quarter Horse fillies galloping at velocities between 13 and 14 m/s. High-speed cinematography was utilized at 243 fps. Body surface markers were applied on approximate centers of rotation of forelimb segments on the wing of the atlas, the center of the scapular spine, the shoulder joint, the elbow joint, the carpal joint, the metacarpo-phalangeal joint, and the proximal interphalangeal joint. These markers and the hoof toe were digitized on alternate film frames for a sampling rate of 122 Hz. Graphical representations were developed of limb motion cycles for right forelimbs on the right lead transverse gallop (leading limb) and the left lead transverse gallop (trailing limb).

RESULTS AND DISCUSSION

The cyclic motion patterns of the gallop stride of 3 horses are displayed in the following figures. Limb segments were plotted for stride cycles in two coordinate systems: an external coordinate system and an internal coordinate system fixed at the scapular marker. In the gallop, the scapular marker has similar vertical ranges of motion (10 ± 1 cm) on each side of the body,¹ and since the timings and positions of maximum absolute angles of the neck segment are similar for each side,¹ diagrams may be adjusted for equivalent neck segment maximal angles. These diagrams are useful in differentiating leading and trailing limb motions and also in identifying idiosyncratic limb kinematic patterns, particularly with regard to hoof motions. The techniques are useful in assessments of individualized equine gait patterns² and may have applicability in studies of the effects of shoeing, lameness and training on equine stride characteristics.

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1. Deuel, NR and Lawrence, LM. 1987. Kinematics of the equine transverse gallop. *J. Equine Vet. Sci.* 7(6):375-382.
2. Deuel, NR and Lawrence, LM. 1987. Individual variation in the Quarter Horse gallop. *Equine Exercise Physiology* 2, p. 567-574. ICEEP Publications, Edwards Brothers, Inc., Ann Arbor, MI.

FIGURE 1: HORSE A, RIGHT FORELIMB GALLOP MOTION CYCLE

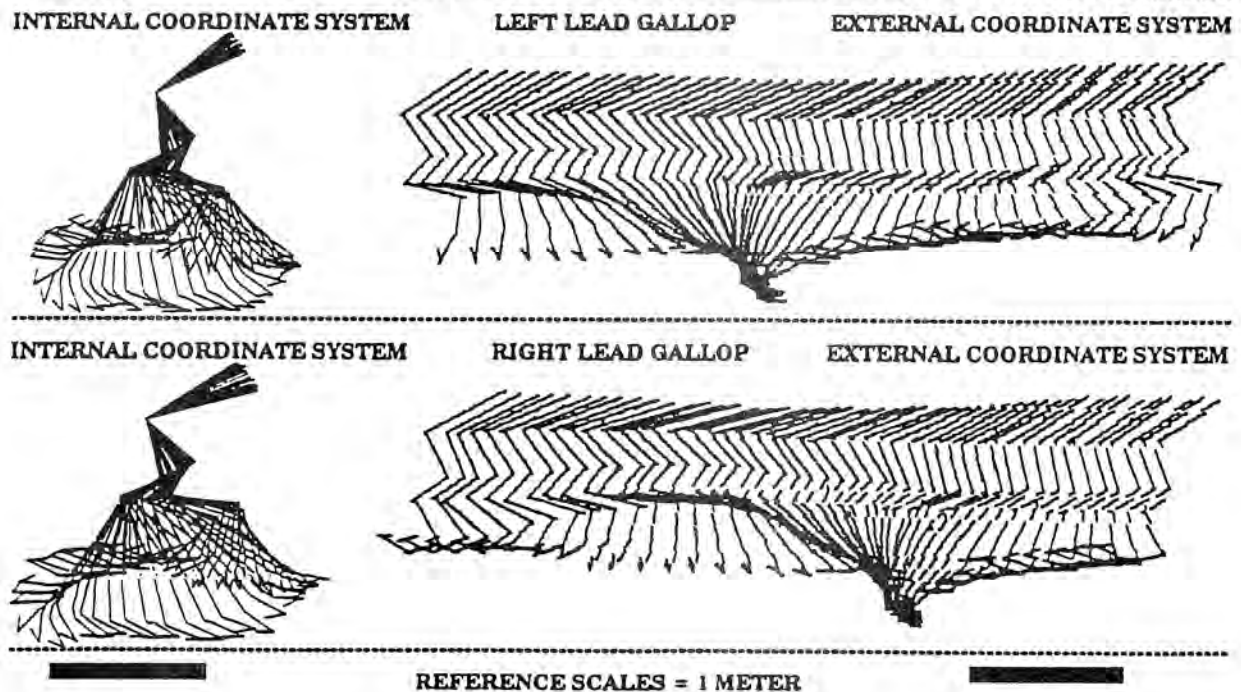
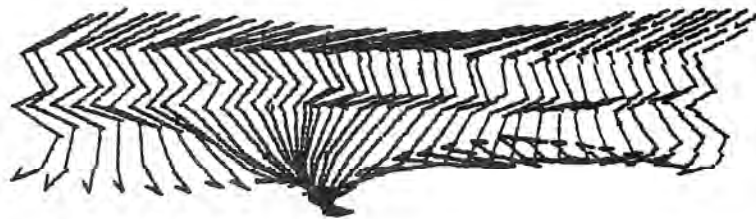


FIGURE 2: HORSE C, RIGHT FORELIMB GALLOP MOTION CYCLE

INTERNAL COORDINATE SYSTEM



LEFT LEAD GALLOP

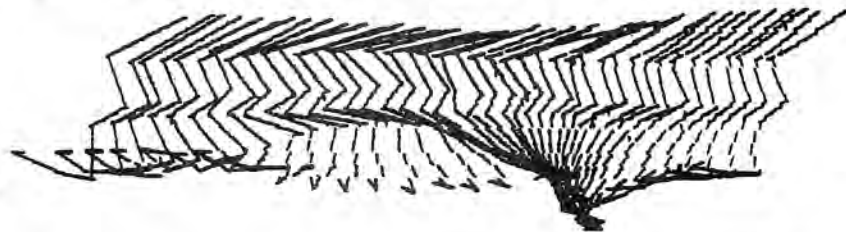


EXTERNAL COORDINATE SYSTEM

INTERNAL COORDINATE SYSTEM



RIGHT LEAD GALLOP



EXTERNAL COORDINATE SYSTEM

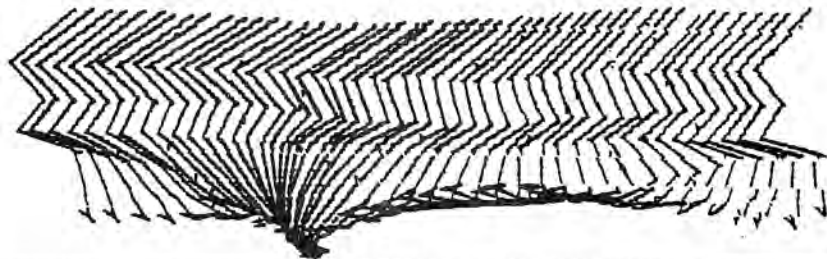
REFERENCE SCALES = 1 METER

FIGURE 3: HORSE D, RIGHT FORELIMB GALLOP MOTION CYCLE

INTERNAL COORDINATE SYSTEM



LEFT LEAD GALLOP

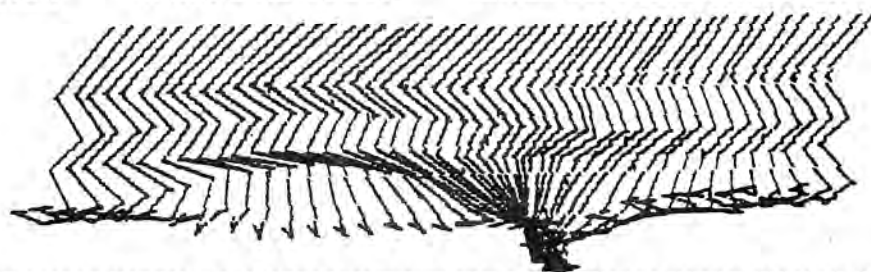


EXTERNAL COORDINATE SYSTEM

INTERNAL COORDINATE SYSTEM



RIGHT LEAD GALLOP



EXTERNAL COORDINATE SYSTEM

REFERENCE SCALES = 1 METER

The Biomechanics of Taijiquan Pushing Hands

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Huifeng Men Guixiang Gan

Matial arts department of Beijing Institute of P.E.

Shadow Taijiquan boxing is one of the famous branches of Chinese martial arts. Taijiquan pushing hands (TPH) is tending to become a new form of competition of pugilist art. The biomechanics of TPH is very complex, so up to now no report about it is documented. The purposes of this study is to find preliminary biomechanical principle of TPH.

(1) METHOD. Four highly skilled TPH male boxers were selected as subjects, young man as a attacker while old man as a defender, both of the old man are very famous Taijiquan masters in China. They can throw opponent 3-4 meters away while they look like nothing to do. Two high speed cameras were used to make 3-D film, set at a speed of 50 fps. Each subject were measured 3-D forces synchronisly with filming using two Kistler platforms, controlling by IBM-PC/XT microcomputer. Digitizing data were smoothed with a Butterworth digital filter set at a cutoff frequency of 7Hz.

(2) RESULT. The "peng" method of the master Wang were presented as a typical example, there were similar biomechanical characteristics in other's. Fig. 1 snows the stick fig. and Fig. 2 presents the CG path and force vectors for both of the attacker and defender. Fig. 3 shows curves of the foreces and CG.

1. In this experiment the defender distinguishes true attacker's force from false and yields to make attacker emptiness, using following methods: 1) to push attacker to full backward by relative larger horizontal force (28.8Kg, 2.7 times of the attacker's force). 2) When the attacker discharges the defender withdraws back immediately leading attacker emptiness (zero horizontal force). 3) The defender's increament of the forces is very small and smooth (attacker can not know him while he knows attacker). 4) The defender transfers the attacker's horizontal kinetic energy into defender's vertical kinetic energy and potential energy, the defender gets 2.5 times of the body's weight, the attacker must make 2.5 times of the horizontal force to push defender fall down. The defender sinks his CG, it makes himself more stable. This is a new finding about theory of stable. The defender's forward stable angle is 6 times of the attacker's angle (24 deg relative to 4 deg.). 5) the center of pressure (PC) of the defender backward moving slight, the range is much smaller than attacker's (0.66cm relative to 7.0cm). 6) Duration of the repaling, distinguishing and yielding is very short so the attacker is difficult to respond (0.14 sec).

2. The defender used elastic buffer, making the attacker to almost zero weight (from 58.7Kg to 12.94Kg). When this opportunity arises the defender hits suddenly with explosiveness of whole body's force. The attacker's increaments of ground reaction forces (134.4Kg) are much more than defender's (12 times of the defender's). The horizontal impulses of attacker's at take off mostly come from himself (60%). It snows the characteristics of the "borrow opponent's force to strike him" and "using four ounces to deflect a thousand pounds".

3. The defender's discharging order is important, that is knee extension (at 0.26"), hip flexion, trunk forward incline (at 0.32") elbow extension (at 0.44"), shoulder extension (at 0.46"). The most of contribution comes from leg (79%). It conforms with the famous Taijiquan scripture very well.

4. The biomechanical experiments shows that TPH has a significant difference characteristics of biomechanics among five phases.

5. The movement of the center of pressure and CG of the boxers, vertical and horizontal force, angular parameters of the joints and moments of the ground reaction forces can provide useful indicators for improving the techniques of the TPH.

6. The biomechanics can make contributions for getting insights into the traditional TPH.

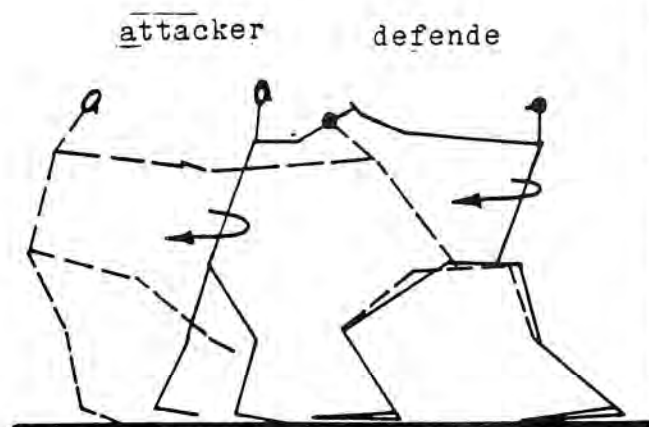


fig.1 attacker, defender stick figure

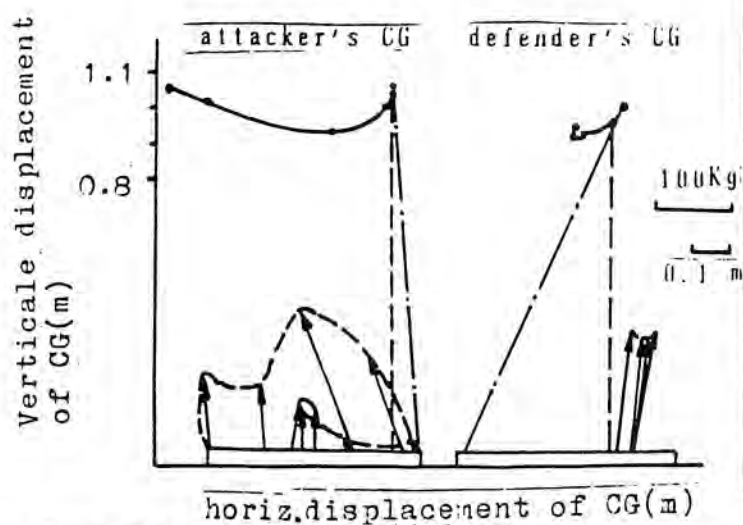


fig. 2 centers of gravity and forces

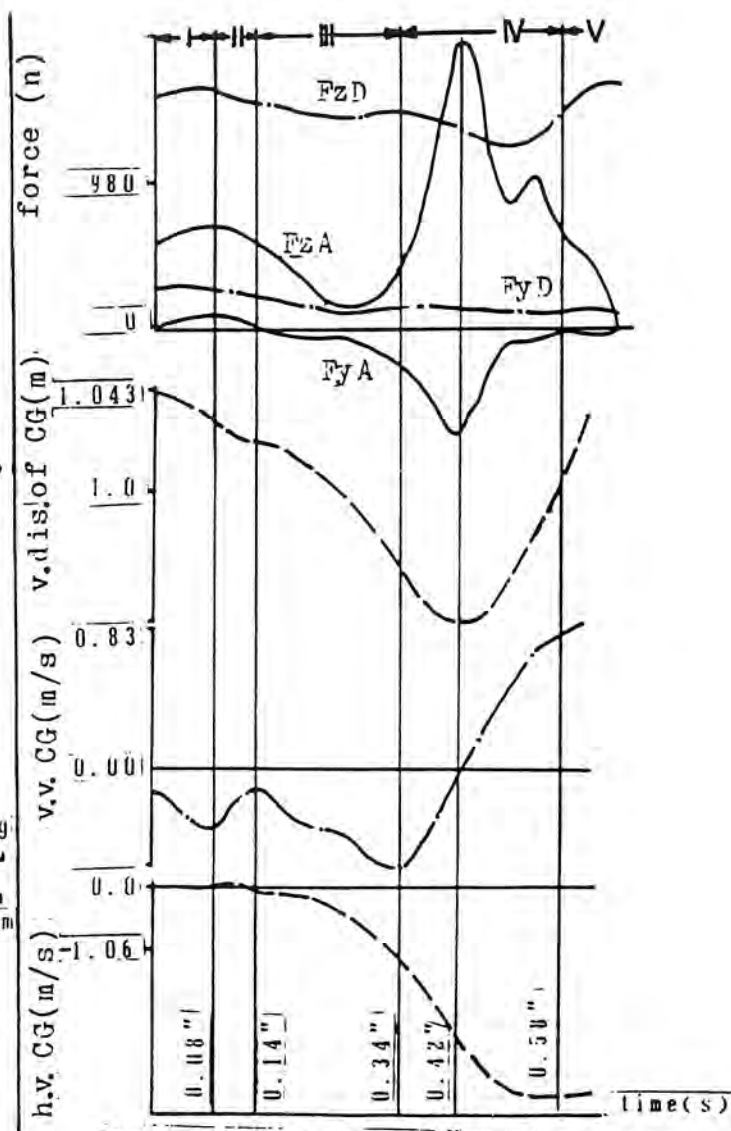


fig. 3 the relationship between CG and forces.

FzD, FyD, vertical, horizontal force of the defender.

FzA, FyA, vertical, horizontal force of the attacker.

FzD, FyD, vertical, horizontal force of the defender.

FzA, FyA, vertical, horizontal force of the attacker.

Biomechanical Analysis of Backward Giant Full Twisting to Handstand on Parallel Bars for Two World Champions

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Tailai Liu

Sports School of Beijing Institute of P.E.

Many gymnasts won PB world titles with their talented series of exercises giant. The coaches and gymnasts are very interesting in Backward Giant Full Twisting on PB(BGFT), but now no report about it is documented. The purpose of this study is to find good techniques to improve our student Li Jing and other gymnasts. The practical result is that Li Jing and Xu Zhiqiang won PB world champions at two international games. The kinematic data was calculated from high speed film(100 fps, digital filter with cutoff 7Hz).

RESULT

Li's technique was excellent and Xu's performance has significant potential required to improve in the future.

(1). Typical technique of "outstretched" style at downward swing.

Fig. 1 shows Li's performance diagram, Fig. 2 shows Li's and Xu's angles of the hip and shoulder. The Li's shoulder and hip was stretched fully(188.9 deg., 180.4 deg.) at position of the bars. Both of the angles of hip and shoulder is changing within 14 deg. The his largest radius of center of body mass in the downward swing can get a greater rotational work. It is important source of energy for following swing up. Path of the center of body mass is smooth and circular of this technique. So load bearing to hands increases smooth without shocking (fig.3). The Xu's CG path shows that his radius of CG were shortened seriously by flexing shoulders and hips, in bigger range (30.1 deg. and 22.8 deg.). Xu did not use downward swing of "outstretched" and had not big enough of kinetic energy, as result, the height of Xu's CG is 15 cm lower than Li's. Xu had not enough time to complete twisting in 360 deg. The path of the Xu's CG appears disturbance by the adding support of arm at third quadrant. It leads the twisting movement cannot complete before the vertical handstand position.

(2). The pattern of changing grasp is one of the key techniques for this exercise.

During swinging up only right hand was released while the pattern of holding bar for left hand is changes progressively. First from the hanging position to the supporting position by pronate joint of articulationes carpometacarpeae while left hand presses bar with caput of fifth ossa metacarpi is risen to top surface of the bar. Third, grasp movement is inversion of left hand in order to complete coming supporting grasp.

(3). Twisting techniques

1. Extra moment. When joints of the knees swings up to the level of the bars, body begins to twist with distal legs, using double grasps of the hands to act force at the bars. 2. Technique of swing arm. Path of swinging right arm is important fact for swinging up with twisting 360 deg. We can see clearly that path of right arm from over head to bottom, passing chest(Fig.4). According Coriolis theorem we know that Coriolis inertial force (F_c) can be created by combining angular velocity of body swinging up (ω) and linear velocity of swinging right arm (V). The formula is: $F_c = 2M * \omega * V$

M , is mass of swinging right arm.

The gymnast felt that the twisting trunk is "active" for this phase, and legs twisting is "pasive". It accords with prvious beamechanical analysis.

Calculations show that average twisting angular velocity in 0-90 deg. is 6.83 rad/sec, but it's average is 9.83 rad/sec for twisting angle between 180 to 270 deg. The reason of increasing twist is influence of the Coriolis inertial force of right arm and decreasing moment of inertial of total body around longitudinal axis of the body. Because the body stretches fully during twisting angle 180-270 deg.

The center of body mass must move to appropriate distant to left bar, because twisting body have to get certain support with left hand during later phase. The horizontal force creating by left hand push CG to right bar and the gymnast will be dropped to the ground if without moving CG to the left bar. It took a long time to solve this problem for Li Jing and other gymnasts, the CG have to move to left side about 6cm from middle plane between two bars.

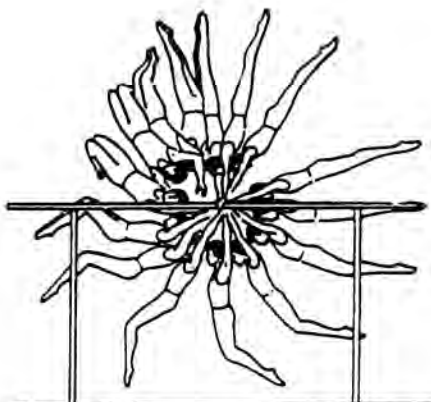


fig.1 Li jing performance

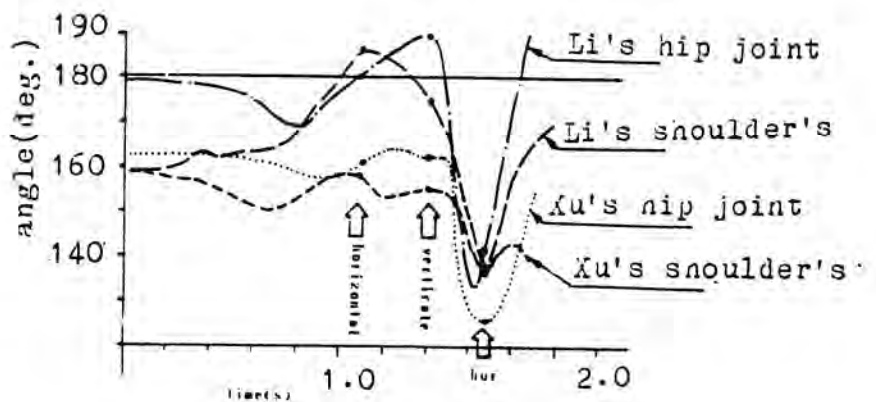


fig.2 the angle of the joints of hip and shoulder

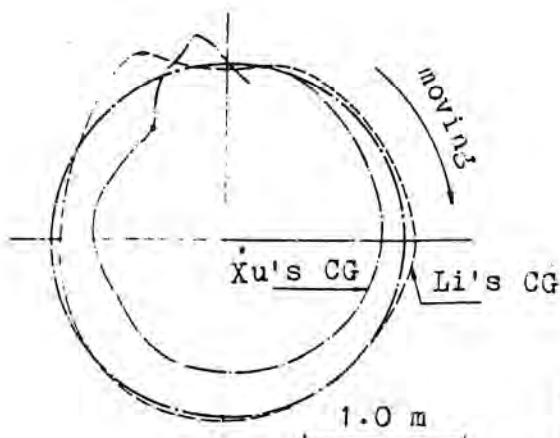


fig.3 the path of the C.G.

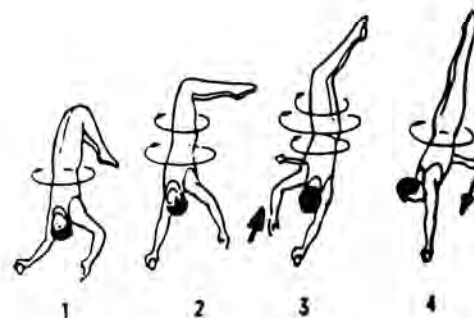


fig.4 the arm swing during twisting

ACCOUSTIC EMISSION DURING COMPRESSION, FROM NORMAL AND DISEASED BONE

Arye Bivas, Isaac Leichter, Isaac Roman, Joseph Y Margulies and Ariel Simkin

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INTRODUCTION

The study of acoustic emission utilizes sonic signals emitted by a material when excited by physical stimuli. During mechanical loading energy is released as a result of dynamic processes such as microcracks and dislocations. The released elastic energy generates acoustic waves within the material. Recording of the acoustic signals emitted by materials under loading has been used for the examination of structural disorders in fiber reinforced composites. In the present study this technique was used for examining trabecular bone tissue from normal subjects and from patients with metabolic bone disorders. It was anticipated that this examination would provide an insight into the mechanisms responsible for the mechanical behaviour of trabecular bone under load.

MATERIAL AND METHODS

Trabecular bone specimens were taken from 31 femoral heads which were excised from osteoporotic patients during replacement of the femoral head following subcapital fractures and from 6 femoral heads of osteoarthritic patients during total hip replacement. These were compared to specimens taken from 13 femoral heads which were obtained at autopsy from subjects under the age of 60, free of known bone disease. The cylindrical specimens were taken from the trabecular region under the cortical layer of the femoral head so that their axis of symmetry coincided with that of the femoral neck. Each cylinder was of 10 mm height and 10 mm diameter. The specimens were kept at a temperature of -18° C during the period of measurements and thawed before loading at room temperature.

Each specimen was compressed along its axis at a constant speed of 1 mm/min, using a J.J. Lloyd M30K Material Testing Machine, until a maximum was obtained on the load-deflection curve. The acoustic signals emitted from the specimen during compression were recorded by a transducer attached to an aluminium clipper embracing the specimen. The signals were analyzed by an Acoustic Emission Testing System (AET 5000).

On each load-deflection curve the yield point and the maximum load point were identified. The ultimate compressive stress (UCS) and the elastic modulus in the pre-yield zone (EM) were calculated. The energy absorbed by the specimen up to the point of maximum load (AE) was calculated as the area under the load-deflection curve. The average rates of acoustic events up to the yield point (RY) and beyond it (RB) were calculated as the ratio between the number of events and the respective time interval. The average peak amplitude (AP) of the acoustic signals was also recorded. The values of these parameters in the

normal, osteoporotic and osteoarthritic subjects were compared and the significance of the differences was assessed by Mann-Whitney U Test.

RESULTS

Table 1 presents the average value and standard deviation of each parameter in the three groups of subjects.

	TABLE 1		
	Normals n=13	Osteoporotics n=31	Osteoarthritics n=6
UCS (MN/m ²)	11.5 ± 3.9	7.6 ± 4.5	9.7 ± 1.6
EM (MN/m ²)	157.1 ± 38.4	116.5 ± 50.0	116.1 ± 17.2
AE (Jx10 ⁻³)	0.867 ± 0.576	0.363 ± 0.299	0.677 ± 0.170
RY (sec ⁻¹)	1.6 ± 1.2	2.9 ± 2.1	3.9 ± 1.0
RB (sec ⁻¹)	4.2 ± 2.1	7.5 ± 4.8	8.1 ± 3.0
PA (dB)	44.4 ± 1.2	47.0 ± 3.4	46.3 ± 3.2

Table 2 presents, for each parameter, the ratios between the average value in the diseased and the normal bones.

	Osteoporotics/Normals	Osteoarthritics/Normals
UCS	0.66	0.84
EM	0.74	0.74
AE	0.42	0.78
RY	1.81	2.44
RB	1.79	1.93
AP	1.06	1.04

It was found that UCS and EM were significantly lower in both diseased groups while RY, RB and AP were significantly higher than in the controls. The AE was lower in both diseased groups, however between OA patients and normals the difference was not statistically significant.

DISCUSSION

The results of the mechanical behaviour under compression showed that trabecular bone tissue of both OP and OA patients was weaker, less stiff and less ductile than of normal subjects. The higher rate of acoustic events both before and beyond yield in both OP and OA patients may reflect the greater extent of trabecular failure during the loading process. The higher peak amplitude of signals in the diseased bones may be attributed to either the higher energy of the acoustic events or to a lower attenuation of the signals in the diseased bones.

SHORT TERM EFFECTS OF VOLUMINOUS HEAVY RESISTANCE TRAINING
AND RECOVERY ON THE SNATCH TECHNIQUE IN WEIGHTLIFTING

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INTRODUCTION. Changes in olympic weightlifting technique among elite weightlifters are usually rather limited, although prolonged training may lead to some further development in their neuromuscular performance (Kauhanen et al. 1988). Due to their extensive training background acute and/or short term changes in selected characteristics of the neuromuscular performance may take place only after exceptionally intensive and/or voluminous training sessions (Häkkinen and Kauhanen 1988). The present study was undertaken whether these types of voluminous heavy resistance training sessions may also lead to short term changes in the biomechanical structure of the lifting and in the maximal weightlifting performance itself during a training period of one week.

METHODS. 7 male elite weightlifters from the Finnish national weightlifting team (from several weight categories from 52 to 90 kg) volunteered as subjects for the present study. No training or other physical strain was allowed during two days before the experiment. The first three days of the experiment were characterized by heavy resistance and high voluminous training with two training sessions per day including the olympic lifts and several assistant exercises. The fourth day was moderate in its intensity and lower in its volume including one session. The fifth day included again two sessions and the sixth day was a resting day. The measurements (every second day; 4 times altogether) were carried out in the mornings of the first (M1; after two resting days), third (M2; after two voluminous training days), fifth (M3; after a moderate training day) and seventh day (M4; after a resting day) always at the same time of the day (9.00-11.00 a.m.). The measurements during the sessions including the snatch as the first exercise took place so that the lifting conditions differed as little as possible from the "normal". The best (in kg's) accepted snatch lift of each session was taken for the final analysis. The following biomechanical parameters were registered during the different lifting phases (first pull, second pull, third pull, and drop under barbell): maximal vertical ground reaction forces (GRF, expressed as percentages of the load of the barbell plus the weight of the lifter) with a force platform, knee angular velocities (KAV, $^{\circ} \times s^{-1}$) with an electrical goniometer, lifting height of the barbell (LHB, cm) with a special measurement device and electromyographic activities (IEMG, mV) of vastus lateralis (VL) and vastus

medialis (VM) muscles with a telemetric recording (Kauhanen et al. 1984).

RESULTS. No statistically significant changes occurred in the maximal snatch performance (in kg's) between the four measurements (M1-M4). During the first pull of the snatch GRF increased ($p < 0.05$) from M1 to M2 and to M3 and decreased ($p < 0.05$) from M2 to M4. During the third pull GRF increased ($p < 0.05$) from M1 to M3 and decreased slightly from M3 to M4. During the drop under the barbell KAV decreased ($p < 0.05$) from M1 to M2 and to M3 and increased slightly from M3 to M4. LHB decreased ($p < 0.05$) from M1 to M4. The averaged IEMG's of the VL and VM muscles remained rather unchanged except for a slight increase during the second pull of the snatch from M1 to M4.

DISCUSSION. The present results showed that after two intensive voluminous training days the knee angular velocity during the drop under the barbell decreased indicating a decreased capacity to perform the final rapid phase of the snatch. At the same time the ground reaction forces increased especially during the first pull of the snatch. The decreased lifting height of the barbell and a slightly increased EMG-activity during the second pull further reflect the changed biomechanical characteristics of the lifting during the present experiment. Despite these changes in the lifting technique the lifters were able to maintain their maximal weightlifting performance the same by some kind of compensatory mechanisms such as a more active role of the upper body while catching the barbell overhead. After one day of rest the knee angular velocity increased slightly and the ground reaction forces reattained their starting levels indicating possible short term effects of the recovery. In conclusion, the present results indicate that selected short term changes may take place in the biomechanical structure of the snatch even in elite athletes during a short but very intensive and voluminous training period, although this may not necessarily lead to changes in their maximal weightlifting performance.

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DYNAMICS AND BIOMECHANICS ON START AND ACCELERATED PHASE OF A SPRINT

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The start can be divided into two stages, of which the first is from the athlete hearing the gun to his back-leg driving off the starting block, and the second is to the fore-leg driving off from the starting block. All the strides in the sprint fall into the stages of flight, buffering, and tension. For the two stages of the start, and each of the stages of buffering, and tension in the strides, three kinetic equations in polar coordinate form (theorem of the moment of momentum in relation to mass center, theorem of the moment of momentum of landing point, and the energy integral form of the differential equation of the motion of the mass center) are given respectively; and also two kinematic equations, the relation of Δr , the changed of the polar coordinate of the mass center to foreblock or landing point, with the radial velocity V_r (mean value) of the mass center and the time interval Δt , and the relation of $\Delta \theta$, the change of another polar coordinate, with the circumferential velocity V_θ (mean value) of the mass center, polar coordinate r (mean value), and time interval Δt , are given too. The flight stage is the equations of the free falling of the mass center. In combination with the constitutive equations for each stage of the athlete, the first stage of the start is the constitutive relation (block reaction force — velocity curve) of the extensor muscles of the ankle of the back leg; the second stage of the start, and the extension stages in the sprint are the constitutive relation of the extensor muscle group (reaction force-velocity curve). In each buffer stage in the sprint, there exists a limit value of reaction-force in supporting the ground due to the athlete's ability (like admissible stress). From these general equations of dynamics and biomechanical constitutive equations, the technical parameters that an athlete can select are: the various geometric parameters for the start; the radial component of the reaction force of the fore-leg in driving the ground in the first stage of the start; the end inclinations in the first and second stages of the start; the each stride length in the sprint; the radial distance of the mass center from the landing point at the initial instant of various strides landing the ground; the end inclinations of each extension; and the moment of momentum in relation to the mass center at the ends of the first and second stages, and the buffer end and extension end of each stride. In the sprint, with the increase in running velocity, the moment of momentum $\sum m_C(mv)$ decreases, and naturally tends approximately to zero in the intermediate running. From this, the method for determining an optimum technical parameter group is obtained, and it is also possible to discuss the effect of a change of a certain parameter on the record of an athlete.

This paper puts forward that for the first stride in the sprint, the buffer reaction is not to decrease the running velocity. The several joint strides between the accelerative running and intermediate running should be decelerative running, instead of accelerative running. (Fig.)

For these few strides, the athlete consumes less energy, and is relatively relaxed. Owing to the fact that the moment of momentum of the landing leg in relation to the mass center changes with the velocity of running, the reaction-force of ground will pass through the rear of the mass center in accelerative running. Both in accelerative and intermediate running buffering reaction-force of the ground is active force, and not resistance. This should not be reduced. At the beginning of landing of various strides, the distance of the landing point in front of the mass center should be determined by the optimum parameter group, although the horizontal component of the buffer reaction force of ground is backwards, it will not do to think that reduce this distance is taken into consideration.

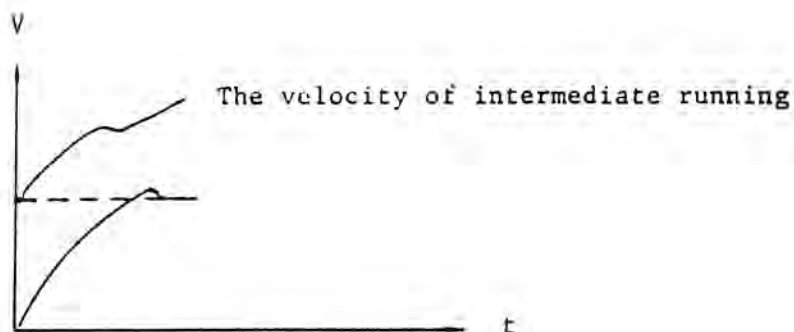


Fig.

Reference

Hu Lin-Yi "GENERAL MECHANICS AND BIOMECHANICS ON INTERMEDIATE PHASE OF A SPRINT" — XI International Congress of Biomechanics 1987

EFFECTS OF SYNERGISTIC WRIST MOTION ON FLEXOR TENDON EXCURSION IN THE HAND

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INTRODUCTION: Flexor tendon repair is frequently compromised by dense adhesions which develop between the synovial sheath and repair site and lead to a limitation of tendon excursion and poor digital function. The concept of immediate protected mobilization after tendon repair to reduce the incidence of tendon adherence has been advocated. Experimentally and clinically, early motion has been shown to improve digital function following tendon repair. Most commonly, clinical mobilization is carried out with a dorsal splint which holds the wrist joint in moderate flexion and utilizes a rubber band to passively flex the finger. It is believed that active extension mobilizes the tendon and inhibits antagonistic flexor activity. A palmar bar modification can be added across the palm to improve flexion of interphalangeal joints and thus tendon excursion. However, realistic measures of tendon excursion and relative tendon glide achieved using these conventional splints and their optimum ranges have not been carefully studied, and the efficacy of these devices is questionable. The purpose of this study is to examine the absolute and relative flexor tendon excursions in a conventional dorsal splint and dorsal splint with palmar bar. In addition, the hypothesis that the incorporation of synergistic wrist motion will increase flexor tendon excursion in the finger but not tendon tension was also tested.

MATERIALS AND METHODS: Four fresh cadaveric forearms with intact hands were used for this experiment. Skin incisions were made on the palmar side of the proximal phalanx, palm, and distal forearm. The tendon sheath was opened between the A3 and C1 pulley to avoid damage to the pulley system. Both the flexor superficialis (FS) and flexor profundus (FP) tendons were tagged with a buried 4-0 wire suture in zones II, III, and V, and the skin was then closed with 4-0 nylon suture. The excursion experiments were performed. Radiographs were taken in the true lateral plane. Excursion was measured from the film based on the displacement of the metal markers with the reference of bony markers. Two groups of experiments were performed to compare tendon excursions in three splint designs (Figure 1). Experiment 1 was designed with the muscles in a simulated, relaxed state with passive tension. Both the FS and FP were loaded by rubber bands with enough tension to maintain the finger and wrist joints in the natural resting position. The excursions of both flexors in zone II, III, and V were then measured in the three splinting techniques studied.

Experiment 2 was designed to compare the tendon excursion by using three different splinting techniques under varying simulated isotonic muscle tension. Instead of using rubber band traction, different weights were hung on these two flexor tendons. The tendon excursions were then measured in the same fashion as that in the first experiment.

RESULTS: The results of experiment 1 are summarized in Table 1. Synergistic wrist motion provided the largest excursions of the FP and FS in zones II and III as well as the largest relative displacement between these two tendons, which are significantly greater than those using the splint with a rubber band or one with an additional palmar bar.

The results of experiment 2 with the muscles under simulated isotonic tension are comparable to those of experiment 1. Tendon excursion increased with the addition of weight on the flexor tendons. The differences in the measured excursion were not as obvious using synergistic wrist motion, despite the addition of weights, and again, synergistic motion produced the largest amount of relative glide between the two tendons.

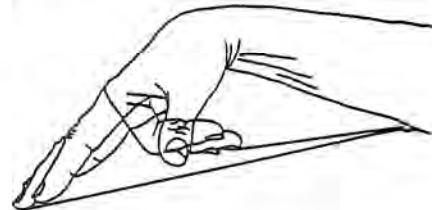
DISCUSSION: As reported in the literature, frictional force within the carpal tunnel rises precipitously as wrist flexion approaches 60 degrees. This friction may lead to tendon buckling rather than gliding in the palm with passive flexion of finger such as would be provided by the rubber band. More proximal muscle force is thus necessary to produce excursion. Synergistic wrist motion increased the absolute tendon excursion and relative glide of the flexor tendons in critical zones II and III without buckling or slackness of tendons in the palm. The data suggest that if a fixed wrist position is desired for a dorsal splint, a position closer to neutral might be preferable to one more severely flexed.

Table 1. Comparison of tendon excursion (mm) using different splints with muscles in simulated passive tension zone.

Zone	Tendon	Splint with with Rubber Band	Synergistic Wrist Motion
		Mean (S.D.)	Mean (S.D.)
II	FS	7.8 (1.9)	15.2*(3.2)
	FP	10.1 (1.7)	19.8*(3.9)
	FP-FS	2.3 (0.4)	4.6*(1.8)
III	FS	12.1 (1.6)	20.7*(2.0)
	FP	13.9 (0.1)	25.8*(3.3)
	FP-FS	1.2 (0.9)	5.2*(1.3)
V	FS	10.5 (1.9)	1.6*(0.7)
	FP	10.5 (0.6)	5.7 (5.9)
	FP-FS	1.2 (0.7)	5.6*(4.0)

*Duncan's multiple range test at $P < .05$ significantly different in the same group of zone and tendon.

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PHYSIOLOGICAL CONSIDERATION OF MUSCLE FORCE DISTRIBUTION

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INTRODUCTION An important task in the analytical determination of muscle force distribution is that of describing the muscles with appropriate models. When skeletal muscle is stimulated, it is rapidly activated and changed from a passive tissue into a dynamic tissue capable of developing force. During activation, the length of the muscle may decrease, increase, or remain the same, depending on the motion of the limb and the external forces opposing this motion. The potential force which the muscle can generate at activation is a function of the muscle length and velocity at that instance. The purpose of this study was twofold. The first was to develop a muscle model that is phenomenologically correct without being overwhelmingly complex for practical applications. The second was to incorporate this relationship into an analytical model to determine the muscle distribution.

METHODS AND MATERIALS Muscle length-tension relationship: Recently, Woittiez (1984) developed a three-dimensional muscle model describing the length-tension relationship of whole muscle while accounting for muscle architecture. The active length-tension relationship was defined as a parabolic function of the architecture index (i_a); the mean fiber length divided by the muscle belly length at optimum muscle length. Further, Otten (1985) addressed the asymmetry of the length-tension relationship. Based on these observations, a model was developed for the normalized muscle active tension (f^A) which accounted for the muscle length as a function of the muscle strain (ϵ) and i_a as follows:

$$f^A = \exp \left[- \left[\frac{0.96 * (1 - \frac{1}{i_a})}{(\epsilon + 1) - 1.0} - 0.35 * (1 - i_a) \right]^2 \right] \quad \text{for } i_a < 1 \quad (1)$$

$$f^A = \exp \left[- \left[2.72 * \ln(\epsilon + 1) \right]^2 \right] \quad \text{for } i_a = 1$$

Joint model: Based on the requirement that force (F) and moment (M) equilibrium must be maintained while the muscle is resisting an externally applied load, a set of constraint equations can be obtained:

$$\sum F = 0; \quad \sum M = 0. \quad (2)$$

The three-dimensional coordinates of the points representing the origin and insertion of each muscle crossing the joint were obtained from anatomic study. Using vector analysis, the components of the muscle force unit vector and moment arms about the three orthogonal axes could be calculated at any joint configuration. Further, using the muscle coordinates, the muscle excursion could also be calculated as a function of the joint angles.

Potential muscle force: The amount of force, $F_{\max}$, a muscle is capable of generating under isometric conditions is:

$$F_{\max} = F^A + F^P = (f^A + f^P) * PCSA * \sigma. \quad (3)$$

in which, F^A is the maximum active force, F^P is the passive force, f^A is described in equation (1), f^P is the normalized passive muscle force (Woittiez, 1983), PCSA is the physiological cross-sectional area, and σ is the maximum stress or force per unit area which can be generated by the muscle.

Muscle force distribution: Due to the indeterminate nature, the muscle force distribution is achieved by satisfying not only the force and moment equilibrium equations (2) but also the physiological constraint:

$$0 < F < \alpha F^A + F^P \quad (4)$$

In which, α is the activation or excitation level of all muscles. The unique muscle force is then obtained by minimizing α .

RESULTS To illustrate this method, the muscle force distribution about the elbow joint was studied. A unit load was applied to the distal radius. The comparison of the results of muscle force with and without consideration of muscle length-tension relationship and the muscle architecture are summarized in Figure 1.

DISCUSSION This study demonstrated how muscle physiology can be implemented in the optimization procedures for muscle force determination. The model can easily be expanded to include the muscle force-velocity relationship by modifying the normalized muscle force calculation. In addition, the model could also be modified for muscles crossing multiple joints or for a single joint muscle with multiple degrees of freedom by considering multiple muscle stimuli, i.e., one α for each degree of freedom.

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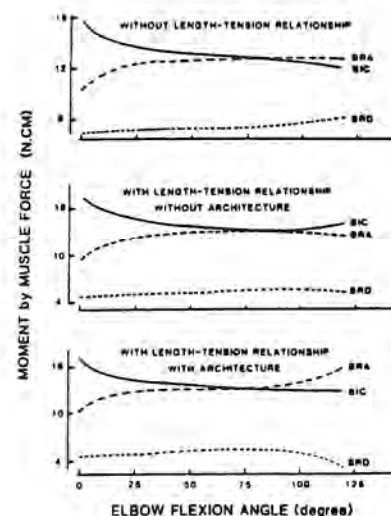


Fig. 1

THREE-DIMENSIONAL KINEMATICS OF GLENOHUMERAL ELEVATION

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INTRODUCTION The shoulder permits more motion than any joint in the human body. Confusing terminology of reference plane for elevation without universal measurement for description of shoulder motion exists in clinical evaluation. The purpose of this study was to define the plane of humeral elevation with respect to the plane of the scapula and to ascertain the effect of humeral axial rotation on elevation. This has potential significance for examination, rehabilitation, and selection of operative procedure performed on the shoulder.

MATERIAL AND METHODS Three-dimensional kinematic analysis using a 3Space™ Isotrak™ system housing three orthogonal coils in a source and sensor was performed on four fresh cadaveric shoulders. Each scapula was mounted on a Plexiglas arm in the upright anatomic position with the source positioned next to the scapula and the sensor bolted to the proximal humeral shaft. The scapula coordinate system, centered at the center of rotation of the humeral head, was derived by transformation of the data from the source coordinate system following digitization of the scapula. In this reference coordinate system, the z axis was perpendicular to the infraspinatus plane. The y axis was obtained by the cross product of z and the unit vector of two digitized points (P,Q) on the medial scapular border. X axis was the cross product of y and z and directed laterally (Fig. 1). Humeral motion with respect to the scapula utilized a Eulerian angle system based on a 1 - 3' - 1" rotation sequence (Fig. 1). The first rotation (ϕ) about x axis rotated the humerus into an appropriate plane of elevation. The second rotation (θ) about z' axis corresponded to arm elevation, while the third rotation (ψ) about x" axis represented humeral axial rotation. The neutral reference position was set with the humeral shaft parallel to the medial scapular border and transhumeral pin (inserted perpendicular to a plane formed by the humerus and supinated forearm at 90° elbow flexion) parallel to the plane of the infraspinatus fossa. In this position, the humeral axes system and the reference coordinate system corresponded. Elevation was affected with a constant load applied perpendicular to the humeral shaft using a pulley system. A universal joint was used to affect a full range of humeral rotation in each plane of elevation at five-degree increments anterior and posterior to the plane of the scapula. The experiment was performed with an intact rotator cuff, following which the cuff was excised, permitting scapular digitization with subsequent transformation of data from the source to the scapular anatomic reference coordinate system.

RESULTS Maximal elevation was found to occur at about 22° anterior to the plane of the scapula and was coupled with external rotation of the humerus. A typical pattern of elevation with respect to the scapula with corresponding humeral axial rotation is shown in Fig. 2. Anteriorly and in the scapular region, external rotation is required for maximal elevation. Conversely, internal rotation is necessary for optimal elevation posterior to the scapula.

DISCUSSION This is the first study to: 1) define the plane of maximal elevation and 2) quantitate the interrelationship of elevation with humeral axial rotation. Maximal elevation is anterior to the plane of the scapula in the region of the coraco-acromial ligament and coupled with external rotation. Anterior to the scapula, any derotation or rotation away from the externally rotated position lowers elevation. Conversely, for optimal elevation posterior to the scapular plane, internal rotation of the humerus was required. The observed mechanism was clearing of tuberosity impingement from beneath the acromion and relaxation of the inferior capsular ligamentous constraints.

CONCLUSION This research has significance for shoulder examination, rehabilitation, and selection of surgical procedure, as any operation that limits humeral rotation will, of necessity, reduce maximal permissible elevation.

DEFINITION OF THE REFERENCE COORDINATE SYSTEM [LEFT SHOULDER (PA VIEW)]

Z: $\perp$ plane of scapula

Y: Z $\times$ PQ

X: Y $\times$ Z

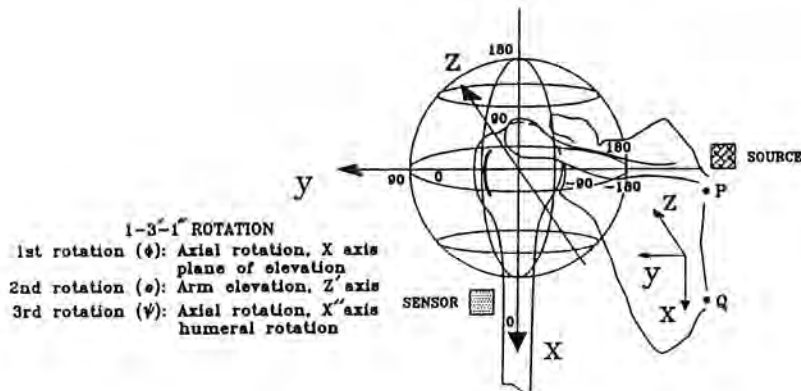


Fig. 1. Anatomic coordinate system and rotation sequence.

Maximum Elevation/Humeral Rotation/Plane

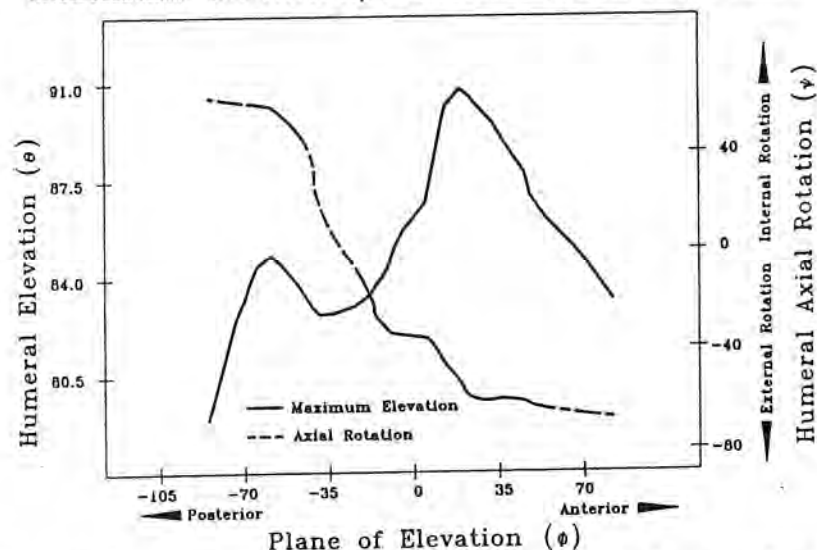


Fig. 2. Humeral elevation and rotation in each plane referable to the scapula.

HARMONIC WAVES IN A PRESTRESSED ARTERY

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In the present work, utilizing the theory of small deformations superimposed on a large initial deformation, the propagation of a harmonic wave in a thick walled cylindrical artery filled with an incompressible inviscid fluid is studied. Although the blood is known to be a Non-Newtonian viscous fluid, for simplicity of the problem, it will be assumed to be an incompressible inviscid fluid, while the arterial wall is taken to be a thick walled, incompressible and isotropic elastic material. Due to variability of the coefficients of governing differential equations a closed form of solution cannot be given. Therefore, a truncated power series method has been utilized and the approximate dispersion relation with several special cases are obtained.

GOVERNING EQUATIONS AND SOLUTIONS

Fluid Equations: Assuming the blood is an incompressible inviscid fluid and the motion is axially symmetric, the governing equations in cylindrical polar coordinates are given by

$$\frac{\partial \bar{p}}{\partial r} + \bar{\rho} \frac{\partial \bar{u}}{\partial t} = 0 \quad \frac{\partial \bar{p}}{\partial z} + \bar{\rho} \frac{\partial \bar{w}}{\partial t} = 0, \quad \frac{\partial \bar{u}}{\partial r} + \frac{\bar{u}}{r} + \frac{\partial \bar{w}}{\partial z} = 0 \quad (1)$$

where $\bar{\rho}$ is the fluid density, $\bar{p}$ is the fluid pressure, $\bar{u}$ and $\bar{w}$ are the velocity components in the radial and axial directions, respectively.

Equations of Arterial Wall: Considering that the blood vessels are subjected to a large static initial deformation and the superposed dynamical deformation are small, in cylindrical polar coordinates the governing field equations for axially symmetric motions are given by [1].

$$\frac{\partial p'}{\partial r} + \beta_1(r) \frac{\partial^2 u}{\partial r^2} + \frac{\beta_2(r)}{r} \frac{\partial u}{\partial r} - \frac{\beta_3(r)}{r^2} u + \beta_4(r) \frac{\partial^2 u}{\partial z^2} - \rho \frac{\partial^2 u}{\partial t^2} = 0 \quad (2)$$

$$\frac{\partial p'}{\partial z} + \beta_1(r) \frac{\partial^2 w}{\partial r^2} + \frac{\beta_2(r)}{r} \frac{\partial w}{\partial r} + \beta_4 \frac{\partial^2 w}{\partial z^2} + [\beta_2(r) - \beta_3(r)] \frac{1}{r} \frac{\partial u}{\partial z} - \rho \frac{\partial^2 w}{\partial t^2} = 0 \quad (3)$$

$$\frac{\partial u}{\partial r} + \frac{u}{r} + \frac{\partial w}{\partial z} = 0 \quad (\text{incompressibility condition}) \quad (4)$$

where p' is the increment in hydrostatic pressure of the solid body, ρ is the mass density of arterial wall, u and w are the incremental displacement components in the radial and axial directions and $\beta_i(r)$, ($i=1,2,\dots,5$) coefficients which depend on initial field. For other details, the reader is referred to [1].

As might be seen from (2)-(4), the coefficients of these set of differen-

tial equations are variable, it is thus impossible to give u , w and p' a closed form of solution. Nevertheless, since the coefficients $\beta_i(r)$ are analytic in r a truncated power series solution for u , w and p' is always possible. For this purpose we seek a solution to (1)-(4) of the form

$$(\bar{u}, \bar{w}, \bar{p}; u, w, p')(r, z, t) = (\bar{U}, \bar{W}, \bar{P}, U, W, P)(r) \exp [i(kz - \omega t)] \quad (5)$$

where k is the wave number, ω is the angular frequency and $\bar{U}, \dots, P$ are complex amplitudes of the wave. When (5) is introduced into (1)-(4) a set of simultaneous ordinary differential equations are obtained. The solution of (1) is expressible in terms of modified Bessel functions. To obtain a solution to (2)-(4), we first introduce a new dependent variable (y) as $r = \bar{r}(1+y)$, where $\bar{r}$ is the radius of midsurface, and expand the field quantities and coefficients $\beta_i(y)$ into a power series of y . In the expansion we kept the terms up to (including) y^2 and neglected the higher order terms. Upon using well defined boundary conditions we obtained the dispersion relation, which gives the phase velocity as a function of wave number and initial deformation. As a special case the phase velocities are obtained for thin membranes and large wavelengths and the result is compared with existing theories.

The more general results are obtained by numerical means. The results indicate that the wave speed increases with pressure while it decreases with axial stretch.

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Propagation of Impulsive Shock Waves in the Human Musculoskeletal System

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One of the major consequences stemming from man's development was the changing of his environment from resilient grassy fields, to the concrete and asphalt reality of the modern industrial city. This alteration of his surroundings has resulted in much higher dynamic loads on the human musculoskeletal system during normal gait. The impulsive loads to which the foot is subjected during gait may be considerably higher than the body weight, depending on gait velocity and foot viscoelastic properties. Because of these higher dynamic forces, the natural shock absorbers of the human musculoskeletal system become insufficient. The impulsive energy which invades the body during each heel strike can no longer be sufficiently attenuated, modified and dissipated by the foot only. This leads to an overloading of the rest of the natural shock absorbers of the human musculoskeletal system. The mechanically induced shock waves resulting from gait generated intermittent waves that propagate through the entire human body from the heel up to the head.

Several studies have associated impulsive loading resulting from gait with excessive fatigue and progressive damage to some elements of the human locomotor system. The elements usually damaged by degenerative changes (the foot, meniscus, intervertebral disc, etc.) are generally referred to as the natural shock absorbers with their main function being to attenuate and modify incoming shock waves traveling toward the skull. This attenuation is a rational form of absorption and dissipation of energy invading the body during heel strike.

Obviously, the significant reduction or even elimination of the injurious effects of gait induced shock waves are contingent upon the development of accurate methods to quantify these waves and the effect of various shock absorbing devices (footwear, insoles, etc.) on the human body.

The main purpose of this study was the refinement of the non-invasive, *in-vivo* methodology for the acquisition of real-time data of the impulsive shock waves at particular locations of the human musculoskeletal system and the investigation of the shock waves transformation by the body's internal shock absorbing elements.

A simple, non-invasive *in-vivo* technique has been used to monitor transient vibration at different points of the skeletal system; namely, medial malleolus, tibial tuberosity, iliac crest, and forehead. These particular locations were chosen since they are covered by only a thin layer of soft tissue. This investigation utilized a novel approach, by which small low-mass PCB piezoelectric accelerometers were placed on the skin surface at the above mentioned locations. They were externally fixed to the human subject with adhesive tape and a tightly wrapped elastic ace bandage to give an actual bone acceleration measurement during gait. Such attachment results in the natural frequency of the recording instrumental setup to be above that seen in walking data, thus avoiding inaccurate data due to resonance.

A total of forty subjects, twenty eight males and twelve females, generally in their early 20's served as test subjects. Each subject walked barefoot eight trials, between 7 and 10 heel strikes were recorded for each trial.

A summary of the acceleration values typically attained at the various measurement points on the bodies of the many individuals tested in this study is shown in Table 1. While it appeared that most of the 40 subjects tested in this study were healthy individuals with traditional walking styles, the data acquired in this study revealed unbalanced shock attenuation capacities for 22 of the subjects tested. Unbalanced gait data was defined as the relative difference between signals resulting from right and left heel strikes being greater than 10%. This observation was rather surprising, since all the subjects tested had balanced results (less than 10% difference) between the signals obtained from their right and left ankles (as expected); the discrepancies between right and left leg data did not appear until tibial accelerograms were compared for each leg. The recorded right and left tibia readings were generally 10-15% different for these "unbalanced" subjects. The main reason for this discrepancy was that one leg was more developed than the other. This is reasonable since a more developed leg would have denser muscle and soft tissue than a less developed one, which would help to attenuate the heel strike-generated shocks even more. This consideration was confirmed when all of the 22 "unbalanced" subjects which were tested acknowledged that they preferred to use the particular leg with the better shock attenuation capacity more than their other leg, for leaning on, kicking with, pushing off, etc. Thus, the tendency to use a favored leg would likely lead to an uneven development of each leg and would affect each leg's shock absorption capacity.

Table 1. Shock Attenuational Capacity of the Human Musculoskeletal System

<u>Recording Location</u>	<u>Heel Strike (Mean+S.D.)</u>	<u>% Attenuation</u>
Ankle	5.09 + .61 g	n/a
Tibia	2.54 + .28 g	50.1
Hip	0.95 + .09 g	81.1
Forehead	0.61 + .07 g	87.8

SEXUAL DIFFERENCES OF THE NORMATIVE TYPOLOGY OF THE RACHIS CURVES AND ANALYSIS OF SPHEROSOMATOGRAMS OF ♀ OLYMPIC CRAWL AND BUTTERFLY SWIMMERS
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Applying of the Improved Radius Method with Intersection Point on 190 Female, and 286 Male students, the relation between the length of the dorsal curve to the length of the lumbar curve was two to one (2.1 to 1 for F, 2.2 to 1 for M) but the relation of the height of dorsal curve to the height of the lumbar curve was 1.7 to 1 for F, and 2.4 to 1 for M. Statistical analysis indicates that 82.6% of F, and 80.3% of M, in this sample constitute the "Type Normal A" of the biological development of the rachis. On the other hand for 7.9% of the F, and 11.5% of M, the relation between the length of the curves was one to one (1.1 to 1 for F, 1.2 to 1 for M), but the relation of the height of the dorsal curve to the height of the lumbar curve was 1.7 to 1 for F, and 2.4 to 1 for M. The curves with these characteristics constitute "Type Lordotic B". Moreover for 9.5% the F, and 9.6% of the M, the relation between the length of the curve was about four to one (3.9 to 1 for F, and 4.6 to 1 for M), and ratio of the dorsal height to the lumbar was 3.6 to 1 for F, and 8.0 to 1 for M. The curves of rachis with these characteristics constitute "Type Kyphotic C". This classification called "Normative Typology" is only a statement of facts allowing the analysis of rachis curve according to the sexes.

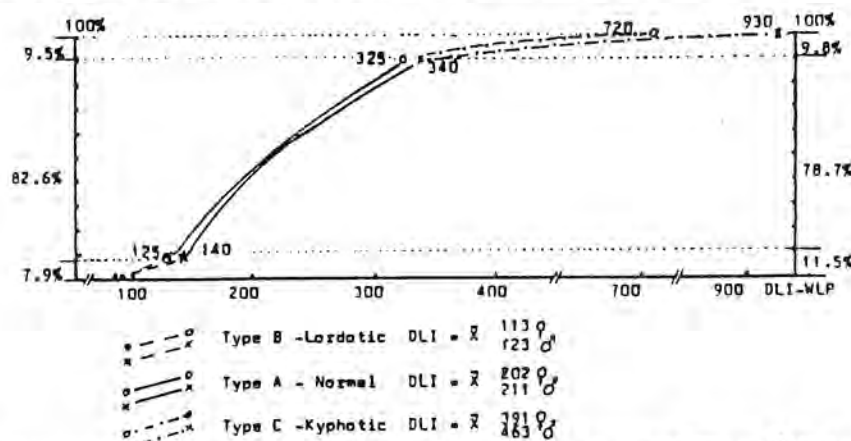


Fig.1. Cumulative Frequencies of Dorso-Lumbar Index concerning ♀ (190) and ♂ (286)

The comparison was done taking into consideration:

Four Indexes for whole curve of Rachis: 1. Dorso-Lumbar Index: relation between length of dorsal and lumbar parts; 2. Curve Relative Index: relation between heights of dorsal and lumbar parts; 3. Relative Summation Index: ratio of sum of both heights to the total length of the rachis; Inclination Index: inclination of the rachis to the vertical.

Three Indexes for the Dorsal part: 5. Dorsal Top Index: position of the top of the dorsal curve; 6. Dorsal Curve Index: relation between height and length of the dorsal curve. 7. Radius Dorsal Curve: size of the radius of the dorsal curve.

Three Indexes for the Lumbar part: 8. Lumbar Top Index: Position of the top of the lumbar curve; 9. Lumbar Curve Index: relation between height and length of the lumbar curve. 10. Radius Lumbar Curve: size of the radius of the lumbar curve.

The means of Female and Male (Fig.2) concerning the relation of the length of the curves DLI are similar (209-222). Sizes of the radius of the dorsal curves RDC are the same (462mm-459mm). An important difference appears in the relation of the heights of the curves. This ratio CRI was 168 for F, and 239 for M. This is related to the ratio between height and length in the dorsal curve DCI, which is smaller in F (9.1) than in M (10.4), and in the lumbar curve LCI where F reaches 11.4 against M 9.4.

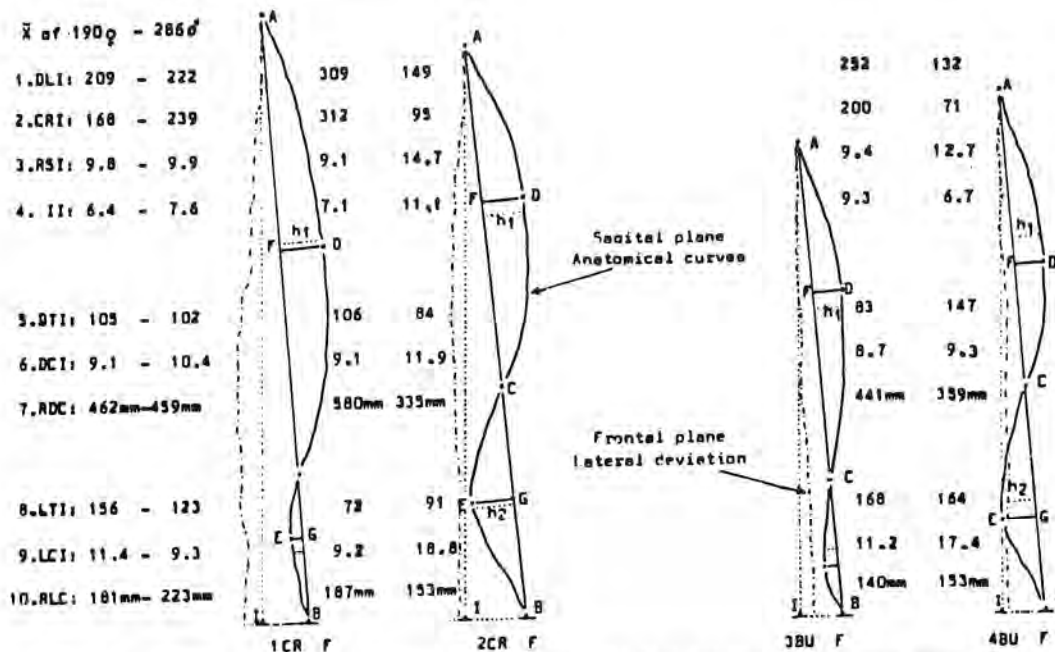


Fig. 2. Means of indexes of Rachis curve of 190 Female and Male subject and Spherosomatograms of selected Olympic Female Crawl and Butterfly swimmers.

The spherosomatograms of four Female swimmers 1CR, 2CR crawl specialist, 3BU, 4BU butterfly specialist is presented. The relation between the dorsal and lumbar curves in length DLI as well as in height CRI show that the data are specific to each subject -due to adaptation in order to reach a better performance- even in the same stroke pattern and although all the swimmers belong to the group Type Normal A. 1CR and 3BU evaluate toward Type Kyphotic C, but 2CR and 4BU do it toward Type Lordotic B.

The crawl swimmers 1CR and 2CR breath on the right side, in the frontal plane their rachis veers away from the vertical (A---I) on the left side. But for 3BU and 4BU the dorsal part of the curve veers away on the right from the vertical (A---I) most likely because the swimmers turn in the sagittal and horizontal planes in water environment without gravitation.

Spherosomatography of the Rachis curve and Interpretation by Normative Typology with 10 indexes, show the high complexity of the spine - a most dynamic part of the body. Analysis taking in consideration the female and male separately by study of spherosomatograms in sagittal plane which shows anatomical curves (lordosis, kyphosis), and in frontal plane for lateral deviation (scoliosis) could contribute to understand the how and why of the changes of the rachis curves under the influence of the way of the life (sport, occupation ...).

Kinematic Variability of the Tennis Forehand Drive

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It is not unusual for research in sports biomechanics to analyze one superior trial of a performer, when intra-subject variability is unknown, on the assumptions that variability is low and the superior trial is representative. Recently, there has been increased interest in human movement variability. Hatze (1986) proposed an generalized measure of movement variability, and several researchers have begun to compile data on the variability of several kinetic variables (Winter 1984; DeVita and Bates 1988). The present study examined the intra-subject variability of upper extremity kinematics in the tennis forehand drive.

Method

Two male varsity tennis players stroked their normal flat forehand drive down the center of a tennis court off balls projected from a ball machine. New tennis balls were projected with a trajectory, velocity ($20 \text{ m}\cdot\text{s}^{-1}$), and frequency similar to skilled tennis play. Two cameras running at 100 and 200 fps recorded the movement and the DLT method was used to reconstruct the motion in three dimensions (Mean error 4.1 mm). Rounded reflective markers were placed on the racket tip, racket CG, hand CG, wrist, elbow, and shoulder. One stroke was filmed in each of fifteen rallies performed by subject MD. Subject ES was filmed for eight normal strokes, and eight strokes from rallies using a grip significantly firmer than his usual grip. Grip forces were measured by two sensors on the racket. Reconstructed position data were smoothed with a quintic spline (Woltring 1986). Curves were synchronized to impact and averaged. Curve Coefficients of Variation (CCV) were calculated using the formula proposed by Winter (1984).

Results and Discussion

Ball velocity prior to impact in the 31 filmed strokes was $12.0 \pm 0.5 \text{ m}\cdot\text{s}^{-1}$ (CV = 4%). The forehand movements performed in this stroking condition proved to be quite consistent. The resultant velocity of the racket CG prior to impact for both subjects ($14.2 \pm 0.9 \text{ m}\cdot\text{s}^{-1}$) showed similar variability (CV = 3.4 and 6.6%). Both subjects exhibited similar patterns of mean position, velocity and acceleration curves in all three dimensions. The variability of these curves, however, was not consistent.

Mean position curves were the most consistent in the direction of the stroke (Y) and in the vertical (Z) direction (CCV between 14 and 67%). In the lateral direction the mean curves were much more variable with CCV's consistently above 100%. Mean velocity curves tended to have CCV's similar to or smaller than the corresponding position curves.

Linear accelerations for all markers tended to be most consistent in the Y dimension ($31.8 \pm 13.2\%$), but in contrast to position were followed by the X ($42.1 \pm 13.1\%$) and Z ($44.4 \pm 20.2\%$) dimensions. The least variable mean linear acceleration curve for both subjects was the Y acceleration of the wrist (CV = 27.0 and 17.1%). The mean and SD curves for subject MD are presented in Figure 1. For all dimensions the movement strategy employed was to create consistent wrist motion, with more variable racket and upper extremity motion.

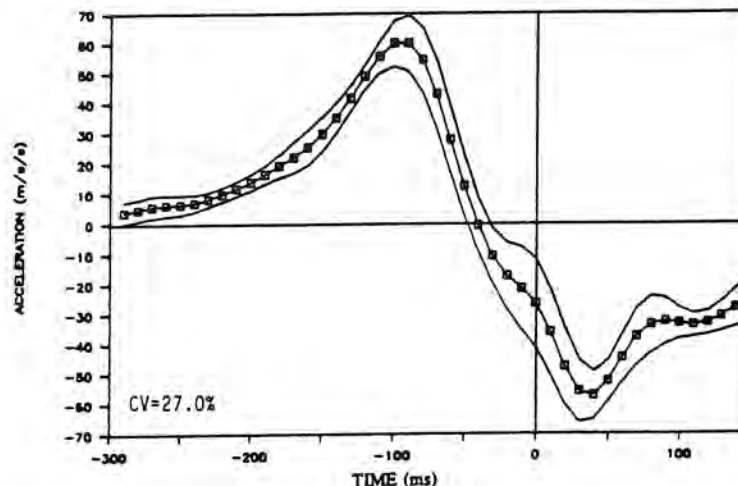


Figure 1. Mean Y Acceleration of the Wrist (Subject MD).

Increasing grip force for subject ES did not significantly influence the shape or magnitude of the position, velocity or acceleration curves. The mean firm grip curves did tend to have smaller variability than the normal grip means. It was concluded that small samples of trials were adequate to establish mean data for these two subjects in this condition. The Curve Coefficient of Variation is a relative measure of variability in curves which may be useful in biomechanical and motor control studies.

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RECENT ADVANCES IN ANALYSES OF THROWING

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Recently there has been increased interest in computer modelling of sports maneuvers. This interest has been stimulated by dramatic improvements in modelling techniques and by increased interest in improving performance, reducing injury, and improving training. In this paper we present results of recent studies of throwing. We employ the finite-segment modelling procedures commonly used in multibody dynamics analyses.

The arm is modelled as a three-body system consisting of the upper arm, the forearm, and the hand. There are three degrees of freedom at the shoulder, one at the elbow, one in the forearm and two at the wrist, giving a total of seven degrees of freedom. At most six degrees of freedom are required to throw an object. Hence, the arm has a redundant degree of freedom. This redundancy can be used to optimize the throwing maneuver.

The governing dynamical equations are developed using Kane's equations as expositied by Kane and Levinson (1980, 1985) and by Huston et. al. (1978). Constraint equations are developed from the desired launching motion of the thrown object. The governing equations thus consist of the combined dynamics and constraint equations which can be written in the matrix forms:

$$A\ddot{x} = f + \hat{f} + \tau \quad \text{and} \quad B\dot{x} = g \quad (1)$$

where A is in general a 7x7 generalized mass array x is a 7x1 column array of joint angles, f is a 7x1 column array of inertia forces, $\hat{f}$ is a 7x1 column array of generalized constraint forces, τ is a 7x1 column array of joint torques, B is in general a 6x7 constraint equation matrix and g is in general a 6x1 forcing function array. Explicit and illustrative expressions for these arrays have been recorded by Huston et. al. (1978) and by Wang and Huston (1987). wang and Huston (1987) have shown that the generalized constraint force array is related to the constraint matrix B by the relation

$$\hat{f} = B^T \lambda \quad (2)$$

where B^T is the transpose of B and λ is in general a 6x1 array of constraint force and moment components.

In general Equations (1) represent a set of thirteen equations for the seven joint angles, the seven joint torques, and the six constraint force and moment components. Hence, in general an additional seven equations are needed to obtain a unique solution. These equations can be obtained from optimization considerations such as minimum energy considerations or minimum joint torque considerations or weighted means of these.

Our current research implies that the joint torques should be proportional to the limb cross-section area and that for optimal control the large limbs (upper and forearm) should move before the small limb (hand and fingers).

Illustrations of the solutions of Equations (1) for simple dart throwing are presented by King and Huston (1985).

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COMPARISON OF MAXIMAL SPRINT RUNNING AND SPRINT SPECIFIC STRENGTH EXERCISES

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INTRODUCTION. In running the most forces are exerted during the contact phase. In order to increase the strength level the runner utilizes many kinds of exercises including weight lifting and jumping exercises in addition to running exercises. The present study was designed to examine sprint specific strength exercises (bounding and supramaximal sprinting) and to compare them to normal maximal sprinting. It was believed that this kind of analysis would give further information about the biomechanics of the exercises with possible practical applications for training.

METHODS. Eleven male sprinters also experienced in supramaximal sprinting served as subjects. Their age was 21.5 ± 2.4 yrs (mean + S.D.); height 1.83 ± 0.07 m; body mass 76.1 ± 6.4 kg, body fat (Durnin and Rahaman, 1967) 10.6 ± 1.5 %; and 100 m record 10.79 ± 0.13 s.

The subjects performed, in an indoor hall, three maximal runs (45 m), three supramaximal runs (45 m) with horizontal towing (Mero and Komi, 1985) and three maximal bounding exercises (45 m). The bounding exercise is very similar to sprinting but the bounding stride is longer. The runs and bounding exercises were filmed during the phase of uniform speed with a Locam 51-0003 camera set to operate at 100 fps. During filming the subjects ran over a long (10 x 1.2 m) force platform (Komi, 1985), which was covered with a Tartan mat. Electromyographic (EMG) activity was telemetered from the m. gastrocnemius cap.lat. (GA), biceps femoris cap.long. (BF), gluteus maximum (GM), rectus femoris (RF) and vastus lateralis (VL) (see Mero and Komi, 1987a).

RESULTS. The main results of the present study are presented in table 1. In EMG activity the three analyzed exercises were very similar and no differences were found between them in any part of the stride cycle.

DISCUSSION. Supramaximal velocity was 7.7 % greater and bounding velocity 18.2 % lower than velocity in normal maximal sprinting. Stride length in supramaximal sprinting contributed 6.4 % of the observed increase and these results are in line with the earlier observations (Mero and Komi, 1987b). In bounding, the stride length was 24.7 % greater than in normal running. Contact time, during which the forces are exerted, was clearly greater in bounding than in the other exercises. The resultant force in the bounding exercise during braking was 21.4 % greater than in maximal sprinting, which is near by the same as the difference in stride length. However, in the propulsion phase there were no differences in the resultant

forces. But in the resultant force impulses the bounding exercise had the greatest values in both braking and propulsion phases, which were 28.3 - 58.3 % greater than in the running exercises.

TABLE 1. Comparison of maximal sprinting, supramaximal sprinting and bounding in velocity and force^a production (mean $\pm$ SD). The observed significant differences between exercises are shown (t-test).

Variable	Maximal sprinting		Supramaximal sprinting		Bounding		Maximal sprinting vs bounding
Velocity ($m \times s^{-1}$)	9.97 $\pm$ 0.28	***	10.74 $\pm$ 0.37	***	8.16 $\pm$ 0.32	***	
Stride length (m)	2.19 $\pm$ 0.13	**	2.33 $\pm$ 0.14	***	2.73 $\pm$ 0.35	***	
Stride rate (Hz)	4.55 $\pm$ 0.19		4.61 $\pm$ 0.24	***	2.99 $\pm$ 0.26	***	
Contact time (ms)	105 $\pm$ 6		101 $\pm$ 8	***	134 $\pm$ 7	***	
Braking time (ms)	46 $\pm$ 11		44 $\pm$ 6	***	60 $\pm$ 7	**	
Propulsion time (ms)	59 $\pm$ 6		57 $\pm$ 9	**	74 $\pm$ 7	**	
Flight time (ms)	125 $\pm$ 6		127 $\pm$ 11	***	198 $\pm$ 20	***	
Resultant forces							
Braking phase (N)	2080 $\pm$ 158	*	2257 $\pm$ 181	**	2526 $\pm$ 264	***	
Direction (deg)	80 $\pm$ 2		79 $\pm$ 3		80 $\pm$ 2		
Propulsion phase (N)	1565 $\pm$ 251		1617 $\pm$ 243	*	1597 $\pm$ 246		
Direction (deg)	76 $\pm$ 2	*	70 $\pm$ 5		76 $\pm$ 2		
Resultant force impulses							
Braking phase (Ns)	96 $\pm$ 28		99 $\pm$ 38	***	152 $\pm$ 46	***	
Propulsion phase (Ns)	92 $\pm$ 22		92 $\pm$ 28	**	118 $\pm$ 39	**	

^a forces are average forces during the phase and body weight is included in vertical forces

* 0.05

** 0.01

*** 0.001

In conclusion, the bounding exercise is a very good strength/power exercise for sprinting, because training stimulus (force impulse) for strength is high and muscle activity is near that of sprinting. Supramaximal sprinting as an exercise is powerful, because velocity, stride length and also stride rate can be increased.

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ELECTROMYOGRAPHIC AND POWER CHARACTERISTICS IN VARIOUS DROP JUMPS

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INTRODUCTION. Drop jumps have been studied by analyzing the force production during contact and the height of the rise of the center of gravity (e.g. Asmussen and Bonde-Petersen, 1974; Komi and Bosco, 1978) and the potentiation of the electromyographic (EMG) activity in the braking and propulsion phases during contact (e.g. Viitasalo and Bosco, 1982; Bosco and Viitasalo, 1982; Viitasalo, 1984). The present study was designed to investigate in more detail the EMG activity and power characteristics in normal drop jumps and also in a "lightened" drop jump where a subject hangs with rubber ropes during jumping. The lightened drop jump has been used in training to develop strength/power characteristics of athletes.

METHODS. Seven male sprinters experienced in normal drop jumping served as subjects. Their age was 23.1 ± 2.8 yrs (mean $\pm$ SD); height 1.81 ± 0.06 m; body mass 75.8 ± 5.5 kg; and 100 m record 10.89 ± 0.09 s. The subjects performed three series of maximal drop jumps (from a height of 0.55 m) consisting of three normal jumps, three lightened jumps and three normal "posteffect" jumps. In the lightened conditions the rigidity of the rubber ropes was controlled so that the mass of a subject was lightened by 10 % when standing on a force platform. EMG activity was telemetered from the m. gastrocnemius cap. lat. (GA), biceps femoris cap. long. (BF), gluteus maximum (GM), rectus femoris (RF) and vastus lateralis (VL) and integrated (IEMG) in 50 ms periods. An angular displacement signal was picked up by an electrical goniometer attached to the right knee. Mechanical power of the positive work phase per body mass was calculated according to Bosco et al. (1983). All three performances in each jumping condition were averaged for every subject.

RESULTS AND DISCUSSION. The average knee angular displacement ranged between 48-51 degrees showing no systematic changes between the various jumps. The height of the rise of the center of gravity was highest (0.47 ± 0.03 m) in the lightened drop jump. The value was greater ($p < 0.01$) than the respective values in the normal (0.43 ± 0.02 m) and in the posteffect (0.43 ± 0.02 m) jumps. Mechanical power was 6.8 % greater in the lightened jump₁ than in the normal jump (53.1 ± 7.1 Wkg⁻¹ and 49.7 ± 6.2 Wkg⁻¹ respectively) but the difference was not significant. There were no differences in contact times and the difference (9.3 %) observed in the rise of the center of gravity (flight time) is primarily due to the lifting

effect of the rubber ropes.

Preactivity (IEMG) values in the measured muscles ranged from 27 to 63 % calculated from the maximal EMG activity during the contact phase. No differences were observed between the various jumps. At the beginning of the braking phase the GM muscle was the only muscle that increased (20 %; $p < 0.01$) activity in the lightened drop jump. The other four muscles decreased their activity slightly in that jump. In the maximal activities during contact RF and GM had slightly greater values in the lightened jump compared to the normal jump, whereas the other muscles slightly decreased their activities. At the end of the contact phase RF and VL nonsignificantly increased activity in the jump with the rubber ropes whereas the other muscles decreased their activities (GA 27 %; $p < 0.01$, GM 22 %; $p < 0.01$, BF 9 %; ns). The normal drop jump and the posteffect jump had the same muscle activity model during contact.

In conclusion, the rubber ropes raise the jumper in the lightened drop jump higher than in the normal drop jump. Muscle coordination in the lightened jump compared to the normal jump changes during the contact phase and especially at the end of the propulsion phase there are great decreases in EMG activities of the GA and GM muscles.

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FIVE MONTH IN VIVO MEASUREMENT OF HIP JOINT FORCES

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INTRODUCTION: A telemetrized, inductively powered hip prosthesis [1] allows long-term, safe in vivo measurement of three-dimensional joint forces, real time monitoring and recording of video-images and force signals together. Measurements on several patients will complete existing data and feasible improvement of implants. After three-year tests in sheep, a very active 82-year old patient got instrumented implants for the left (May 88) and right (Aug.88) hip joint. This is a preliminary report from more than 20 hours measurements, valid only for this patient.

COORDINATE-SYSTEM: The resultant force R and its components X, Y and Z act on head of prosthesis and are measured in a femur-based system. -Z acts from proximal, parallel to the femoral axis, -X from medial and -Y from ventral. Wzx is the angle between -Z and R in the frontal plane, Wxy between -X and R in the transversal plane.

GENERAL OBSERVATIONS: R mostly varies only slightly in direction for periodic activities, fast movements, high loads and vice versa. It typically acts from proximal--medial--ventral upon the head. At loads over about 100%BW (bodyweight), the direction Wzx in the frontal plane stays remarkably constant between 10 and 25 deg., mechanically a compromise between bending of neck resp. distal femur. The component -Y always stays small, but varies more. Wxy has a tendency to be in the range of the neck's natural antetorsion, decreasing stem- and femur torsion. This indicates the importance of a well chosen antetorsion of prosthesis. Load directions at both joints of our first patient are very well balanced by muscular control, a finding which can not be generalized, however.

PHYSIOTHERAPY, (first 2 p.o. weeks, numbers /.../ from table): During passive movement of the leg, R is always small /1/. Combined passive adduction and inner rotation created no danger of luxation /2/. Active movements in a lying position can result in forces up to 150%BW in the joint /3/. When lifting the straight opposite leg while lying on the back, 150%BW were observed /4/. Working against resistance produced up to 250%BW /5/. Lifting the pelvis when lying on the back caused up to 250%BW /6/.

STATIC POSITIONS & SLOW MOVEMENTS (3 w.p.o. and up): Symmetric two-leg-stance resulted in forces bigger than the expected 33%BW /7/. The wide variation of Wxy indicates antagonistic muscle balancing. Lifting on toes gave force an increase to 112%BW /8,9/. One-legged-stance caused magnitudes as high as during slow walking /10/. Standing on both legs and bending the upper body by 80 deg., resulted in 230%BW with small Wzx /11/. During sitting and rising from a 47 cm high chair, for-

ces up to 107%BW acted in the joint /12,13,14/ with small Wzx and Wxy.

WALKING: In practice, two crutches remarkably reduced the load only during the first p.o. days /15/. With improved walking ability and overall physical condition, loads were high /16/. After some weeks occasionally 250%BW were observed. One crutch normally reduced the joint loads by not more than 25% and only for short distances /17/. Unaided walk lead to loads between 245% and 330%BW /18..21/, stronger varying from one day to the other and depending from the p.o. time than from the walking speed.

range(*)or

peak val. R [%BW]	range of Wzx[degr]	range of Wxy[degr]	No. Activity
30...50			/1/ Physiother., Pass.movements
80	7	30	/2/ " Adduct.&inner rotation
100..150	10 .. 17	10 .. 26	/3/ " Active movements
140..150	14 .. 16	15 .. 26	* /4/ " Lifting opps. leg
250			/5/ " Against resistance
210..250	10 .. 17	0 .. 14	* /6/ " Lifting pelvis
60...80	4 .. 12	-35 .. 35	* /7/ Symmetr. two-legged-stance
112	13	-10	/8/ Rising on toes
74...88	3 .. 7	-21 .. 23	* /9/ Standing " "
220..310	18 .. 21	-11 .. 35	/10/ One-legged-stance
190..230	9 .. 11	17 .. 20	* /11/ Bending upper body
18.. 30	32 .. 36	5 .. 6	* /12/ Sitting on chair
73..107	19 .. 23	17 .. 34	* /13/ Starting chair rising
40.. 46	20 .. 37	-22 .. 14	* /14/ Stopping " "
85..150	14 .. 23	5 .. 25	/15/ 2 crutches, 8 d.p.o
190..200	14 .. 20	20 .. 27	/16/ 2 " after 10 "
<25% reduction			/17/ 1 " from 2 w.p.o
..280	.. 16	.. 20	/18/ no aid,slow walk 5 "
245..255	19 .. 21	11 .. 18	/19/ " " " 13 "
260..310	20 .. 23	-5 .. 17	/20/ " " " 16 "
270..330	22 .. 25	-4 .. 18	/21/ " " fast walk 17 "

COMPARISON WITH LITERATURE: Calculated joint forces of other autors were smaller for the two-legged stance. This might indicate the importance of antagonistic muscle activities for balancing static positions and slow motions. Forces during walking were higher than calculated by others, so optimization functions may have to be improved or muscle lever arms in the mathematical models might be incorrect. Our results compare well with the data measured in [4,5]. In contrast to [2] we found small forces in swing phase of walking. Moderate loads acted during chair rising, for which high joint pressure were reported in [3].

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LOAD DIRECTIONS AT HIP PROSTHESES MEASURED IN VIVO

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INTRODUCTION: A telemetrized, inductively powered hip prosthesis [Bergmann, G. et al., J. Biomech. 21 (1988) 169-176] allows long-term, safe in vivo measurement of three-dimensional joint forces, real time monitoring and recording of video-images and force signals together. Measurements on several patients will complete existing data and feasible improvement of implants. A very active 82-year old patient got such implants at the left (May 88) and right (Aug. 88) hip joint. This is a preliminary report from more than 20 hours measurement with this patient.

COORDINATE-SYSTEM: The resultant force R (Fig.1) and its components X , Y and Z act on head of prosthesis and are measured in a femur-based system. $-Z$ acts from proximal, parallel to the femoral axis, $-X$ from medial and $-Y$ from ventral. W_{zx} is the angle between $-Z$ and R in the frontal plane, W_{xy} between $-X$ and R in the transversal plane.

INFLUENCE OF LOAD DIRECTION ON BENDING AND ROTATION OF PROSTHESIS AND FEMUR: By an increase of W_{zx} , neck and prox. femur are bended less in the frontal plane, bending of dist. femur may be higher. Torsion of prosthetic and femoral shaft as well as additional neck bending in the transv. plane become minimal if W_{xy} is directed towards the antetorsion of the neck.

SELECTED MEASUREMENTS: Except for very small R , as in the swing phase of walking (Fig.4), $-Z$ always dominates the other components, indicating load directions close to the femoral axis. Average ratios $X/Z=0.65$ and $Y/X=0.80$, as seen during leg flexion (sec.4 to 8 of Fig.2), were the highest observed so far and represent the steepest angles rel. to the femur ($W_{zx}=33$ deg., $W_{xy}=39$ deg). At higher joint forces (Fig.3) and longer postop. time (Fig.4), the correlation between $-X$ and $-Z$ increases and the load direction W_{xz} in the frontal plane becomes nearly invariable. After some month and when R is bigger than $50\% \cdot R_{max}$ in the stance phase of walking, W_{xz} is in the range of 19 to 25 degrees. The variation between $-X$ and $-Y$ is bigger, and therefore that of W_{xy} in the transversal plane. We found W_{xy} between -35 and $+36$ deg. For high joint loads during walking, this range was -5 to 20 deg. (Fig.4). The load direction in the transv. plane obviously has a tendency towards the antetorsion of the femoral- or prosthetic neck.

These findings indicate that load directions at the hip joint are very well controlled by muscle activities. By this a good compromise is achieved between stresses in different bone areas, caused by bending and torsion. This outlines the importance of restoring the individual geometry of the proximal femur by the implantation, particularly the antetorsion angle.

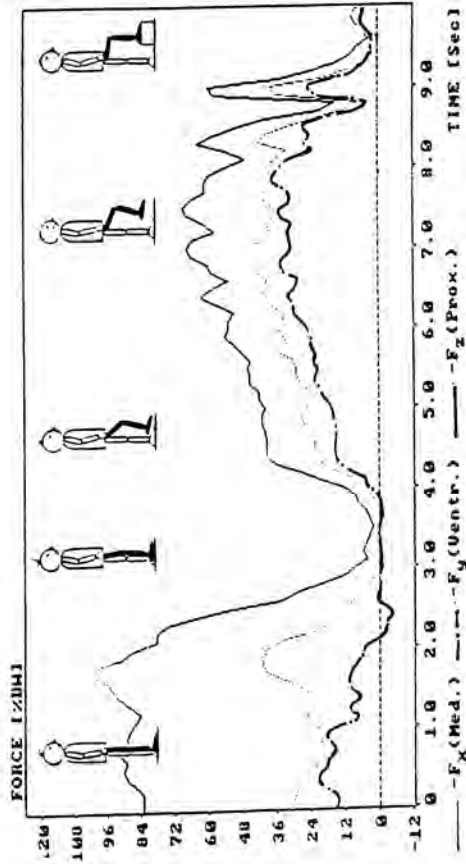


Fig.2:

Force components left leg, 12 weeks p.o. (BW= 630N)
Loading: two leg stance -- leg taken from ground --
thigh bent & lifted -- foot on 30-cm-step

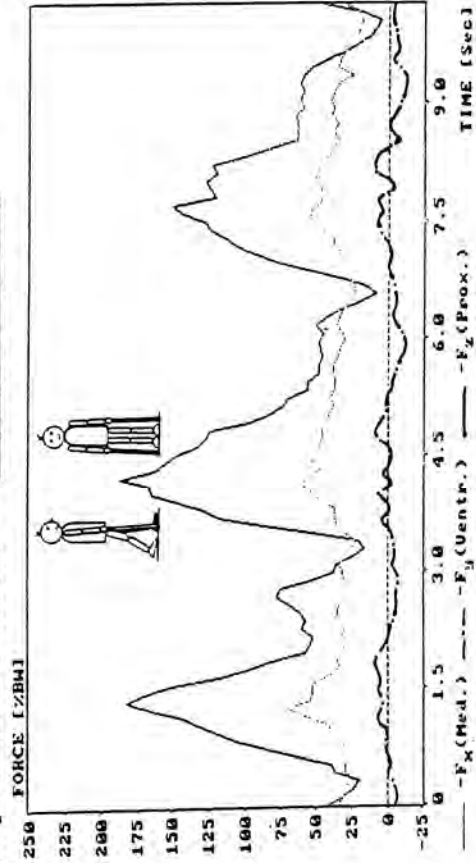


Fig.3: Force components right leg, 6 days p.o.
Slow walking with two crutches, appr. 0.3 m/s

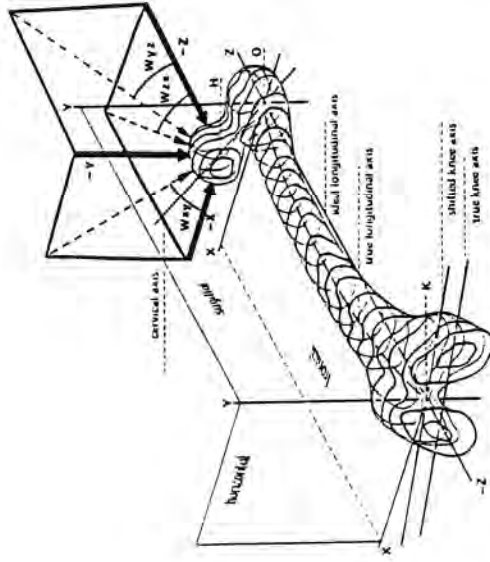


Fig.1: Coordinate system, force components -X, -Y and -Z are chosen, because their values are mostly positive.

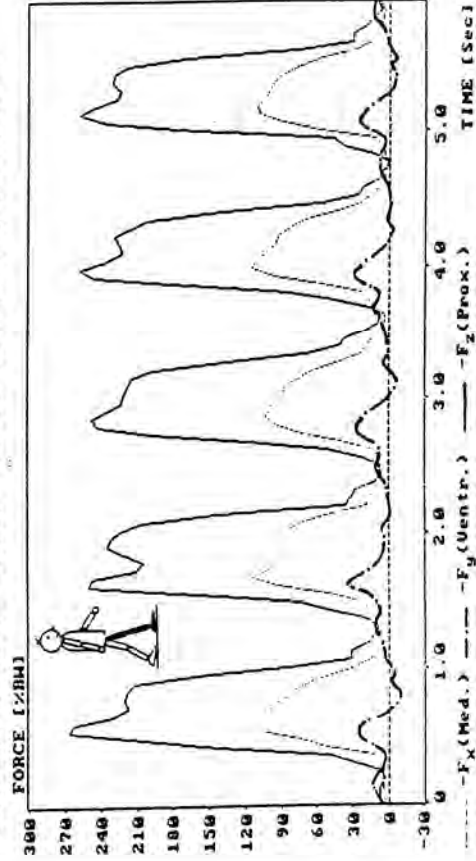


Fig.4: Force components left leg, 17 weeks p.o.
Free walking, appr. 1.2 m/s

THE INFLUENCES OF GRIP TIGHTNESS ON
HAND-ARM'S DYNAMIC CHARACTERISTICS

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Among the local vibration to which human body has subjected, the vibration caused by hand-held vibrating tools is the most frequently encountered one and brings about more harms to human body. Reynolds D.D (1972) et al pointed out: "For normal arm positions used for the operating vibrating power tools, the arm position had little influence upon the vibration response of hand." But few researchs about grip tightness influences have been reported.

In this paper, we present the experimental method to measure the changes of grip tightness and the hand-arm's dynamic characteristics (HADC), and give the preliminary measuring and analyzing results. The experimental scheme: during experiment, strength of grip is chosen as one of the five given values, while the frequency and acceleration of vibration input to hand changes continuously within 5-50HZ and $3-30\text{m/s}^2$, and the HADC is expressed by response in acceleration. We obtain the following results: 1) The angle between forearm and uparm influences the HADC. 2) The varying of grip tightness can change the HADC, generally, the difference is about 20%, the maximum 64%. 3) The vibration responses of hand are not in proportion to the strength of grip, for example, when the strength of grip is 1Kg and the frequency of vibration 18HZ, the response in the joint of thumb's phalanges is the highest. 4) With the changes of the grip tightness and the frequency of vibration, in certion scope the hand-arm's soft tissue will shake obviously, for example, when the strength of grip is 5Kg and frequency 22Hz, the response in the musculus brachioradialis near articulatio cubiti increases from 6m/s^2 to 15m/s^2 , and a resonance peak occures.

From those results, we have made sure that the grip tightness have influences on the HADC, especially on soft tissue. Those results will have an active significance when we deeply study the mechanism, the prevention and cure of occupational diseases caused by local vibration.

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MORPHOLOGICAL REACTION TO STRESS ACTING ON THE SHOULDER
AND ACROMIOCLAVICULAR JOINTS IN GYMNASTS

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Many gymnasts complain about pain in the shoulder region which is specially noticeable during practice periods on the rings. A little while ago Bodem and his co-workers (1985) were able to show that exercise on the rings can result in the shoulder-joint having to bear up to five times the weight of the body. We therefore used CT densitometry (Müller-Gerbl et al. 1988) to examine stress on the shoulder and acromioclavicular (AC) joints more closely by calculating subchondral bone distribution in the glenoid cavity, and beneath both surfaces of the AC joint, in gymnasts and in ordinary people. According to Pauwels (1980), the distribution pattern is able to provide information about the local adaptability of the joint, and particularly about the nature of the stress (direction of the resultant force). This is specially important for the AC joint, the function of which is still far from being understood.

MATERIAL:

CT datasets (section thickness 2 mm) from both shoulder and AC joints of 11 top German athletes (age range: 16 - 29 years) who had completed at least 5 years professional training, and 15 ordinary people (age range: 23 - 46 years; 9 male, 6 female).

METHODS

1. CT densitometry: determination of the boundaries between areas of equal density, expressed in Hounsfield Units (HU) and obtained with the radio-therapy programs EVA1 and SIDOS TELE.
2. False colour-display pictures built up with the help of the picture-analyser IBAS 2000 (Siemens, Erlangen).
3. Surface display of the bone density distribution beneath the surfaces of all joints examined.

RESULTS

1. The bone density of both the shoulder-cup and the AC joint was significantly higher in the gymnasts than in ordinary people.
2. Whereas the highest bone density was found in the central region of the glenoid cavity in ordinary people, in 2/3 of the gymnasts the highest value appeared at the back of the cup.
3. With one exception, the highest subchondral bone density was found in both athletes and controls in the acromial half of the joint.
4. In the normal subjects, the greatest degree of mineralisation was found in the middle part of either surface of the AC joint; in the athletes, on the other hand, it was almost always more pronounced posteriorly.
5. The athletes showed a correlation between the distribution patterns in the glenoid cavity and in the AC joint. In those cases in which the greatest density was found at the back of the shoulder-cup, it was also found in a similar position in the AC joint.

DISCUSSION

Our findings quite clearly show that severe stress over a long period of time (as is, for instance, experienced by athletes) eventually produces an increase in local bone density in those joints which we have examined. The distribution

pattern of the subchondral bone provides information about the quality of the principal stress; that is to say, the prevailing position of the point of action of the resultant load. A peripheral location of this point with consequent overloading of a single part of the joint surface may possibly produce pain. The nature of the transmission of the load through the AC joint appears to be dependent upon the position of its point of action at the shoulder-joint, from which point it is further conveyed either dorsally through the scapulo-thoracic "joint", or ventrally through the AC and sternoclavicular joints towards the sternum. Unphysiological stress on the shoulder joint must therefore inevitably lead to similar stress on the AC joint, which is why gymnasts frequently complain of pain there. These abnormal stresses, which appear to attack only some gymnasts, are probably attributable to insufficient training of the muscles or inadequate technique, since they are particularly likely to involve young athletes. It therefore seems reasonable to suggest that, in order to avoid injuries due to sport, certain exercises should not be undertaken until the appropriate muscles have been fully trained.

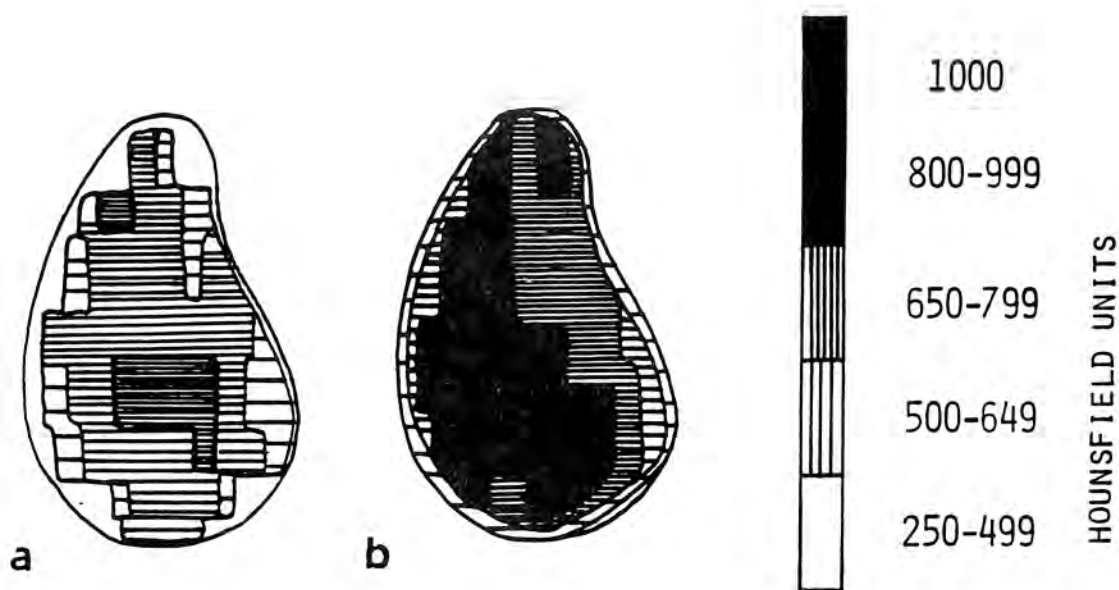


Fig. 1: Lateral view of a right glenoid cavity showing surface distribution of subchondral bone density (copied from a false-colour display)
a.) Control
b.) Gymnast

ESTIMATING THE DISTRIBUTION OF SUBCHONDRAL BONE DENSITY BY MEANS OF RECONSTRUCTED SECONDARY SECTIONS

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INTRODUCTION

We have already demonstrated in a previous investigation (Müller-Gerbl et al. 1988) that CT densitometry may be used to calculate the density distribution of subchondral bone, and therefore to judge - in Pauwel's sense - the mechanical adaptability of certain joints in the living subject.

There are several diseases of the shoulder in which it can be shown that the mechanical situation obtaining in the shoulder-cup is reflected by a subchondral bone density distribution pattern which differs from the normal. Similar changes have been described by Bentzen and his co-workers (1988) in the upper articular surface of the tibia before and after meniscectomy.

Because of certain sources of error which are inherent in the system (in particular, the "partial volume effect"), the CT sections must lie as nearly as possible at right angles to the surface of the joint. Owing, however, to the relative positions of the patient and the CT apparatus in space, it is in general only possible to obtain transverse sections; and this unfortunately does not allow the method to be applied to certain important clinical regions - of the knee and elbow-joint, for instance - for which the technique is therefore not yet available. We therefore decided to see whether the CT densitometry of reconstructed secondary sections would make it possible to assess the subchondral bone distribution with sufficient accuracy.

To check this, we compared the results of such sections with those arrived at directly.

MATERIAL

We obtained data from sagittal, frontal and transverse sections of one specimen each of a knee and an elbow-joint.

METHODS

1. The use of data from transverse sections to reconstruct secondary sections in the frontal and sagittal planes by means of the appropriate software.
2. CT densitometry of the direct sagittal and frontal sections, and the reconstruction of secondary sagittal and frontal sections by means of the programs EVA1 and SIDOS TELE.
3. The determination of unit fields of density in Hounsfield units (HU) in the single section photographs.
4. False colour representation of the density distribution in all parts of the joint surface of the primary and secondary sections.

RESULTS

A comparison of the results from the two systems showed that the distribution of the density levels in the primary and secondary sections (both for the individual sections and on the surface maps) was very similar indeed in all the specimens examined (Fig. 1).

DISCUSSION

Bringing together the results appearing in "isodensit" contour maps made from the primary and secondary sections of both specimens, makes it easy to show that CT densitometry can quite safely be used to obtain information about the distribution of the subchondral bone density in the secondary sections also. This means that, in spite of the inconvenient orientation of the apparatus, it is also possible to obtain accurate information in living patients about joints the surfaces of which do not lie perpendicular to the beam; and the availability of the method for biomechanical use can therefore be extended accordingly. Certain inherent errors of the system (partial volume effect, the production of artefacts) which may lead to inaccurate results, apply just as much to the reconstituted as to the primary sections. These errors are, however, not summated.

In our opinion, these results offer further evidence that this method of assessing the long-term mechanical results of an osteotomy or assessing its general condition under physiological stress, is very well suited to the purpose.

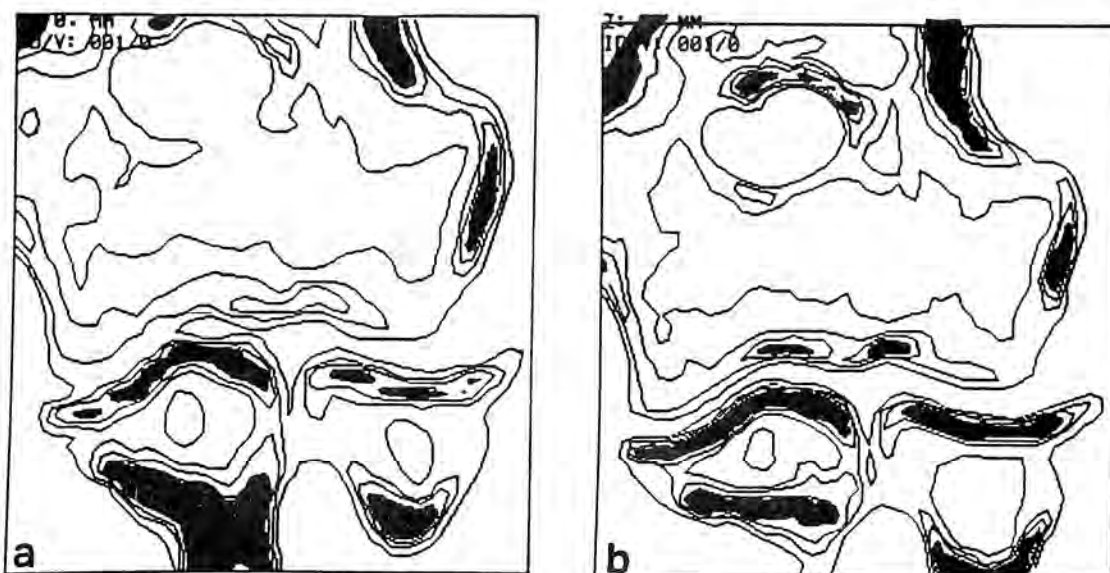


Fig. 1 Density of the subchondral bone in corresponding frontal sections of the elbow-joint. Zone of highest density black

a Result obtained from the primary dataset

b Result obtained from the reconstructed image

DYNAMICS OF THE SWING IN THE CROSS REST

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In this paper, a study of the dynamics of the human swing in the cross rest on parallel bars is presented.

Firstly, a personalized mathematical model is constructed for a human performer. The human body model consists of three links: the two arms (not including the two hands), the head and torso, and the two legs. The three links are connected to each other through hinges. These links are constructed from the segments of the human body model of Hanavan (1964). Twenty-five anthropometric measurements are used to determine the dimensions and the inertia properties of the three links. Calculations are made by a computer subprogram.

Secondly, the method of Kane (1965) is used to develop the governing equations of motion of the three-link model for the swing in the cross rest. The generalized coordinates θ_r ($r=1, 2, 3$) are shown in Fig 1. Let generalized speeds be linear combinations of θ_r ($r=1, 2, 3$)

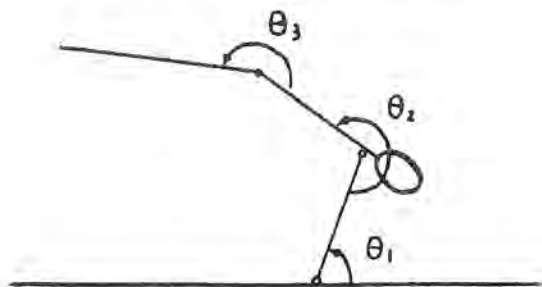


Fig 1

and be defined as follows:

$$u_1 = \dot{\theta}_1$$

$$u_2 = \dot{\theta}_1 + \dot{\theta}_2 \tag{1}$$

$$u_3 = \dot{\theta}_1 + \dot{\theta}_2 + \dot{\theta}_3$$

The governing equations of motion can be written in the form

$$\underline{P} \dot{\underline{u}} = \underline{Q} \quad (2)$$

Where $\underline{P}$ is a 3×3 matrix, $\dot{\underline{u}}$ is a 3×1 matrix, and $\underline{Q}$ is a 3×1 matrix. After the motion of the model is determined, the supporting force can be obtained using the principle of the motion of the mass center.

Four professional gymnasts were chosen as the performers. The experiments of the swing in the cross rest were recorded by filming. The smoothed film data of angles were substituted into equations (1) and (2). The time histories of the moments at the wrist, the shoulder and the hip joints, and of the supporting force were computed. The analysis of the results of the computation leads to the following conclusions:

1. In order to obtain a good takeoff, it must increase the moment of gravity as far as possible during the down swing, and correspondingly increase the reverse wrist moment at the same time.
2. It must pay attention to controlling the time of the stretch of hip to get an effective extra supporting force.

Finally, according to analyses and experiences, the joint moments computed might be modified. Then the equations (1) and (2) are solved numerically for the generalized coordinates θ_r . Thus, a modified swing motion of the model can be obtained through calculation, which can be used to compare with the original motion recorded by filming.

All calculations are incorporated into a computer program. The swing motion of the model can be displayed on the screen as animated drawing. Comparison of the original motion picture with the modified motion picture can be used to suggest guidelines for improving performance.

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THE MEASUREMENT OF FORCE AND DISPLACEMENT OF
THE TOOTH BEING CORRECTED ON LIVE BODY

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In orthodontia, various equipments are used to force tooth being corrected to move to its desired position. However, there are few publications on the relationship between the displacement of the tooth and the force exerting on the live body, neither the available method to measure them.

In this paper, we present the basic principles of the Space Three Point Location Approximation (STPLA) and L Positioner, and their experimental effects on clinical cases. The resolution of displacement in our experiment is 0.2mm. Our method has been verified by measuring and analysing the displacement and force of the tooth being corrected through out the whole course of treatment of ten clinical cases of protruding tooth. From analysing and processing the experimental results of six cases chosen randomly, we have got the quantitative data of the displacement of the tooth being corrected and force exerting on the live body. And we find: 1) The maximum horizontal displacement of the corrected tooth 1|1 is 4.2mm, the minimum one is 2.0mm. 2) The horizontal displacement of the support tooth 6|6 is within 0.2mm. 3) The average initial force needed in correcting is 299g, the average corrected force of the incisor is 150g, and the maximum 349g. 4) The STPLA and L Positioner are the feasible measurement means when we do biomechanics researchs of the tooth being corrected by a moveable correcting instrument. Clinical applications proved that the STPLA is a direct, simple, accurate and reliable method. It is worth spreading the STPLA to research area of the experimental measurement of the displacement of the tooth being corrected and the force exerting on the live body.

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A SYSTEM FOR THE ANALYSIS OF THE BIOMECHANICS OF THE KNEE IN IN VITRO TESTS

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Introduction

Important theoretical and experimental work on the biomechanics of the knee has been done to date but little is known about the effects of muscles on the joint loading. The theoretical models were mainly based on gait analysis with inherent limitations in the prediction of muscle forces. The experimental investigations frequently neglected the influence of the different muscle groups by resecting them completely or modeling only quadriceps and hamstrings in a planar fashion. As a result the knee had to be mounted in a rather rigid manner producing artefacts during loading. The fact, that no transducer was available which could be inserted permanently into the knee and record both static and dynamic joint contact pressures continuously seems to be the main reason for these difficulties.

It was therefore our goal to develop a system which would enable us to rely entirely on the ligamentous apparatus and the musculature (or its replacement) for stability (no reposition necessary) and record simultaneously external loading (eg. body weight), muscle forces and joint contact pressure.

The System

The system consists essentially of three components : pressure transducer, mounting rig and data sampling hard and software. Each pressure transducer consists of an array of 6 individual button-like, capacitive sensors per compartment. (Fig.1). Each sensor (10x0.4 mm) is pliable, able to adapt to the shape of the condyles and insensitive to changes of the electrical potential adjacent to the transducer. Their resolution is better than 3%.

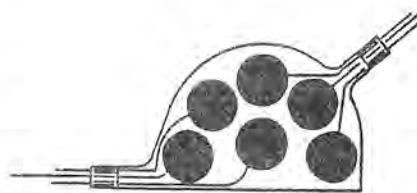


Figure 1

The complete mounting rig is shown in Figure 2. The table is mounted to the material testing machine and allows free adjustment of the knee in a horizontal plane. The base is attached to the table through four studs. It is balanced on nuts which facilitates tilting in any direction. The slots in the base allow the knee to be rotated around a vertical axis. The muscle replacements consist essentially of grip, force transducer and tensioner. Different lengths of cable compensate for the different location of the tendon insertions. The grip has fine serrations on the inside of the jaws and is tightened by a bolt going through the centre of the jaws. The force trans-

ducer is instrumented with strain gauges. The arm is bolted to the top plate through numerous predrilled holes to allow accurate, anatomical alignment of the "muscles". The top plate is interfaced to the testing machine through a frictionless planar ball-bearing.

DISCUSSION

Several problems have been identified during the use of the system. The initial set-up requires some time. All soft tissue is stripped of the bone preserving only the capsule with the ligaments and the tendinous insertions of the muscles. The bones are cast in Plaster of Paris, the grips attached to the tendons and the pressure transducer inserted into the joint and sutured to the capsule adjacent to the menisci. The transducers are inserted through a small posterior incision in the capsule leaving the oblique popliteal ligament intact.

Two errors might occur as a result of the design of the muscle replacements. Firstly, the whole structure is rigid (rather like a ligament) and does not self-adjust dynamically to a change in position. Secondly, the "muscles", in particular the long muscles of the thigh, are not restricted by their adjacent muscles and the fascial envelope of the thigh. This allows abnormal sideways movement of the muscle replacements. Finally, each position, eg. angular deformity, has to be measured and recorded individually since no fixed pre-positioning of the knee is possible due to the (desired) flexibility of the system.

CONCLUSIONS

The system allows simultaneous recording of applied load, muscle tension and contact pressure in the joint, both statically and dynamically.

A first, qualitative analysis of the data obtained indicates that the "timing" of muscle contraction during the gait cycle is of great importance in the loading of the knee joint.

Acknowledgement : This work was supported by the Medical Research Council of South Africa.

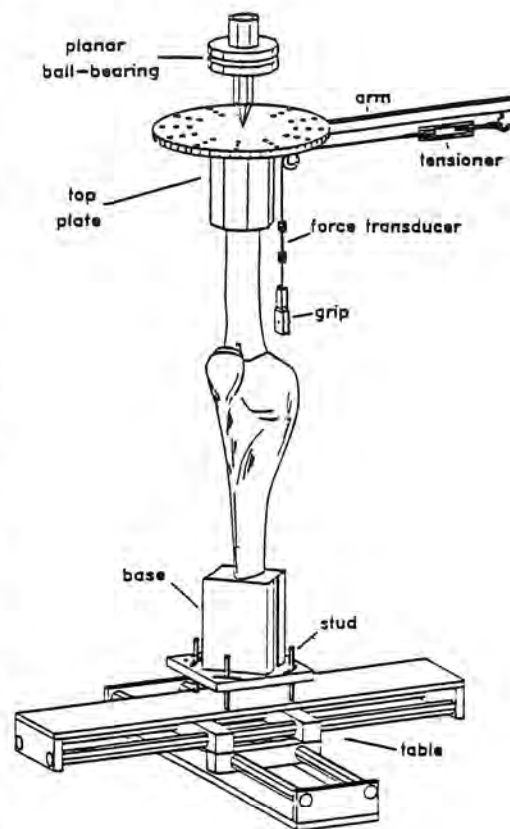


Figure 2

SPIRAL ANISOTROPIC PROPERTIES IN THE MEDIAL PART OF THE COMPACT FEMORAL BONE

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I. Introduction

At present the compact femoral bone is generally considered to be a transversally isotropic body as it follows from the measurements of elastic properties with the help of ultrasound. In a number of other works was determined that the orientation of the lamellae of the osteons, as well as the position of each osteon in a compact bone, are directly dependent upon the character of the mechanical loading of the locality. Boyde A. et al./1/, Portigliatti Barbos et al./2/ have proved that the different types of the osteons have their own precisely defined places in the osteon population, dependent upon the character of their loading.

In view of the fact that the structure of a compact bone must be closely related with the elastic properties, we have concentrated on the study of the elastic properties of the medial third of the diaphysis of human femoral bone.

II. Experimental Measurements

Samples of a compact femoral bone have been removed from two rings of the medial part of the diaphysis of two right and two left human femoral bones. Little cubes oriented in such a way that their top and the bottom superficial areas lay in the planes perpendicular to the longitudinal axis of the femur and their lateral parallel superficial areas /perpendicular to these planes/, were taken from anterior, medial and posterior walls. Electrical strain gauges of type EA-XX-062TT-120, manufactured by VISHAY Co., were pasted on the superficial areas of the cubes, and strains ϵ_{11} , ϵ_{22} , ϵ_{33} were measured under static loading. The moduli of elasticity in compression, E_1 , E_2 , E_3 , were calculated from these strains. The same sample /cube/ was then ground off so that the normal to the top area of the new cube enclosed the angle of 8 to 12 degrees with the longitudinal axis of the femur. The electrical strain gauges were again pasted on the cube's surface and the moduli of elasticity in compression, E'_1 , E'_2 , E'_3 , determined. The same procedure of the grind-off /of the original sample/ was applied to the third cube with the gradient of the normal to the top superficial area with respect to the longitudinal axis of the femur of 18 to 22 degrees. The new strain gauges were pasted on the cube's surface and the moduli of elasticity in compression, E''_1 , E''_2 , E''_3 , determined. Measurements of the moduli of elasticity in compression were simultaneously carried out on the cubes excised in close vicinity of the samples under examination, which were oriented so that the normal to the top surface of the cube enclosed the angle of 349 to 353 degrees with the longitu-

dinal axis of the femur.

III. Results

From the experimental measurements carried out it follows that the moduli of elasticity in compression in the medial part of femoral diaphysis do not reach their maxima in the directions parallel with the longitudinal axis of the femur. These moduli are highest in the direction of the tangent to the steep right-hand spiral in the right femur /with the orientation from the distal end to the proximal end of a femoral shaft/, and in the direction of the tangent to the left-hand steep spiral in the left femoral bone.

In view of the results of our experimental measurements of the highest moduli of elasticity in compression in the medial third of femoral diaphysis it is obvious that in any compact femoral bone the elastic properties are curvilinear. The femoral compact bone is curvilinearly anisotropic. The flows of the elastic properties are oriented in the direction of a very steep spiral. The curvilinear behaviour of the elastic properties corresponds with the nonlinear structure of the compact bone, and vice versa. With the primary change in loading, or a change in the orientation and the magnitude of the dominant principal stresses, the bone gets restructured so that the elastic properties and the directions of the elastic properties are conform to the directions of the dominant principal stresses.

The results of the measurements of the elastic properties of the compact femoral bone allow us to draw the conclusion that at a given point of the compact bone the principal axis of elasticity is identical with the direction of the greatest dominant principal stress, as well as with the longitudinal axis of the structural elements.

Let us conclude the paper by formulating the theorem concerning the identical orientation of the dominant principal stresses, the principal structural direction and the principal elastic properties: THE DIRECTIONS OF THE GREATEST DOMINANT PRINCIPAL STRESSES AT A GIVEN POINT OF A STATIONARILY LOADED HEALTHY FEMORAL COMPACT BONE ARE ORIENTED IDENTICALLY WITH THE PRINCIPAL STRUCTURAL DIRECTION AND WITH THE DIRECTION IN WHICH THE MODULI OF ELASTICITY IN COMPRESSION ACQUIRE THE HIGHEST VALUES.

This theorem is very important one for the design of new adaptable isoelastic hip-endoprostheses.

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MECHANICAL STABILITY OF THE HUMAN LUMBAR SPINE

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One of the main mechanical roles of the spine is to transfer the compressive force acting in the back. The spine must also allow movement between the thoracic cage and the pelvis. These two roles demands two different mechanical characteristics. First, to be able to transfer the compressive force, the spine must be stiff in order to avoid mechanical buckling. Second, the demand for movement talks for a flexible spine.

Experiments performed by Lucas and Bresler in 1961, showed that the weight bearing capacity of the spine is very small. Lateral buckling occurred at a vertical load of about 20 N. The real compressive force acting is of the order 300 N or more. Obviously the spine must be supported by its surrounding muscles.

The work presented here is focused on the question: How is the spine supported by its surrounding muscles?

A buckling analysis of a mechanical system can be made if the elastic properties of all parts included are known. The spinal system is formed by passive (skeletal elements and ligaments) and active (muscles) components. The elastic properties of the passive components, mainly the motion segments, are fairly well investigated. To be able to perform a buckling analysis, the elastic properties of the activated muscles must also be included.

From data presented in the literature, the conclusion can be drawn, that the stiffness of an activated muscle is a function of the force transferred by the muscle. For a small sudden elongation ($\epsilon < 2\%$) the relation is approximately linear. The proportionality factor between force and stiffness can be determined. The problem to be solved is therefore to find the muscle forces in the highly statically indeterminate musculo-skeletal back system.

The mechanical system is modeled as six rigid elements, five lumbar vertebrae and the thoracic cage, with the pelvis as the foundation. The rigid parts are connected to the foundation and to each other at the disk midpoints. In all there are 40 muscle fibers (modeled as straight lines from the origins to the insertions on the skeletal elements) included.

The number of muscle forces and unknown reaction forces and moments are far more than the number of available equations of equilibrium (equilibrium for each of the five lumbar vertebrae and the thoracic cage). To treat this a functional division of the system has been made in a global and a local system. The global system consists of the muscles acting directly from the thoracic cage to the pelvis. The local system consists of the muscle fibers with origin or insertion on either of the five lumbar vertebrae. Several different assumptions are made of how the forces are distributed between the two systems and also within the local system. Each assumption gives a unique set of muscle forces and thus muscle stiffnesses. Passive torque stiffnesses for the motion segments are taken from the literature. In this way, a normal engineering buckling analysis can be performed and the relation between force distribution and the mechanical

stability of the spinal system is obtained.

Calculations of possible muscle force distributions and and depth of the lumbar curve are made for symmetrical vertical loading and for the standing posture. The conditions for stability and that muscle tension levels must be below the physiological limit are considered. Only back muscle activity is assumed which corresponds to the behavior of about 75% of the subjects in an investigation by Klaussen and Asmussen.

The results show that lateral stability is maintained if the force in the global erector spinae exceeds a critical value which rises with increasing vertical loading on the shoulders. A maximum allowed global erector spinae muscle force is set by the condition that no uncontrolled extension of the spine is allowed. The condition for sagittal stability implies that the horizontal distance from the gravity line (including upper body weight and outer load) to the midpoint of the most forward disk must be greater than about 20 mm in the normal standing posture with no outer load. When 1000 N is carried on the shoulders, the distance must increase to about 40 mm.

The condition for muscle tensions below the physiological limit implies a flatter back for increased vertical loading.

From a mechanical point of view, three different autonomous stabilizing muscle systems can be identified:

1. The global system formed by the the muscles acting directly between the thoracic cage and the pelvis.
2. The sagittal local system. This is formed by the interspinal and the multifidi muscles and it gives local sagittal support to the spine.
3. The lateral local system. This is formed by the intertransverse and the quadratus lumborum muscles and local erector spinae muscle fibers from the erector spinae and the lumbar intermuscular aponeurosis to the transverse processes of the lumbar vertebrae.

Reference: Bergmark, Anders. Mechanical Stability of the Human Lumbar Spine. Thesis. Dept of Solid Mechanics, Lund Institute of Technology, Sweden, Lund 1987.

SIMULATED TEAM BOAT ROWING USING MECHANICALLY COUPLED WIND RESISTANCE ERGOMETERS

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Rowing ergometers of various types are frequently used to assess the physiological performance of rowers and to condition and train their bodies during off season. Up to now only the use of single seat rowing ergometers had been reported in the available literature. This paper describes a team rowing device assembled of Concept II rowing ergometers, which has actually been built and tested according to the following steps: (1) basic theoretical considerations with respect to the dynamics of a racing shell; (2) set up the system's equation describing its longitudinal dynamics; (3) transformation of the system equation into an equation of the equivalent rotating system; (4) comparison of the transformed system with the dynamic equation of a Concept II rowing ergometer; (5) set up the equations of a team rowing ergometer cascade; (6) design, computing and manufacture of the mechanical parts for the basic module of the team rowing device; (7) presentation of the prototype two seat rowing ergometer cascade in figure 1.

Basic Theoretical Considerations: When rowing a team boat the athletes' performance is affected by the dynamic behaviour of the racing shell and in addition by visual, acoustic, tactile and vestibular feedback due to reciprocal interactions of the team members. In order to reveal the effects of the boat's dynamics on a team member a simplified model has been used to describe its longitudinal dynamic behaviour. By looking at the mathematical models which can be found in the biomechanic literature, it becomes obvious that two major effects, which may be voluntarily changed are immediately responsible for the longitudinal motion. One of them is the sum of the propelling forces acting on the blades during the pull phase and the other one is the sum of the forces acting on the boat due to acceleration caused by rowers when moving their bodies back and forth. To come close to the feel of rowing, the physical model used for simulation must work at least according to the above mentioned simplified mathematical models and provide visual, acoustic, tactile and vestibular feedback similar to reality.

Mathematical Model: The basis for the development of a team boat rowing simulator is an equation published by Schneider 1980. Extending that equation from 2 to n team members we obtain:

$$\ddot{x}_B = \frac{1}{m_{tot}} \left(-k_{on} \dot{x}_B^2 + \sum_{i=1}^n F_{ix} - \sum_{i=1}^n m_i \ddot{\xi}_i \right) \quad i = 1 \dots n \quad (1)$$

$F_{ix} > 0$ during pull phase, $F_{ix} \equiv 0$ during recovery

Transformation Of The System Equation: Substituting $x_B = r_0 \phi$ and multiplying both sides with r_1 , equation (1) changes after rearranging to

$$\ddot{\phi} = \frac{1}{\Theta_{tot}} \left(-c_{on} \dot{\phi}^2 + \sum_{i=1}^n M_i - r_1 \sum_{i=1}^n K_i \right) \quad i = 1 \dots n \quad (2)$$

$K_i = m_i \ddot{\xi}_i$ $M_i \equiv 0$ during recovery

which is the equivalent rotating system description of equation (1).

Concept II Ergometer: The dynamic equation of a Concept II ergometer can be obtained as:

$$\ddot{\phi} = \frac{1}{\theta} (-c_{on} \dot{\phi}^2 + M) \quad M \equiv 0 \text{ during recovery} \quad (3)$$

Comparing the structure of equation (3) with that of equation (2) the necessary modification to be carried out on the mechanics of the Concept II ergometer can be deduced from the differences. To achieve moment addition two or more flywheels must be connected to a common spinning shaft. The original chain driven free-wheel must be changed to allow independent pull and recovery phases of the rowers. In addition all ergometers used in cascade must be equipped with wheels, placed on rails, centered with a spring damper system and connected to each other. After having these modifications taken into consideration equation (3) changes to

$$\ddot{\phi} = \frac{1}{\theta_{tot}^*} \left(-c_{on}^* \dot{\phi}^2 + \sum_{i=1}^n M_i^* \right) \quad M_i^* \equiv 0 \text{ during recovery}$$

$$K_i^* = \frac{m_i \xi_i}{1 + \sum_{i=1}^n m_i / \sum_{i=1}^n m_{ei}} \quad i = 1 \dots n \quad (4)$$

which is the

Equation Of The Team Rowing Ergometer Cascade.

The realization of equation (4) for $n = 2$ can be seen in figure 1.

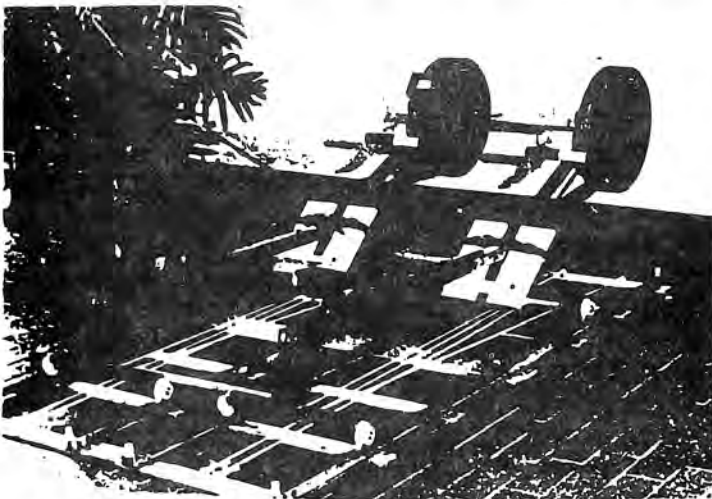


Fig. 1: Team rowing ergometer

NOTATION

m_i : mass of rower i
 ξ_i : relative acceleration
 n : number of rowers
Translatory System
 x_B : acceleration of the b.
 m_{tot} : mass of boat and team
 k_{on} : drag coefficient
 F_{ix} : blade force of rower i in longitudinal direc.
Rotatory System
 $\dot{\phi}$: angular acceleration
 θ_{tot} : moment of inertia
 c_{on} : drag coefficient
 M_i : moment of rower i
 r_i : effective lever
 m_i : mass of ergometer i
 m_{ei}^* : referring to simulator

Conclusion: Serious high performance rowers who tested the device of figure 1, say that this cascade comes closer to creating the feel of rowing than any other type of rowing machine, despite the differences in equation (4) compared to equation (2).

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A BIOMECHANICS RESEARCH ON THE PEDALING TECH. IQUE OF CYCLING

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The purpose of this article is to study the leg movements of cyclists during the pedaling in terms of sports biomechanics.

A synchronous collecting system developed by the Institute was used in the research. This computerized system was designed to collect the data from cinematograph, electromyograph and strain gauge simultaneously. The film impulse as the synchronous signal was fed by the motion-picture cameras. During the test, the triggering signal was given by the external controlling unit to the micro-computer which could eliminate the unsteady frames caused by inertia of the camera at the onset of the test, perform the high-precision calculation for the entire photographing and collect such data as the time constant of every two frames and the transient analog variables (for instance, EMG and force). The whole test was automatic and when it was complete, data were printed and stored. When the resolution of the film was accomplished, data fed into the micro-computer, and thus, the results of the variable above-mentioned during the synchronous test were obtained.

The synchronous signals were produced the moment the apertures of moving film passed through photo-coupler. During the test, the central processing unit would fetch the real-time values from the counter whenever it received the impulse signals and keep collecting the analog variables successively until the set number of collecting times was reached. The length of the collecting time in the synchronous system was 0.6ms.

Eight cyclists on the team of Tianjin City were chosen as subjects.

Twenty-four trials were performed, and the averaged data obtained and analysed in order to get the information concerning the kinesiology of leg movements during the cycling and the changes of EMG in major leg muscles and the pedaling force.

The angles of thigh, knee joint and the pedal varied with the turning of crank in a sinusoid pattern. Compared with the first two angles, the third one, to some extent, changed irregularly, which might be explained as the movement accumulation of all segments in legs. The sequence of muscle groups involving in the cycling would be elucidated according to the changes of EMG. This

sequence, together with other kinesiologic and kinetic data could make it possible to carry out biomechanical analysis on the cycling techniques effectively, and based on that, the useful suggestion of improvement in the cycling technique could be put forward. As far as the pedaling force was concerned, it was observed that the pedaling force was greater when the angle of crank ranged from 30 to 200 degree with the maximum of the force appearing within the range of 90 to 195 degree, that the pedaling efficiency changed in advance of the pedaling force by 40 degree or so, and that from 350 to 170 degree, the pedaling efficiency presented positive with its maximum being within the range of 50 to 90 degree.

A synchronous analysis was utilized to evaluate the specificity of the cycling techniques in individual cyclist. This analysis associated with the changes of the pedaling force, EMG and kinesiologic variables at the same phase of cycling may provide the accurate biomechanics diagnosis towards the cycling technique.

Therefore, we conclude that this synchronous system may be applied to the biomechanical analysis on all segments in leg during cycling. With this system, the changes of EMG, pedaling force and kinesiologic index in subjects can be recorded. Through the analysis, we found that there were two major problems in the cycling technique, i.e. weak in ankle joints and less of active flexion in knee joints.

SPECTRAL ANALYSIS OF IMPACT SHOCK DURING RUNNING

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Introduction

Repetitive impact loading of the lower extremity has been linked to joint disease and athletic overuse injuries. As a result, impact forces and the propagation of impact shock waves during walking and running have been widely studied. Typically, impact has been quantified by measuring ground reaction force peaks^{1,2,3,4} or peak lower extremity accelerations^{5,6} during the first 50 ms of the support phase. Neither force plate nor accelerometer has led to entirely satisfactory measures of impact, however. The force plate is sensitive to the net force acting on the whole body and therefore reflects the average acceleration of the whole body center of mass. With surface-mounted accelerometers, low frequency components from limb motion and high frequency components due to resonance in the attachment are superimposed on the acceleration-time signal and the pure impact component is not readily distinguishable.

Spectral analysis offers a convenient method of separating the different components of the acceleration signal and hence of distinguishing the lower extremity shock wave. The purpose of this study was to determine the effects of running speed on the spectral characteristics of head and shank accelerations during treadmill running.

Methods

Twelve experienced male runners (mass: $73.3 \text{ kg} \pm 5.1 \text{ sd}$; height: $1.797 \text{ m} \pm 0.035 \text{ sd}$) ran on a motorised treadmill at 2.0, 3.0, 4.0 and 5.0 ms^{-1} . Axial acceleration of the shank was measured with a Kulite sub-miniature piezoresistive accelerometer attached to the skin overlying the distal anterior-medial aspect of the right tibia. The accelerometer was held in position by an elastic strap tightened to the limit of subject tolerance. The strap had the effect of compressing the soft tissue between the bone and the accelerometer, resulting in a stiffer attachment with a higher natural frequency⁷. An Entran accelerometer mounted on a bite-bar held firmly between the teeth was used to record the vertical acceleration of the head. The same pair of custom-made running shoes with a midsole durometer hardness of 70 (Asker C) was worn by each subject for all trials. A footswitch mounted in the right shoe was used to distinguish the right support phase of each gait cycle.

Accelerometer and footswitch signals were sampled at 512 Hz and processed on-line using a PDP 11/44 computer with LPA11 analog to digital converter. Each trial was maintained for approximately two minutes during which the spectral characteristics of stance phase accelerations from 25 strides were determined by Fast Fourier Transformation⁸. For each trial, the average power spectral density of shank and bite-bar accelerations were calculated for frequencies in the 0 to 100 Hz range.

Results

The typical shank acceleration signal contained two major frequency domain components. The first component occurred at 5-8 Hz and was associated with the low frequency motion of the leg. A second peak, associated with the higher frequency, high amplitude shock transient was found at 12-20 Hz. These frequency-domain components coincide with the "Active" and "Impact" phases of the time-domain ground reaction force². The frequency and amplitude of both motion and shock components increased with increasing running speed (Figure 1) with changes in frequency inversely proportional to changes in the duration of the stance phase. The peak power of the leg shock transient increased disproportionately with running speed from $0.04 \text{ g}^2 \text{ Hz}^{-1}$ at 2 ms^{-1} to 0.15, 0.40 and $0.65 \text{ g}^2 \text{ Hz}^{-1}$ at 3, 4 and 5 ms^{-1} , respectively.

The power spectrum of the bite-bar signal was characterised by a single peak at 4-8 Hz. Total attenuation between the shank and the bite-bar in the frequency range 0-100 Hz averaged -0.88 dB at 2 ms^{-1} , -4.26 dB at 3 ms^{-1} , -7.01 dB at 4 ms^{-1} , and -9.29 dB at 5 ms^{-1} . The components of the

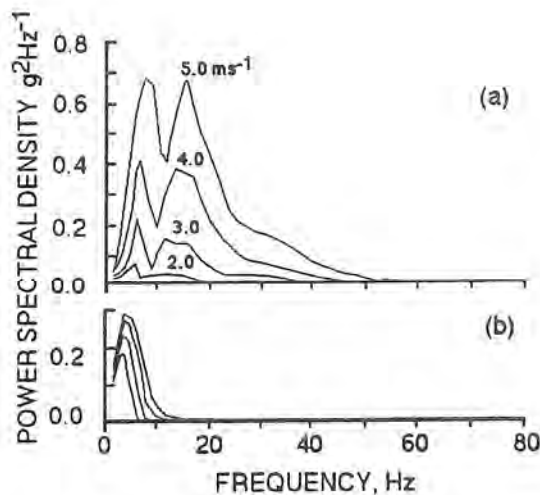


Figure 1:
Stance phase power spectra of (a) shank and (b) bite-bar accelerations at 4 different running speeds. Each spectrum is the ensemble average of 300 strides; 25 from each of 12 subjects.

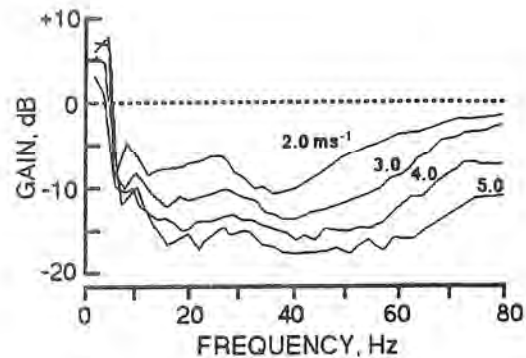


Figure 2:
Gain between shank and bite-bar accelerations.

skeletal shock transient were attenuated at frequencies above 6 Hz. Greatest attenuation occurred in the 15-50 Hz range (Figure 2). At lower frequencies (below 6 Hz) the bite-bar signal was amplified.

Discussion

Both the amplitude and the frequency of leg shock transients increased with increasing running speed. At higher speeds, contact time is reduced and the fundamental frequency of support phase kinematics increases. As a result, axial accelerations are shifted in the frequency domain.

The skeletal shock transient produced by impact may be attenuated by cartilage and other soft tissue at the joints of the leg and spine, by the cushioning effects of a shoe or surface and by kinematic changes. Since soft tissue and shoe properties are unaffected by running speed, changes in attenuation with running speed may be attributable to changes in running kinematics. Kinematic changes which may reduce shock at higher speeds include increased magnitude and rate of knee flexion during contact^{9,10} and a shift from a heel to a midfoot strike pattern¹.

While further research is required to clarify the effects of cushioning systems on the spectral characteristics of skeletal shock transients, the results of this study suggest that the design of a shock absorbing shoe should take into account the higher frequency and disproportionately greater power of skeletal shock transients at higher running speeds.

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GRIPPING FORCES AND HAND ACCELERATION WAVES DURING THE GOLF SWING

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INTRODUCTION

It is necessary to comprehend the mechanism in order to achieve effective movement of the golf swing. Golf swings have been depend on effective club movement from the fingertip of both hands. Various gripping style of the golf have been used by many golf players, who have trained those patterns of the golf swing. Gripping forces could be appeared with subjective observation alone during the golf swing. Kawashima(1988) and Budney (1979) measured the pressure of only center of the grip during the golf swing. Then, this study has the purpose to investigate the grip forces of seven fingers of both hands and acceleration waves with two directions during the golf swing.

METHODS

Fourteen males were subjected with using the driver tool of golf. Gripping forces and acceleration waves of the left hand were measured with the small sensors quantitatively. The force of each finger was recorded on the osillograph with this mini-sensor from the strain amplifier, and the acceleration wave was recorded on the erectro-magnet-graph with the piezo-electric accelerator from charge amplifier. Also, segmental motion analysis were practised in those golf swing by the measure of the 16 mm high speed camera, cinematographically (see Figure 1).

RESULTS AND DISCUSSION

Figure 2 showed typical data of finger force and acceleration wave during the golf swing. Table 1 showed averaged peak values force of at seven fingers of both hands (overhand grip-stye) during golf swings. Averaged values according to classified trained and untrained groups, and then statistical difference of two values caliculated with significant values. At just impact the peak force of the left little finger were recorded 14.3 ± 7.3 N for the elite, and 3.8 ± 2.1 N for the beginners. Comparison between those values was a significant defference statistically ($p < 0.05$).

Forces of the left ring finger, the elites value was 15.2 ± 7.7 N and those of beginners were 9.25 ± 5.0 N at impact point($p < 0.05$). While, the force of the left middle finger of the elite were increased higher than that of the beginners.

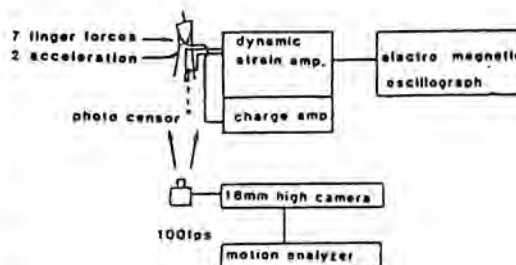


Figure 1 Diagrammatic illustration of the experiment system.

Table 1
The maximum values of force fingers and the acceleration during the golf swing.

Finger force & Acc.	Elite(n=7)		Beginner (n=7)		t
	m	sd	m	sd	
L.Little (N)	14.3	7.3	3.8	2.1	*
L.Ring (N)	15.2	7.7	9.2	5.0	*
L.Middle (N)	17.4	4.1	9.5	4.7	*
L.Thumb (N)	62.2	21.2	52.7	26.7	ns
R.Middle (N)	17.1	7.6	15.3	8.4	ns
R.Index (N)	10.8	4.9	8.9	5.3	ns
R.Thumb (N)	19.3	6.8	17.7	8.1	ns
Dorsum manus (G)	4.5	1.0	2.4	0.8	*
Processus styloideus (G)	5.3	0.8	3.9	1.0	*

* = Significant differences between the groups, $p < 0.05$.

Both forces of the middle and the index fingers of the right hand did not get difference between trained and untrained subjects. However, the maximum forces of two fingers of the right hand were appeared at the time about 120-230 ms prior to the impact. The left hand movement of the elite were accelerated faster at prior of the impact than that of the beginners ($p < 0.05$). Then, formation of the golf swing-pattern would be suggested to be important to make without drawing out three force of the small, the ring, and the middle fingers.

CONCLUSIONS

The results of this study would indicate the following conclusions:

- 1) Peak forces of the little, the ring, and the middle fingers of left hand were recorded remarkably larger swing of elite players than during that of beginners at the golf experiment.
- 2) The swing of the elite were accelerated faster at prior of the impact than during that of beginners at the decline of the acceleration curve.

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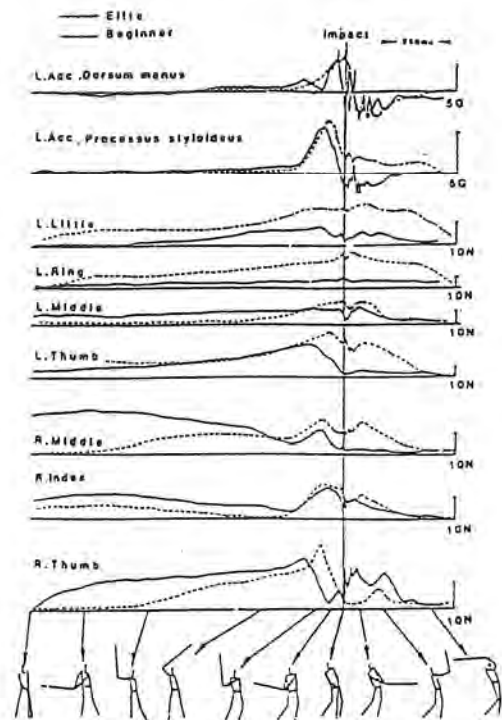


Figure 2 The typical waves of the finger forces and the acceleration during the golf swing

THE EFFECT OF ADDED LOAD ON LANDING STRATEGIES

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Mass added to a body segment represents a mechanical constraint imposed on an individual when engaging in activities requiring equipment such as shoes, boots and skis. Furthermore, individuals readily engage in activities requiring landing from considerable heights when using such equipment. The purpose of the study was to assess strategies used by subjects to accommodate to the addition and removal of masses during landings.

METHODS

Four semi-skilled male university students performed three conditions (C) of 25 landings from a 60 cm height on each of two test days. Additional masses (1024 and 1800 g) were attached to each leg for C2 and removed for C3 on each test day. Ground reaction force (500 Hz) data for fore- (F1) and rearfoot (F2) impact and EMG (1000 Hz) data from five lower extremity muscles were used in the analysis. Strategies used to accommodate to the addition and removal of the masses were assessed through use of group and single subject ANOVAs and simple and multiple regression techniques. Single subject regression analyses were performed to evaluate strategies used to accommodate the loading and unloading manipulations. For the loading analyses, the first 50 trials (C1 and C2) of a single test session were used and the final 50 trials (C2 and C3) were included in the unloading analyses. Separate analyses were performed for each load (1024 g and 1800 g) and force parameter (F1 and F2). Two regression models were used to assess a Newtonian response component and a combined Newtonian and neuromuscular response component.

RESULTS AND DISCUSSION

Repeated measures ANOVAs were conducted for each force variable using the 25 trial subject-condition mean values. A significant interaction for Day x Subject on F1 indicated that all subjects did not respond equally for this parameter on the two test days, which were also different load conditions. No significant main effects were observed. The general lack of significant differences supports the conclusion that the subjects responded similarly to the three conditions using neuromuscular strategies that resulted in attenuation of the maximum force increases that should have resulted from the added load.

Single subject ANOVA results, however, indicated that differences were present for individual subjects (Table 1). Sixteen of the 32 comparisons for the individual subject within day data were significant ($p < 0.05$) indicating the presence of experimental effects. Subject 1 exhibited a significant Newtonian response component for seven of the eight load-variable conditions. Subjects 2 and 3 accounted for the addition of both loads and Subject 4 for the 1024 g load through neuromuscular response mechanisms. Load removal resulted in three, two and one significant Newtonian responses for subjects 2, 3 and 4, respectively.

The simple regression analyses produced the same results as the within-subject ANOVA analyses. However, in addition they also allowed for quantification of the common variances between load and force making between subject and model comparisons more meaningful. The three significant F1 and two significant F2 loading responses accounted for an average of 25 and 15% variance, respectively. The significant unloading responses (10) explained 25 and 22% of the variance.

The multiple regression analyses using the IEMG values in addition to load improved all predictions considerably with an average increase of 34% in explained variance. The load variable entered only two (13%) of the loading models but contributed significantly to nine (56%) of the unloading models. The average number of IEMG variables included in the models was 2.5. Total explained variance for F1 was slightly greater than for F2 (31 vs 27%), the heavier load produced better results than the lighter load (50 vs 41%), and the unloading models accounted for greater variance than the loading models (52 vs 39%).

Individual subject results were extremely variable as should be expected since different strategies were employed as previously discussed. Average ratios between the explained variances were 1.33 (0.49 to 2.62) for F1 compared to F2, 1.61 (0.92 to 2.40) for loading compared to unloading, and 7.04 (2.34 to 15.25) for the Newtonian compared to the neuromuscular model. The average explained variances for the four subjects were 33.4, 22.8, 24.7 and 31.2%, respectively.

The repeated measures group ANOVA results using force parameter condition means demonstrated the ineffectiveness of a between subjects design for accurately assessing response strategies. The results indicated that all subjects performed equally under all conditions with respect to all parameters, thus totally accounting for the load. Single subject ANOVA results, however, demonstrated differential response strategies for load, parameter and loading/unloading conditions. Simple and multiple regression analyses employing the independent variables of load only, and load plus IEMG, respectively, for predicting F1 and F2 generally supported the individual ANOVA results. The observed significant increases in explained variance with the inclusion of IEMG data in the regression analyses demonstrated that all response strategies included a neuromuscular component while 44% of the responses also included a Newtonian component.

Table 1. Twenty-Five Trial Mean Values For All Conditions

	S1		S2		S3		S4	
	F1	F2	F1	F2	F1	F2	F1	F2
<u>1024 g</u>								
C1	11.71	23.29	11.73	27.90	18.72	38.90	13.31	47.96
C2	13.51*	25.37*	12.10	25.66	18.48	40.40	15.55*	43.69*
C3	12.31*	26.86	10.18*	21.83*	19.37	35.26*	14.09*	46.21
<u>1800 g</u>								
C1	12.72	26.81	12.91	28.35	17.80	36.27	17.11	40.81
C2	14.89*	29.60*	12.81	26.30	16.94	35.45	17.05	39.47
C3	11.31*	26.39*	11.40*	24.28	15.97*	42.66*	16.41	38.85

* indicates significant differences between conditions: C2* between C1 and C2; C3* between C2 and C3.

Note: All values expressed in N/kg body mass; C1 and C3 are the no-load conditions preceding and following the added load condition (C2); F1 and F2 are the forefoot and rearfoot maximum impact forces.

Effects of Asymmetrical Load Carrying on Ground Reaction Forces During Walking

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Researchers (1,2) have shown that symmetrical load carrying activities can result in injuries to the musculoskeletal system. However, few investigations have been reported on the effects of asymmetrical carrying tasks. The purpose of this study was to investigate the effects of an asymmetrical, sidepack carrying system on ground reaction forces (GRF) during walking. The sidepack was carried on the left side with a vertical strap over the left shoulder. Three load conditions, 0, 10 & 20% of body weight, along with the effects of load on left and right foot contacts were analysed. Five healthy males (mean age: 25 years, weight: 67 kg) volunteered as subjects. The experimental setup consisted of a force platform interfaced to a laboratory computer and a timing system to monitor walking speed (1.34 ± 0.07 m/s). For each subject, 10 trials were collected for both left and right stance phases for each of the three load conditions. Four parameters describing the vertical (V) and mediolateral (ML) GRF curves were analysed with a single subject statistical procedure (3) to identify significant load and support leg effects ($p < .05$).

Table 1 lists the mean and SD values for all subjects and Figures 1 and 2 present a representative subject's mean V and ML GRF curves. A small left-right asymmetry was evident in the mean values during normal walking (0% load), however with additional loads the left-right differences were increased by an average of 6 times in 7 of 8 load/parameter measures. The V GRF peak magnitudes increased with additional loads but the response was different for each foot. With increased loads, left foot contact had greater first and second maximum force peaks and a lower minimum peak compared to the right foot. The asymmetrical load also produced left-right differences during the single support phase in the ML direction. During this period, larger medially directed GRFs were exerted on the right foot as load increased. Carrying the sidepack moved the body/load mass center closer to the left leg therefore requiring less frontal plane accelerations during left leg support compared to those during right leg support.

The results of the statistical analysis are presented in Table 2. Four comparisons were performed (2 load & 2 contact foot) for each parameter. All subjects had significant load effects in the V force measures however the different response trend for left and right foot contacts in the ML parameter reduced the overall differences between loads, producing a low rate of significance. The left-right comparisons identified 16 and 18 significant differences out of a possible 20 for the 10 and 20% loads, respectively.

The observed left-right asymmetry in GRFs indicated that the sidepack carrying system had a greater injury potential than symmetrical load systems due to the imbalance of forces exerted on the lower extremities. Further investigations are indicated

to analyse the extent to which these differences are transmitted up the legs into the pelvis and spine.

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Table 1. Mean & Standard Deviation Data for GRF Parameters

Parameter	R 0%	L 0%	R 10%	L 10%	R 20%	L 20%
First V Peak	11.62 (0.23)	11.64 (0.32)	12.65 (0.44)	12.91 (0.52)	13.52 (0.56)	13.84 (0.65)
Minimum Peak	7.77 (0.26)	7.64 (0.33)	8.51 (0.48)	8.25 (0.45)	9.17 (0.58)	8.91 (0.60)
Second V Peak	11.89 (0.49)	12.09 (0.57)	12.91 (0.62)	13.36 (0.71)	13.84 (0.79)	14.39 (0.77)
Avg ML Force	0.21 (0.07)	0.11 (0.04)	0.14 (0.21)	0.07 (0.05)	0.25 (0.10)	0.08 (0.06)

All values in N/kg

Table 2. Summary of Statistical Analysis

Parameter	0 vs 10%	0+10 vs 20%	R vs L 10%	R vs L 20
First V Peak	5	5	4	5
Minimum Peak	5	5	3	4
Second V Peak	5	5	4	5
Avg ML Force	1	0	5	4

All values are number of subjects with significant results

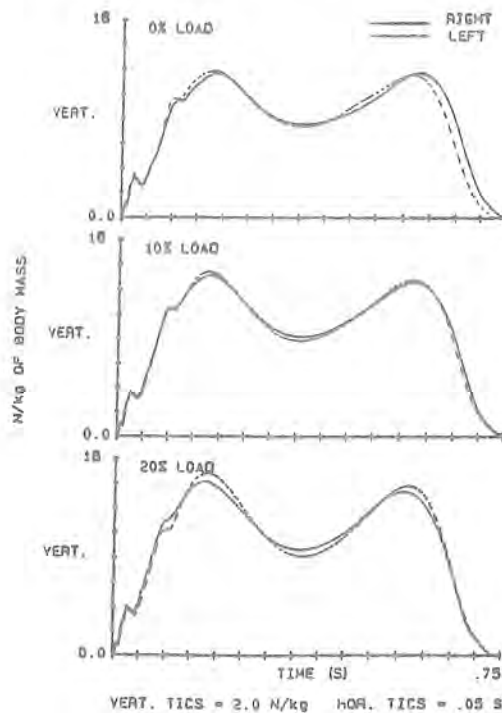


Figure 1. Average Vertical GRF Curves For Subject #5

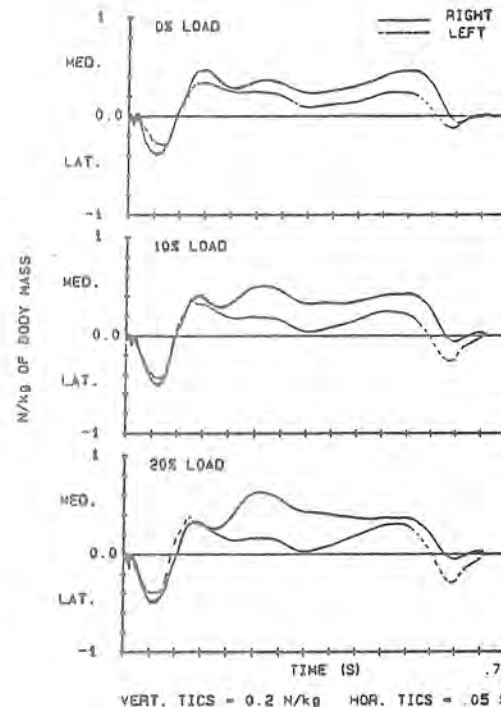


Figure 2. Average Mediolateral GRF Curves For Subject #5

Intrasubject Variability in Lower Extremity Joint Moments of Force During the Support Phase of Running

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Lower extremity joint moments of force (JMF) have been reported for running along with estimates of intersubject variability of these measures (1). However, few estimates of intrasubject variability (ISV) in JMF patterns have been reported. Also, ISV has been identified in ground reaction forces (2) suggesting the need to investigate lower extremity joint kinetics. The purpose of this study was to quantify ISV in lower extremity JMFs during the support phase of running. Six males (mean age: 24 years, weight: 70.3 kg) volunteered as subjects. The experimental setup consisted of a force platform interfaced to a computer and a timing system to monitor running speed ($4.3 \pm .2$ m/s). A 16 mm cine camera was used to film the subjects in the sagittal plane. The testing protocol consisted of 4 identical test sessions per subject: days 1 (morning & afternoon), 2 & 8 to provide for within test session (c1) & day (c2), and between day (c3) & week (c4) estimates of ISV. During all sessions, each subject performed 25 trials from which 10 were randomly selected for analysis. An inverse dynamics approach which combined the film and force platform data along with subject anthropometric measures was used to calculate the hip, knee & ankle JMFs during the activity. Thirty mean parameter values describing the JMF curves from each session were compared with a single subject statistical procedure ($p < .05$) (3) to identify differences between sessions (c2-c4). Also, coefficients of variation (CV) were computed for the total JMF curves to quantify the average normalized standard deviation across periods c1-c4.

Figure 1 shows a subject's mean JMF curves for c1 and c3 and Table 1 presents the results of the CV and statistical analyses (percent of significant differences). The CV values indicated that the average SD among JMF curves increased a small amount over that of a single test session but was consistent over longer periods c2-c4. The observed CV values represent approximately 50% of the corresponding intersubject CVs previously reported (1). However, the intersubject CVs continually decreased from proximal to distal joints whereas the values presently observed were similar for knee and ankle JMFs. The statistical results showed an increasing number of significant comparisons as the time between tests increased. The absolute differences between means (MAD) for the significant comparisons from 3 parameters are shown in Table 2. Generally, these results indicated that the mean parameter values had greater differences between days and weeks compared to within day. However, the trend was not linearly related to the length of time between tests since most between day values were larger than the between week differences. The observed hip and ankle MADs represent 52 & 31% of previously reported (4) intrasubject MADs between 2 running speeds (4.09 & 4.60 m.s) for the same parameters.

The results of both analyses can be combined to produce an overall evaluation of ISV. The magnitude of the observed CV values compared to intersubject CV values and of the MADs compared to those between different running speeds indicated that ISV is functionally meaningful

and should be considered in biomechanical analyses. The CVs provided an average estimate of ISV for each JMF over the entire support phase and indicated that the hip JMF varied more than knee and ankle JMFs. In addition, the average ISV for these measures was best estimated by combining samples from 2 consecutive days since the CV and most MAD results had peak values in the c3 comparison. However, within the ISV envelop expressed by CV and MAD values a greater number of specific performance characteristics approached these limits as the time between measurements increased.

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Table 1. Coefficients of Variation & Statistical Analysis Summary

Joint	c1	c2	c3	c4	c2	c3	c4
Hip	37	40	40	40	21	23	24
Knee	20	22	22	23	24	32	40
Ankle	19	22	22	22	14	40	45

All values in percent

Table 2. Mean Absolute Differences For Significant Comparisons

Parameter	c2	c3	c4
Max Hip Extensor Mom.	0.416	0.454	0.433
Max Knee Extensor Mom.	0.302	0.331	0.390
Max Ankle Plantar. Mom	0.136	0.261	0.157

All values in Nm/kg

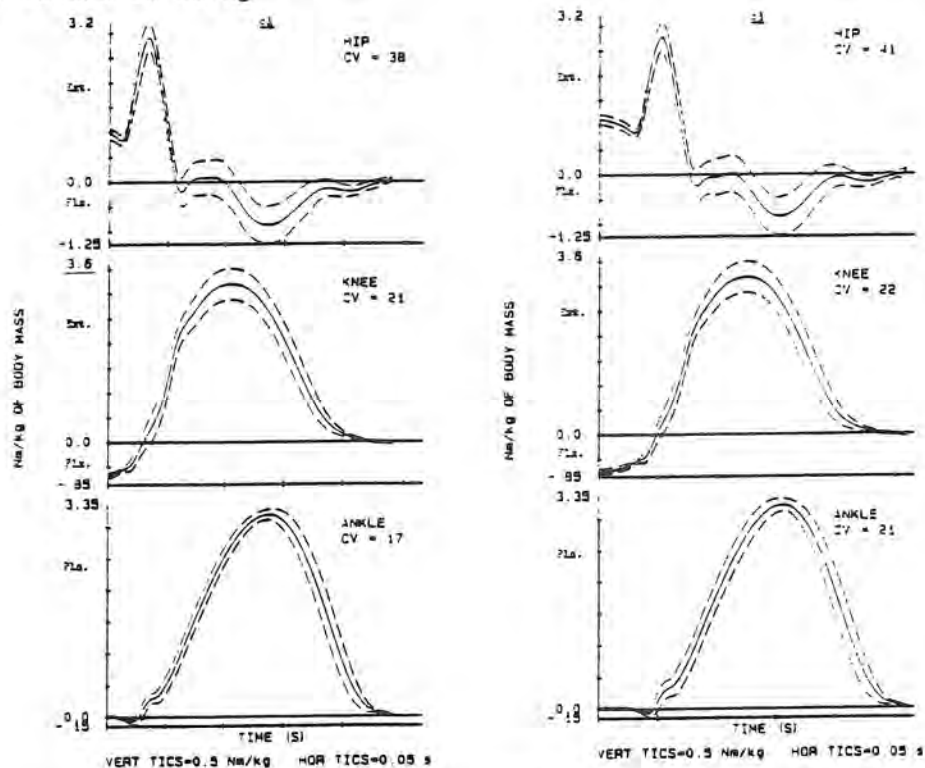


Figure 1. Mean & standard deviation curves for subject 4

DOUBLE EXPOSURE HOLOGRAPHIC INTERFEROMETRIC EVALUATION OF PLATED FRACTURE SYSTEMS

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Holography is a photographic technique which records the intensity and phase of monochromatic light which is reflected from the object and compared to a reference beam. Holographic interferometry is a non-destructive, contactless method for studying overall displacements of irregular or complex systems. Alternative methods for assessing systems in biomechanics have used a brittle lacquer coating which does not adhere very well to wet bone or strain gauges which give point by point information which has to be interpolated later. Double exposure holograms produce a three-dimensional image overlayed with dark stripes; the dark stripes represent lines of equal displacement - every point along a fringe has displaced the same amount from its original position as a result of the load which has been applied.

To produce a hologram a beam from a He-Ne laser is split into two parts known as a reference beam and an object beam, Figure 1. The reference and object beams are exposed to the photographic plate simultaneously and the interference leads to a diffraction pattern which makes up the hologram. The photographic emulsion is then developed and fixed as occurs with normal photography. To record a double exposure hologram the object is recorded in its undeformed state, it is then deformed by loading and a second exposure is taken. The hologram, once processed, then reconstructs both images which interfere and produce interference fringes. It can then be viewed by replacing the processed photographic plate in the same position as had been used to record it, and illuminating it using the reference beam only. Previous work [1,2] using double exposure holographic interferometry has shown the technique to be useful for assessing a variety of complex orthopaedic systems. In this study the role of different screws in a plated system has been investigated.

Experimental

Horse metacarpal bone was received fresh and stored at -20°C . The bones were thoroughly cleaned, then turned on a lathe, inside and outside, into regular cylinders. The ends of the cylinders were then filled with an epoxy reinforced with metal, and two 10mm diameter holes drilled either end, along the cylinder axis. Holes of 5mm diameter were then drilled transverse to the longitudinal direction. The cylinder was loaded in torsion through these holes and a double exposure hologram taken at various loads. A six holed stainless steel 4mm thick plate with Sherman self tapping screws was applied to the cylinder with an osteotomy through the middle of the shaft. Initially the screws were all tightened to 3Nm using a torque limiting screw driver; the bone was loaded in torque, in a series of load increments and double exposure holograms were taken. Screws were sequentially loosened then removed, as shown in the results section, double exposure holograms being taken at each stage. The load increments applied to the bone were varied according to the stiffness of the whole system. A fringe density that was visible while not being too coarse was required.

Results and Discussion

The photographs shown in Figure 2 show examples of the holograms that were recorded. A shows the effect of loading a complete cylinder, the fringes are continuous and diverge towards the right hand side where the cylinder was built in. B shows the cylinder, with a saw cut made through the cylinder, a six holed plate attached and the screws tightened to 3Nm. The fringe density either side of the fracture can be seen to be different indicating relative movement at the fracture site and showing how the plate itself accommodates the angular deflection. Loosening the two outer screws (C) or the two middle screws (D) does not make much difference in the fringe pattern, although for D the fringes do deflect more on the plate around the loosened screw. Diagrams E and F show the effect of loosening and of removing respectively, one screw on the

right hand side of the fracture. The fringe pattern has been greatly disturbed in both these cases, more so when the screw has been completely removed. The plate transmits little of the deformation or load in torsion to the right hand side.

The holograms show the effect that changing one variable can have on the strains and therefore stresses of a system. This series of holograms also demonstrates the micromovement that occurs at a fracture site if the bone ends are not put into compression. The most important screws are the ones which are closest to the fracture site, and that although it is probably necessary to have a plate attached at 12 points of the cortex, the middle two screws of each half, do not in fact have a large effect on the structural torsional stiffness.

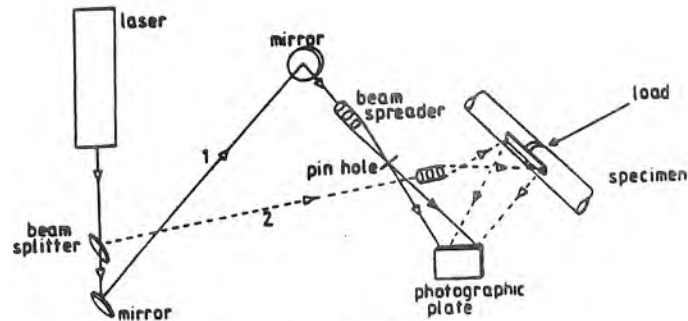


Figure 1

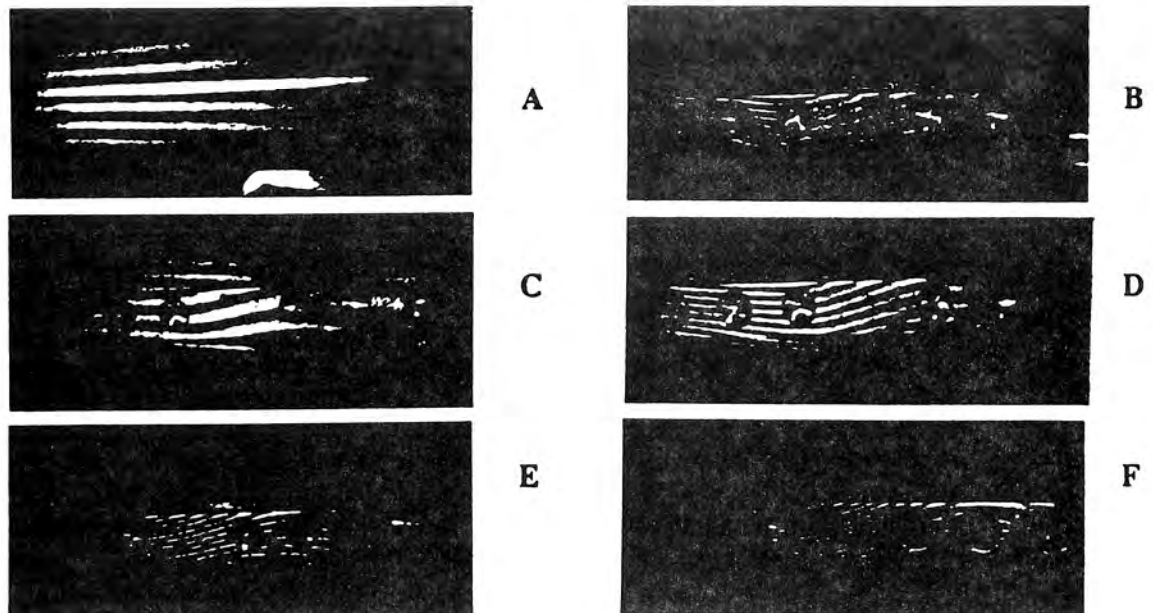


Figure 2

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Acknowledgements

The support of SERC and Johnson & Johnson Orthopaedics Ltd in the provision of a CASE Research Studentship for one of the authors (J.S.) is gratefully acknowledged.

THE FRACTURE PROPERTIES OF CORTICAL BONE IN THOROUGHBRED RACEHORSES:
RACECOURSE DESTRUCTION CASES COMPARED WITH NORMAL BONES

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Introduction

The seriousness of racecourse orthopaedic injuries frequently leads to the destruction of the animal. Around 150 horses a year are destroyed on British racecourses, of which 70% are a consequence of fractures in the lower foreleg. An understanding of why these fractures occur may lead eventually to strategies for their prevention. Linear elastic fracture mechanics attempts to quantify the resistance of brittle materials, such as bone, to crack growth and is therefore a measure of the material's fracture toughness. In this study specimens cut from both fractured and non-fractured (intact) bones have been evaluated at in vivo strain rates in order to assess the fracture mechanisms in equine cortical bone.

Methods

Compact tension specimens conforming to ASTM standard E399-83 were machined from the Cannon bones of a number of animals. Fractured bones were obtained from racecourse destruction cases while intact bones came either from the same source or from Royal Veterinary College cases. Prior to testing the density of each specimen was calculated using Archimedes principle. Specimens were maintained wet throughout all machining and testing procedures. The specimens were loaded in a Schenk Trebel universal testing machine fitted with a 2KN load cell at strain rates varying from 1.67×10^{-4} to 0.5 s^{-1} . Initially a crack was propagated from a notch of length 3.9 mm in a slow controlled manner using one of four slow strain rates ($<0.01 \text{ s}^{-1}$), until it progressed to a length of 9.3 mm ($a/W = 0.6$), marked by a scribe line and observed via a cathetometer. The load at this point was recorded on a chart recorder and the crosshead of the testing machine reversed until the load reached zero. The strain rate was then increased to one of four fast strain rates ($>0.01 \text{ s}^{-1}$) with an introduced crack row of length 9.3 mm. The peak load in these fast tests were recorded via a digital output. Thus two values of K_{IC} could be calculated for each specimen using the ASTM standard. However, pre-tests showed that sub-critical crack growth occurred during the unloading phase which resulted in lower K_{IC} values being measured from the reloaded tests. A correction factor of 20% was calculated to allow for this.

Results & Discussion

Figure 1 shows that initial increases in strain rate result in increased toughness. A peak is apparent, however, for both fractured and intact bones beyond which further increases in strain rate result in decreased toughness. For the intact bones this peak occurs around 0.1 s^{-1} which is at the top end of the in vivo strain rate range ($0.01 - 0.1 \text{ s}^{-1}$). For the fractured bones the peak occurs earlier at 0.05 s^{-1} . Fractographic analysis revealed that this drop off coincides with a transition from ductile failure producing rough fracture surfaces, to brittle failure producing smooth fracture surfaces. The drop off effect is illustrated in a single fractured bone in Figure 2 where again the peak appears near 0.05 s^{-1} . Since the load on the galloping horse increases the faster he runs this drop off in K_{IC} at high strain rates means that the critical flaw size is decreased, thus putting the horse at greater risk of fracture.

Figure 1 Fracture toughness versus strain rate for fractured and intact bones.

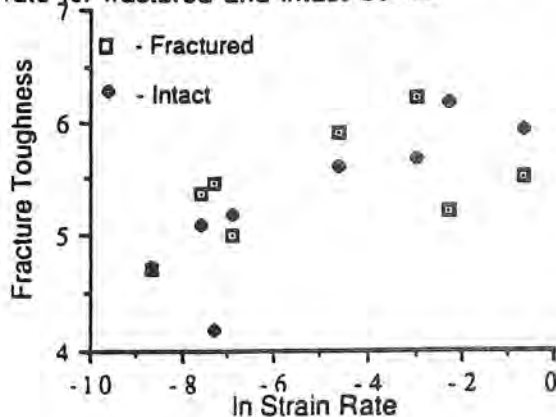


Figure 2 Fracture toughness versus strain rate for a 4 year old gelding.

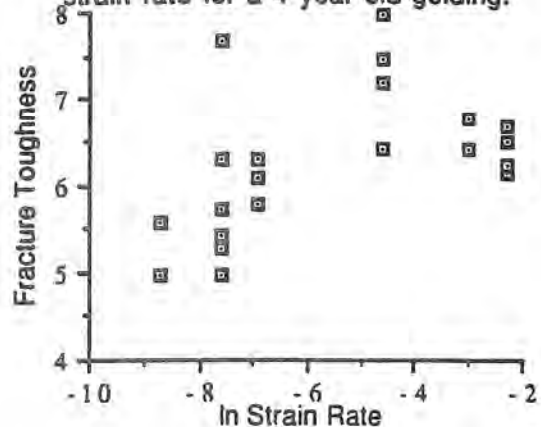


Figure 3 shows a negative relationship between toughness and density. Figure 4 shows increasing density with age along the line shown. The circled data points were the results obtained from fractured bones and have a higher density than would be predicted for their age. These results suggest a congenital condition which may pre-dispose the bone to catastrophic crack propagation at lower than normal strain rates.

Figure 3 The relationship between fracture toughness and density.

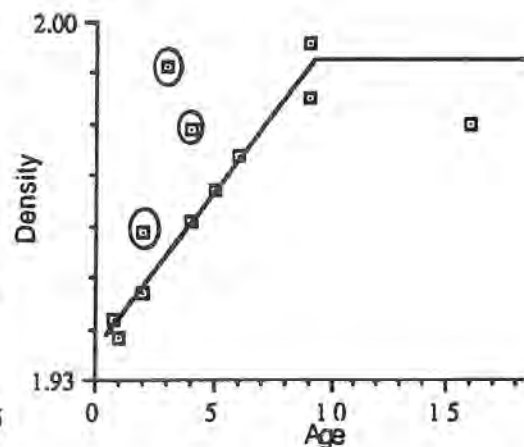
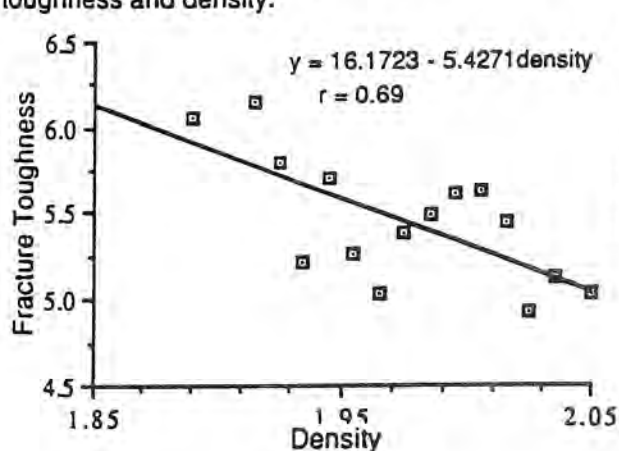


Figure 4 Density variation as a function of age.

Acknowledgements

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EFFECTS OF DYNAMIC FEATURES OF TENNIS RACKETS ON MUSCULAR LOAD OF TENNIS PLAYERS

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Among the injuries caused by sports activities, epicondylitis, so-called tennis elbow is one of the most common and well-known diseases. Tennis elbow might be considered a degenerative disease, the onset of which is hastened by overuse of the arm and elbow (Gruchow and Pelletier, 1979). Yoshizawa et al. (1987) reported that the total activity levels of the forearm muscles were somewhat lower in highly skilled tennis players than in moderately skilled players. Thus the muscular load on the forearm may be affected not only by the level of skill of the players and also by the characteristics of the rackets. In the present experiments, electromyographical and kinesiological studies were carried out to elucidate effects of dynamic features of various models of tennis rackets on muscular load on the forearm of tennis players.

Materials and Methods

Tennis rackets used were eight types of four models, which differ in weight, face area, hardness of shaft, and gut tension, of different manufacturers. Subjects employed were 11 University Club Members ranging in age from 19 to 22 who had participated in the official contests and gained good results. Electromyograms (EMGs) were recorded by means of bipolar surface electrodes, 5mm in diameter, attached over the extensor carpi radialis brevis, the extensor carpi radialis longus, the flexor carpi radialis and the flexor carpi ulnaris muscles with an interelectrode distance of approximately 2cm. EMGs, strain curve of the racket and video motion pictures were simultaneously recorded during forehand and backhand strokes with controlled ball, every four seconds by tennis ball machine, and service. The EMGs recorded were analyzed quantitatively by the amplitude probability distribution analytical method described by Jonsson (1982).

Results and Discussion

Results obtained in the present experiments and discussions were summarized as follows: 1. In the case of the heaviest test racket in comparison with the light one, the stronger discharge of the extensor muscles were observed from the end of the back swing phase to the onset of the forward swing. And, strong discharges of the flexor muscles were observed before the ball impact. There was a tendency that the larger the moment of inertia of the test racket, the higher the amplitude level of the EMGs.

2. The smaller face area of the test racket induced the more prolonged vibration of the shaft and the higher amplitude level of the EMGs of the extensors after the ball impact of the backhand

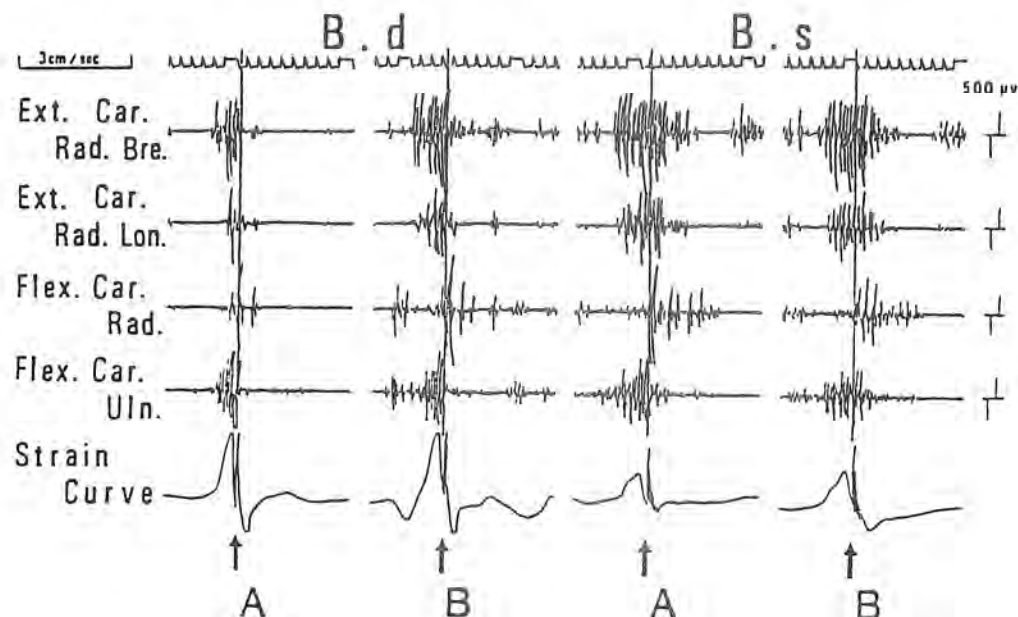


Figure 1. EMG patterns and strain curves of the shaft during the backhand stroke with the double (B.d) and single (B.s) hands. A:largest face area of the test racket, B:smallest face area. Arrows show the ball impact points.

stroke with single hand. The double hand stroke could markedly attenuate the racket vibration and the amplitude level of the EMGs after the ball impact. (Fig. 1)

3. The hardest shaft of the test racket attenuated the activity of the lowest amplitude level of the EMGs.

4. The higher gut tension of the test racket increased the activity of the middle amplitude level of the EMGs than in the given gut tension which was recommended by each manufacturer.

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BIARTICULAR MUSCLE ACTIVITIES DURING FRONT HANDSPRING IN TUMBLING

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The present experiments were carried out to elucidate functional characteristics of the antagonistic biarticular muscles of upper extremity which participated in supporting the dynamic load during hand contact period of front handspring in tumbling. According to Yamashita et al. (J. J. Sports Sci., 2:318-324, 1983), the discharge pattern of the biceps brachii long head (Blo) and the triceps brachii long head (Tlo) could show which joint force, the elbow or the shoulder, more prominently contributed to developing the resultant force output of arm extension rather than the other.

Subjects employed were 4 skilled male adults and 3 unskilled. The EMGs were recorded from the upper extremity, shoulder girdle and trunk muscles utilizing conventional surface electrodes. Horizontal and vertical components of floor reaction force, side-view motion picture and angular change of the elbow joint were simultaneously recorded with the EMGs.

In the skilled subject (Figure 1 SKILLED), the Blo showed the strong discharge before the hand contact; however, this discharge ceased and the Tlo increased after the hand contact. In the monoarticular elbow and shoulder joint muscles, the triceps brachii lateral head (Tla) showed the marked continuous discharge during hand contact period, and the marked discharge of the anterior portion of deltoid (Da) which was observed before the hand contact showed the tendency to decrease after the hand contact. On the other hand, in the unskilled subject (Figure 1 UNSKILLED), these discharge patterns of biarticular muscles reversed, that is, the discharge of the Blo abruptly increased after the hand contact and the marked discharge of the Tlo which was observed before the hand contact showed the tendency to decrease after the hand contact. As to the monoarticular elbow and shoulder joint muscles, the Tla showed the marked continuous discharge during hand contact period as was seen in the skilled subject, but, the discharge of the Da showed the tendency to increase after the hand contact. In the hand contact period, the elbow and shoulder joints are required to support the load by the elbow extension and shoulder flexion forces. Therefore, the Blo acts as the shoulder flexor and the Tlo, the elbow extensor in this period during the handspring. Referred to the paper dealing with the EMG and dynamic significances of the upper arm extension (Yamashita et al., 1983), the discharge patterns of the biarticular and monoarticular muscles obtained in the present experiments strongly suggested that the elbow extension force more prominently contributed to developing the floor reaction force than the shoulder flexion force in the skilled subjects, and vice versa in the unskilled subjects.

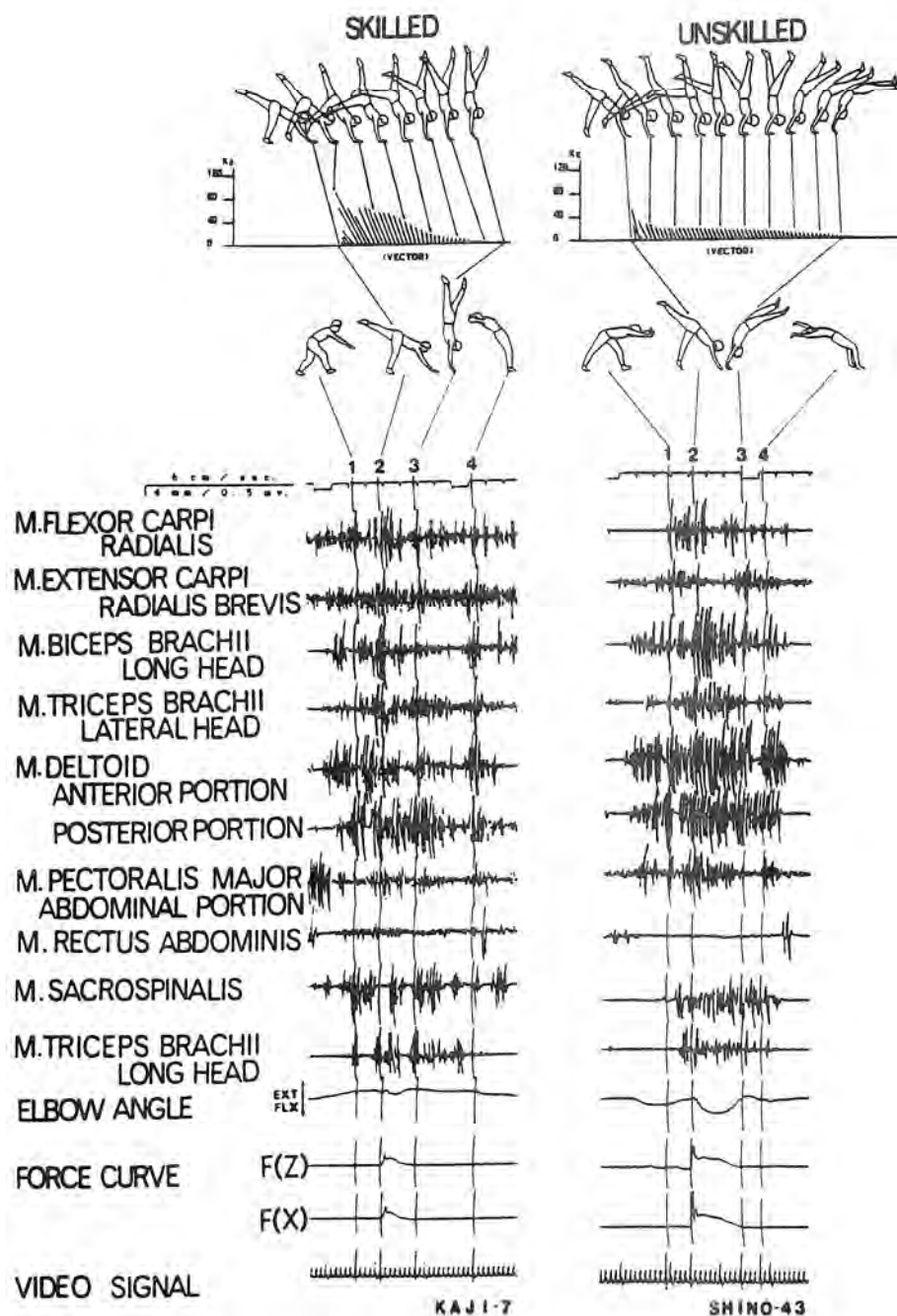


Figure 1. EMGs of front handspring of the skilled and unskilled. Postures 1, 2, 3 and 4 show the point where the forward foot contacts the floor, the hand contacts the floor, the hand leaves the floor, and the backward foot contacts the floor, respectively.

COCONTRACTION OF THE ELBOW MUSCLES WHILE PERFORMING COMBINED TASKS OF PRONATION-FLEXION AND SUPINATION-FLEXION

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The execution of simple contractions often involves reciprocal activity of agonist and antagonist muscles which is referred to as cocontraction. This cocontraction is believed to allow for a better pressure distribution over joint surfaces⁴. It has been demonstrated that activation of antagonists increases with agonists activity⁶. Increased EMG activity in extensor muscles of the elbow has been reported while performing isometric flexion with the forearm in pronation as compared to the forearm in supination or in neutral position³. These results suggest that cocontraction of antagonist muscles could be modulated by force levels⁶ and forearm positions³. Previous studies^{1,2} have shown that either facilitation or inhibition of agonist muscles can be obtained from combined isometric contractions around the elbow. Furthermore, it has been shown that recruitment of different motor units within a given muscle is task dependant⁵. The purpose of the present study was to determine if a) antagonist activity is modulated by the combined contractions of pronation-flexion and supination-flexion as compared to pure elbow flexion, and b) cocontraction activity of extensors, relative to each of these contractions, is similar across elbow angles.

METHODOLOGY

With the forearm in mid-pronation, mid-supination, the subjects (n=6), seated and stabilized, were required to produce isometric contractions such as: a) a ramp flexion contraction (8.0 sec) of the elbow ranging from 0 to 100% maximum voluntary contraction (MVC); b) a similar flexion contraction preceded and combined with 20% MVC pronation of the forearm (pronation-flexion); c) a similar flexion contraction preceded and combined with 20% MVC supination of the forearm (supination-flexion). These tasks, executed at random, were repeated for elbow angles of 50°, 90° and 130° of flexion (0° = anatomical position). The EMG activity of biceps brachii (BB), brachioradialis (BR), triceps brachii (TB) and anconeus (AN) were amplified (TECA AA6, CMRR >100dB, input impedance >100Megohms) and sampled at 1000Hz. Two dynamometers measured the moments produced in flexion of the elbow and in pronation or supination of the forearm. The root mean square (rms) values on 0.05 sec intervals taken at 20%, 40%, 60%, 80%, and 100% MVC (torque signal) in flexion were obtained. The ratios of agonist versus antagonist rms values (BB/TB, BR/AN) were then used as indexes of cocontraction.

RESULTS AND DISCUSSION

The descriptive analysis of the BB/TB ratios across different force levels (%MVC) shows a progressive increase of the ratios at levels of 20% MVC and 40% MVC for all elbow angles. This BB/TB ratio increases with supination-flexion and decreases with pronation-flexion. At higher force levels, the ratio remains stable and is not affected by the combined contractions (fig.1a). Friedman ANOVAs at 90° of elbow flexion confirm that a significant increase in BB EMG activity occurs for 20%, 40% and 60% MVC ($p < .01$) whereas the TB EMG activity remains constant. For a 50° elbow angle,

the increase of BB EMG activity is still responsible for the increase of BB/TB ratio at a 20% MVC level ($p < .05$). For a 130° elbow angle, the observed tendency of BB/TB ratio could not be statistically explained by the individual BB or TB EMG activities. The BR/AN ratio curves across different force levels indicate a progressive increase of this ratio with force in all contractions (fig.1b). The differences across contractions are not statistically significant ($p > .01$).

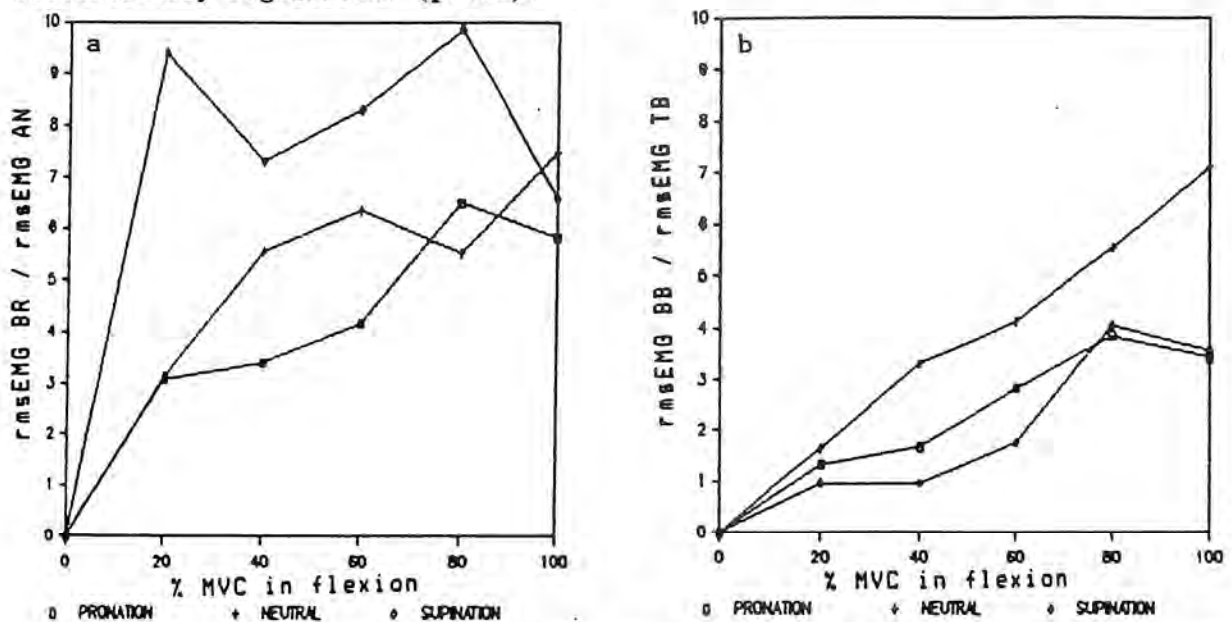


Figure 1-. Cocontraction indexes across force levels at 90° of elbow flexion a)biceps brachii/triceps brachii (BB/TB) b)brachioradialis/anconeus (BR/AN)

The increase of BB/TB ratio is a direct result of an increase in BB EMG activity. Also, the combined contractions of supination-flexion and pronation-flexion do not modulate TB EMG activity. This confirms the facilitatory effect of supination and the inhibitory effect of pronation on BB as previously shown^{1,2}. The stability of BB/TB ratios above 60% MVC could be explained by an equilibrium required between BB and TB EMG activity which allows for a better joint stability⁴. The BR/AN ratios found are more stable. Even though the BR/AN ratio is higher in pronation-flexion than in supination-flexion, it has not reached a significant level ($p > .01$). Finally, it appears that the antagonist EMG activity increases with increased force levels in elbow flexion as shown previously⁶, but this increase of antagonist activity is not statistically significant. From these results, we conclude that in isometric conditions, combined tasks of supination-flexion and pronation-flexion do not modulate the activity of elbow extensors. In addition, cocontraction activity during the combined isometric contractions tends to be similar across elbow angles.

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CARDIAC-LOCOMOTOR COUPLING DURING SUSTAINED NATURAL HOPPING

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During rhythmic muscle contractions, intramuscular pressure may exceed that of systolic blood pressure² thereby intermittently impeding the flow of oxygenated blood to the working muscles. With some types of rhythmic exercise there may be an opportunity to optimize the flow of oxygenated blood to skeletal muscle and to minimize the load on the heart if the peak intra-arterial pressure due to cardiac contraction were to occur when the intramuscular pressure is at or near its lowest point. Coupling of cardiac and locomotor rhythms (1:1) has been identified at some exercise intensities as normal humans walk, run and cycle at cadences natural to them.³

In this study we chose to assess cardiac-locomotor coupling during hopping for several reasons. First, hopping requires that the muscles of both legs contract at the same time, very different from the reciprocating activities already tested. Second, the peripheral arterial resistance should be greater than during alternating rhythmic activities and the potential benefits of coupling in minimizing cardiac load proportionately greater. Third, as other investigators have used hopping to study neuromuscular and stored elastic energy phenomena as determinants of exercise frequency, much is already known about this activity. The object of this study was to test the hypothesis that, during sustained natural two-legged human hopping, coupling of cardiac and locomotor rhythms occurs, with an integer ratio (e.g. 1:1, 2:1, 3:3, 4:3).

METHODS

Eighteen young normal individuals, 10 men and 10 women, were studied with their informed consent. After an identical familiarization session at least 4 days earlier, the subjects were asked to hop on two feet for as long as they could, at a rate and in a manner which was natural to them. Cardiac contraction was represented by the linear envelope of a telemetered electrocardiography signal. For hopping we used a Kistler force platform to generate a rectangular signal each time 500 N was vertically applied to the platform. The signals were amplified, filtered (10-500 Hz) and recorded on FM tape for subsequent analysis. The signals on the FM tape were recorded on a computer after A-D conversion at 200 Hz. Automatic threshold detection of the digitized signals permitted 5-s averages of rates and ratios to be calculated. Coupling of the heart and hopping rates was defined as being present when the two rates (or a multiple of them) were within 1.0% of each other for 4 or more samples in any single minute.

RESULTS

Fourteen subjects coupled at one or more integer ratio -- 5 at a heart:hop rate ratio of 1:1, 4 at 3:4, 3 at 5:4, 2 each at 3:5 and 4:5, and 1 each at 2:3 and 4:3. However, only four subjects showed clear histogram peaks at integer ratios, the most dramatic of which is shown in Figure 1. Figure 2 shows the histogram of all subjects combined. Of the 2830 ratios calculated, 354 of the integer ratios could be expected on the basis of chance alone; we saw 454.

DISCUSSION

If cardiac-locomotor coupling is functionally significant, it should occur in a wide variety of rhythmic exercises. Hopping is not a commonly utilized locomotion method in humans, unlike some other species.¹ It is nevertheless a common activity performed at play, in dance and in ritual ceremonies in some cultures. There are also other forms of hopping (e.g. on one foot), less

sustained hopping, hopping during other activities (e.g. skipping rope) and hopping to the beat of an external signal (e.g. music). Our findings cannot be generalized to these. This study provides limited support for the notion that, for some subjects, cardiac-locomotor coupling is one of the determinants of the preferred rate of sustained natural hopping.

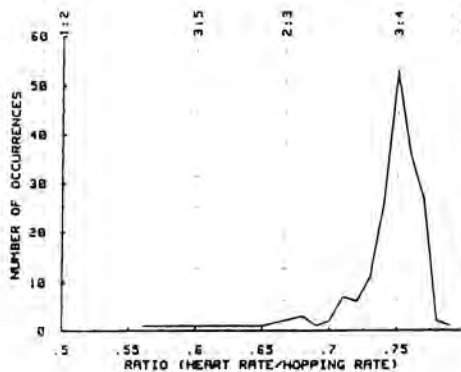


Fig 1. Histogram of a single subject.

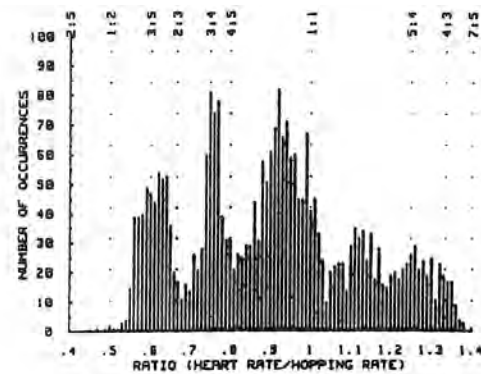


Fig 2. Histogram of all subjects.

Acknowledgement: We thank the AudioVisual Division, Faculty of Medicine and Mrs. E.C. Gregory. This study was supported by the Medical Research Council of Canada.

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DYNAMIC FORWARD INSTABILITY OF A WHEELCHAIR WITH ONE FOOTREST ELEVATED

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Although precise figures on the incidence and circumstances surrounding forward-tipping accidents are not available, they seem to be most common when deceleration occurs while moving forward, particularly down an incline.¹ An elevated wheelchair footrest is most often used by a patient who has recently had an injury to the leg and is wearing a plaster cast. In this study we test the hypothesis that elevating one footrest reduces the dynamic forward stability of an occupied wheelchair, particularly with a simulated plaster cast on the leg and a high-friction floor surface.

METHODS

We studied 20 young normal adults, with their informed consent, in a conventional wheelchair. Subjects descended a 17° ramp by gravity alone, the wheelchair having been released from a position 60 cm from the end of the incline. At the bottom of the ramp was a concrete floor covered with linoleum or a rubber bathmat to represent low- and high-friction surfaces. In a balanced order, the subjects were assessed with both footrests down, with the left leg elevated to 90° (with and without a simulated cast based upon measures of 10 actual casts), and with the leg elevated and the high-friction floor surface (with and without a simulated cast). Symmetry was assumed. We analyzed a slow-motion videotape of each trial using a 4-point ordinal scale: "no tip" if the rear wheels remained in contact with the incline and floor; "transient tip" if the rear wheels lifted temporarily; "full tip" if the rear wheels lifted and remained in the air until the wheelchair had come to rest balanced on the elevated footrest and caster; and "fall" if the occupied wheelchair came to rest on its side on the floor. The conditions were compared using the Wilcoxon matched-pair signed-rank test.

RESULTS

The results are presented in the Table and Figure 1.

Table: Dynamic Forward Stability of a Wheelchair With One Leg Elevated

Stability	Legs Down	One Leg Elevated Without Cast		One Leg Elevated With Cast	
		Low friction	High friction	Low friction	High friction
No tip	18	0	0	0	0
Transient	2	20	2	20	1
Full tip	0	0	8	0	10
Fall	0	0	10	0	9

Each setting was different from the "legs vertical" setting ($p < 0.005$) and the high-friction settings were different from the low-friction ($p < 0.005$).

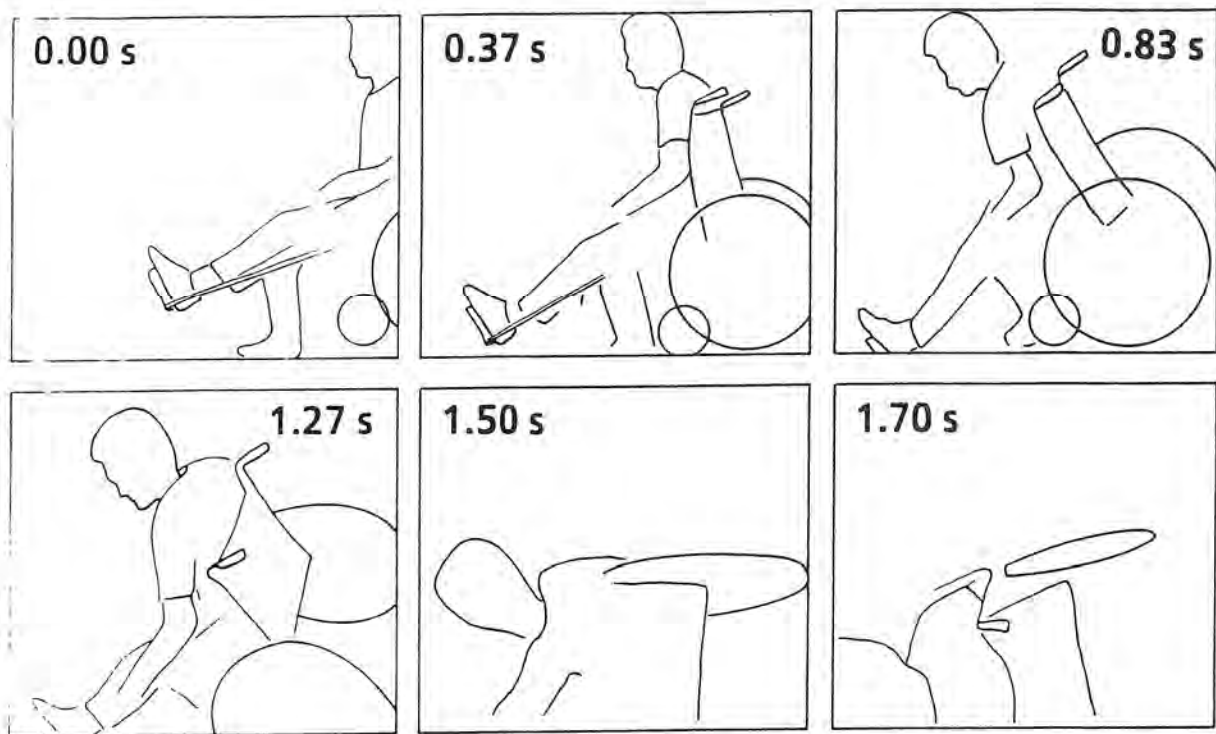


Fig 1 - These line drawings were done from photographs of stop-action videotape. The rotation of the chair to the left was initiated by contact of the lowered footrest with the floor and was facilitated by the freedom of the front wheels to rotate around the vertical axis.

DISCUSSION

Elevating a footrest increases the dynamic forward instability of an occupied wheelchair, particularly on a high-friction floor surface. A simulated plaster cast on the leg has no additional effect. The rough undersurface of the lowered footrest probably accentuated the instability; a smoother profile, with a low-friction material or rollers would be more appropriate. The limitations of this study include the use of a qualitative assessment and an ordinal scale for the dynamic testing rather than a kinematic and kinetic analysis, and the use of able-bodied subjects rather than patients. The results on a single wheelchair and ramp angle should only be generalized with caution. Yet, elevating a footrest significantly increases the dynamic forward instability of occupied wheelchairs, particularly on a high-friction floor surface, a finding with implications for wheelchair design and use.

Acknowledgements: We thank the AudioVisual Division of the Faculty of Medicine, the Nova Scotia Rehabilitation Centre, the Canadian Paraplegic Association and Mrs. E.C. Gregory.

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THE EFFECTS OF SIMULATED KNEE FLEXION CONTRACTURES ON STANDING BALANCE

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Periarticular contractures are among the most ubiquitous problems in rehabilitation, most commonly disabling patients affected with disorders of the neuromusculoskeletal system.⁵ The tissue biomechanics of such contractures have been examined,^{1,3} but little has been done with respect to their consequences. In this study we test the hypothesis that simulated knee flexion contractures affect standing balance.

METHOD

Fifteen normal subjects were studied with their informed consent. Standing balance was tested on a Kistler 9281B multicomponent force platform, the output of which we sampled at 100 Hz (Hewlett Packard 6942A) and stored on a microcomputer (Hewlett Packard 9826). Postural sway (mediolateral [ML] and anteroposterior [AP] travel) and the mean position of the centre of pressure (COP) (as a percentage of the distance between the midlines of the feet and from heels to toes) were determined from 20 s of data. The feet were positioned at a 25° angle to each other (toes out) and 30 cm apart at the mid-heels.² Knee flexion contractures were simulated using an adjustable line from the subject's waist to the sole of each foot. Balance while standing with unilateral and bilateral contractures of 15° and 30° was compared to normal standing balance, using paired t-tests. The average of each subject's data for right and left unilateral contractures was used as there was no consistent and significant difference between them. A level of $p < 0.05$ was considered statistically significant.

RESULTS

Table: Standing Balance with Simulated Knee Flexion Contractures (n = 15)

Measure	Normal Standing	Extent of Contracture			
		15° Unilateral	15° Bilateral	30° Unilateral	30° Bilateral
Travel (cm) ¹					
ML	19.5 (± 13.1)	19.9 (± 12.3)	20.8 (± 14.2)	23.1 (± 13.2)	24.5 (± 12.7)
		0.5 (± 2.6)	1.3 (± 3.6)	3.6 (± 4.4)**	5.0 (± 6.3)**
AP	22.0 (± 12.8)	24.1 (± 15.0)	26.6 (± 13.1)	25.5 (± 13.7)	30.8 (± 18.3)
		2.1 (± 7.1)	4.7 (± 7.5)*	3.5 (± 9.6)	8.8 (± 18.3)
COP (%) ²					
ML ³	43.8 (± 4.8)	4.8 (± 5.8)	0.9 (± 5.9)	15.6 (± 5.1)***	1.4 (± 8.0)
AP	37.5 (± 6.2)	0.9 (± 4.8)	2.8 (± 6.7)	3.7 (± 5.0)	8.3 (± 7.6)**

Level of statistical significance: * = $p < 0.05$, ** = $p < 0.01$, *** = $p < 0.001$.

1 Mean (± 1 SD) values are shown above and mean differences (± 1 SD) from normal standing values below.

2 Except for normal standing, where the mean (± 1 SD) position of the COP is shown as a % of the distance from the right to left foot midlines (ML) and from heels to toes (AP), only the mean differences are shown.

3 A positive value indicates a shift toward the flexed side.

As shown in the Table, with 30° unilateral and bilateral contractures the ML travel increased. AP travel was increased with 15° bilateral contractures, but due to the high standard deviation for bilateral 30° contractures, statistical significance was not reached ($p = 0.08$), although mean values exceeded those with the 15° contractures. With 30° unilateral contractures the COP moved toward the flexed side and with 30° bilateral contractures it moved forward.

DISCUSSION

The results support the hypothesis that standing balance (both the extent of travel and the mean position of the COP) is affected by simulated knee flexion contractures, particularly at the greater extent of contractures (30°). The movement of the COP toward the unflexed side when only one knee is bent is similar to Mahar *et al.*'s ⁴ finding that the COP moves toward the longer leg when leg length discrepancies are simulated. The movement of the COP forward when both knees are bent is consistent with what is expected in the sport "ready position". The limitations of this study include the variability of the data (as shown by the high standard deviations obtained), presumably reflecting differences in the ability of the subjects to maintain standing balance. Using a 90% power calculation and the variability demonstrated, 40 to 100 subjects would be necessary to demonstrate significance with the 15° contractures. A further limitation exists with the use of simulated contractures rather than studying patients before and after resolution of their contractures. The results, nevertheless, provide insight into how knee flexion contractures alter standing biomechanics and they underline the importance of preventing and treating this common disorder.

Acknowledgements: We would like to thank the AudioVisual Division of the Faculty of Medicine and Mrs. E.C. Gregory.

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Three Dimensional Analysis of Ground Reaction Forces For Cutting Maneuvers

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Many of the movements that are of most interest to the athlete, coach, physician and biomechanist occur in more than one plane. The complexity of three dimensional movements requires the development of new tools and techniques for biomechanical analysis and interpretation. Analysis techniques that work well for planar movements are inadequate to characterize the changing force vectors associated with changes of direction. Specifically, the rate of change in direction of the force vector was evaluated for its potential application for studying three dimensional locomotor movements.

Expectations based upon the laws of mechanics suggest that differences in cutting angle or change of direction must be the result of differences in reaction forces. A question of interest, therefore, is how are differences between force time data for change of direction movements to be evaluated? The purpose of the study was to develop techniques for analyzing cutting maneuvers that are sensitive to the mechanical variations in magnitude, direction, and timing exhibited within the force vectors as well as reveal characteristics of individual performance patterns.

METHODS AND RESULTS

Three recreational male athletes between the ages of 18 and 40 having no lower limb clinical pathologies served as subjects. Subjects performed three trials for each of six inside cutting movements (15, 30, 45, 60, 75, and 90 degrees) on the force platform (1000 Hz). The order of performance was randomly assigned to minimize any possible ordering effects. All three trials for each angle were evaluated using the orthogonal (X, Y, and Z) ground reaction force components.

The equipment used to collect data included: a) a force platform b) a datalogger analog to digital converter and storage device, and c) a Compaq 286 digital computer. The angle of cut was measured by having the subjects chalk their shoes prior to performing each trial. Angle of cut was determined by measuring the angle formed by the step preceding the pivot step, the pivot step, and the first step following the pivot step. The angle was measured from the apex of the foot print.

The force vectors generated during the inside cutting maneuver were resolved into three space and graphically displayed in chronological order for analysis. The vector curves that were generated could be rotated on the display to any viewing angle to facilitate the analysis along with the X, Y, and Z axes.

The principal analysis approach employed evaluation of the rate of change of the force vectors. This phenomenon was described using the direction cosine. Average forces in the X, Y, and Z directions, average direction cosines, average velocity of the force vector, and time for each period were computed and evaluated. The force-time curves were divided into four

phases based upon the rate of change of the force vector. Phase I commenced at foot contact and continued until the instant when the force vector changed direction significantly in the X-Y plane. Phase II included the time from the end of phase I until the force vector velocity approached or was less than 10 rads/s. Phase III began when the force vector was changing direction at 10 rad/s or less and ended when the force vector proceeded toward the origin. Phase IV began when the force vector proceeded toward the origin and ended when the force vector was the zero vector.

SUMMARY

Linear gait force-time curves have been characterized by two phases, an impact phase and a loading phase. In contrast a three dimensional analysis of inside cutting maneuvers identified four descriptive phases for these movements: 1) impact phase, 2) rapid change of direction phase, 3) constant direction phase, and 4) lift off phase.

As the severity of the angle of cut increased the proportion of the time spent in phases I, II, and IV decreased and the time spent in phase III increased (see Tables 1 & 2). Based upon these results, phase III is hypothesized to be the most important portion of the activity for controlling the movement. This phase was characterized by a relatively stationary force vector.

The ability to recognize individual patterns is encouraging and suggests that the method has merit for evaluating multidirectional gait activities. The results from this preliminary study indicate that the force patterns were consistent across the different cutting angles for individuals. The observed differences in force patterns between individual subjects were, however, quite variable.

Table 1. Average Time on Force Platform of Relative Phases (sec)

Angle (deg)	Phase I	Phase II	Phase III	Phase IV	Total
15	0.012	0.057	0.084	0.098	0.252
30	0.010	0.071	0.089	0.108	0.279
45	0.010	0.066	0.103	0.101	0.281
60	0.009	0.064	0.127	0.118	0.317
75	0.013	0.061	0.172	0.108	0.353
90	0.011	0.050	0.191	0.134	0.386

Table 2. Percent of Time on Force Platform

Angle (deg)	Phases 1&2	Phase 3	Phase 4
15	27.3	33.3	38.9
30	28.9	31.9	38.7
45	27.0	36.7	35.9
60	23.0	40.1	37.2
75	21.0	48.7	30.6
90	15.7	49.5	34.7

Muscle Activation Changes Associated with Loads at Adjacent Joints *or* Why Muscle Moment Arms are not Sufficient to Predict Functional Role

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The control of multiple muscles about a single joint or in a multi-joint system is a matter of great interest to the biomechanician or motor control physiologist. We have found that in constructing models that predict muscle activation levels, there are some common assumptions that are often made that may be without warrant. The most common of them, which seems quite intuitive, is that a muscle's action can be predicted solely by its moment arms. For example, a muscle that has a moment arm that would mechanically enable it to assist in flexion would be predicted to be activated during flexion tasks and not during extension. Although this may appear to be a common sense assumption, it has not been examined in any detail. It is particularly important when considering how muscles are used during multi-joint tasks. If a single-joint muscle is modulated during load changes at a second joint, then it could be argued that its activation cannot be predicted solely on the basis of its biomechanical contributions (since it would have no contribution at the second joint).

The purpose of this project is to present the idea that muscle force cannot be predicted solely on the basis of its moment arms and the loads at the particular joints at which it acts, but rather must be examined in the context of the loads at nearby joints as well, even if the muscle of interest has no mechanical contribution at those particular joints.

METHODS

We recorded a combination of surface and intramuscular EMGs from biceps brachii (BIC), brachialis (BRA), brachioradialis (BRD), triceps brachii (TRI), and, in most subjects, the pronator teres (PRO) in 13 different subjects for 39 different data sets during various isometric tasks. The subjects were seated with their shoulders at 0° flexion, 65° abduction, their elbows at 90° flexion, and their wrist at one of three fixed positions: neutral, pronation, or supination. They were instructed to match a 3-d.o.f. force target indicating a particular flexion/extension, varus/valgus, and pronation/supination load. The load state was recorded via a 3-d.o.f. strain gauge apparatus attached to the subject's wrist. They were asked to maintain that particular load for 2½ seconds and to keep their wrists flail during the experiment. Three forces along with EMGs were digitally recorded for each trial and were later analyzed. Data was collected at eight flexion/extension/varus/valgus load directions (covering a complete circle orthogonal to the axis of the wrist) for three load magnitudes at each of three wrist positions. After the experiment, the EMGs were rectified and averaged over the trial period and then averaged again for each repetition.

RESULTS

An example of the type of activation patterns we have observed is shown in figure 1. Each plot represents a given elbow extension load of 30% of maximum with superimposed pronation loads of five magnitudes (horizontal axes), against which are plotted the corresponding EMGs normalized to maximum (vertical axes). Note that the BIC activity increases with supination load. This is curious because this is a flexor acting during extension tasks. The BRD, as well as the BRA, showed little change in activation, as they were relatively inactive. The PRO was active during pronation torques for these extension tasks, despite its being a mild flexor. Finally, the TRI showed a V-shaped pattern, with a minimum at or near the neutral pro-supination torque. This V-shape was observed 21/39 times for the TRI and is

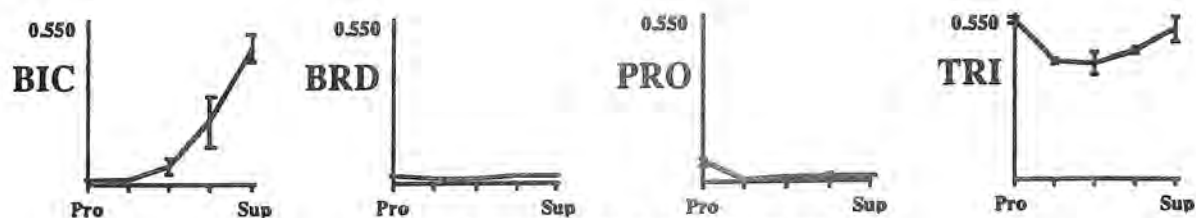


Figure 1: Normalized EMG vs pro-supination load.

noteworthy because the TRI can act as neither a supinator nor a pronator. Based solely on its anatomical location and the biomechanical advantage of its moment arms, one would expect it to act at a fixed level independent of pro-supination load.

The most likely explanation, in our opinion, for these EMG patterns is as follows. Although the subject is instructed to produce the same amount of extension force for all of the extension trials, when pronation torques are superimposed, the PRO becomes active to supply the need. This then produces extra flexion force, requiring the TRI to become more active to compensate. At the middle of the plot, with no additional forearm loads, the PRO can deactivate and the TRI can decrease. However, as the subject begins to create supination torque, the BIC must become active and therefore the TRI must once again increase its level of activation in order to compensate for the increased flexion load (again provided that a constant amount of extension force is to be maintained). Here we see that muscles not related to supination/pronation must be modulated to compensate for inappropriate activation by multi-joint muscles. Interestingly, this implies that a muscle's activation level cannot be predicted solely on the basis of the external load on the joints at which it acts, a common assumption of biomechanicians (e.g. 1, 2, 3).

DISCUSSION

Why is a muscle called a flexor or a supinator or an abductor? It is determined by the muscle's anatomy—its origin and insertion sites—and not by its activation. Biomechanicians often predict a muscle's activation based solely on its geometry (e.g. 1, 2, 3). However, if the EMG for a muscle increases with increasing supination load, is it a supinator, even if biomechanically (i.e., as determined by its moment arm) it has no mechanical contribution to supination? Will not such action essentially cause it to act as a functional supinator?

The TRI for example, in this study acts with increasing activity as the amount of supination load is increased beginning from a neutral pro-supination load. However, it has no moment arm to contribute to supination torque production. It also shows similar activity with pronation torque. These types of activities could not be predicted by solely examining the muscle's moment arm. Nor could they be predicted by examining all of the muscles that act at the elbow joint (i.e., flexion/extension) alone. If such actions were to be theoretically predicted, they would have to be based on a model of the true elbow joint (flexion/extension) that also included the forearm (supination/pronation). This is an important observation that biomechanicians are prone to miss in most models. The activation of muscles can be significantly influenced by the loads at adjacent joints at which the muscle may have no mechanical advantage.

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BIOMECHANICAL CHANGES IN UPRIGHT STANCE CONCURRENT WITH AGING

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INTRODUCTION

To understand changes in the postural control system with age we have focused on biomechanical characteristics of upright stance. Kalchthaler et al (1978) documented a marked increase in the incidence of falls in the elderly. It is reasonable that deficiencies in this control system may be associated with the balance deficits of older persons. Quantification of sway has been accepted as an appropriate measure of postural stability and many studies have sought to quantify postural sway in the aged while standing erect. The physiological significance of sway is unknown; however, increased sway is thought to represent instability. From a mechanical standpoint, the control system must maintain the COG over the base of support. Sway may be evidence of control deficits or an indication of the control strategy itself. Sheldon (1963) first suggested that the inability to control sway in advancing years may result in falls. Investigation of age-related changes in postural sway have demonstrated significantly increased sway amplitudes in the aged during quiet stance with eyes open and eyes closed (Brockhurst, 1982). Therefore, the influence of vision and other subject variables must be accounted for when assessing that of aging. The purpose of this study was to evaluate the relationship of biomechanical measures of upright stance with aging to characterize age-related changes in postural control.

METHODS

Twenty-two normal subjects (age 21-78; 11 female, 11 male) volunteered to participate in this study. All subjects signed consent forms and were given a complete neurological examination including visual acuity, peripheral fields and other visual characteristics. Subjects in this study demonstrated no evidence of visual or neurological deficit upon examination. Kinematic data were sampled (50 Hz) with a modified video based infrared marker detection system (VICON). Four video cameras recorded the two dimensional position of reflective markers (lightweight 25 mm foam spheres) which were attached at eight bony landmarks on the right side of the body. Bilateral symmetry was assumed and therefore kinematic characteristics were calculated in the vertical (Z) and anterior-posterior (Y) directions. Kinetic data were obtained (200 Hz) from a force platform (AMTI) which provided center of pressure (COP) measurements in the medial/lateral (X) and anterior/posterior (Y) directions. Subjects stood upright with a freely chosen foot position for 30 s with eyes open and eyes closed. Data were collected on a PDP 11/73 and subsequently transferred to a VAX 11/750 for reduction and analysis. Three dimensional coordinates (3D) were generated from the four camera views using software specifically designed for this system (AMASS, Adtech Co.). Center of gravity (COG) of a six segment model was calculated from 3D segment spatial coordinates, segment masses and inertial characteristics based on data from Dempster (1955). Kinetic measurements were collected and scaled with the same system and COP coordinates temporally coincident with COG position values were extracted from the force platform data sets. Kinematics were calculated over a twenty second period (1000 samples), commencing ten seconds after the trial began and were evaluated in each direction of whole body (COP and COG) and segmental (head, shoulder, hip and knee) motion. The total displacement of the COP is commonly used as a measure of sway. Herein, this measure was called the total excursion (TEX) and was calculated from the sum of the absolute value of the displacements of a given coordinate. For each video frame $i=1$ to n , marker position $i = x_i, y_i, z_i$; where

TEX=total excursion and abs=absolute value:

$$TEX_x = \sum \text{abs}(x_{i+1} - x_i)$$

$$TEX_y = \sum \text{abs}(y_{i+1} - y_i)$$

$$TEX_z = \sum \text{abs}(z_{i+1} - z_i)$$

To evaluate movement variability, the standard deviation about the quadratic fit to the respective data set was calculated. The quadratic equation predicts: $ax_i^2 + bx_i + c = 0$; therefore the deviation about the quadratic fit (VAR) can be expressed:

$$VAR_x = \sqrt{[\sum (x_i - x_i')^2 / (n-1)]}$$

The relationship of subject age, gender and vision with these whole body and segment measures of TEX and VAR was assessed using a stepwise regression procedure. Only results related to age are discussed herein. Individual traces of each whole body data set were digitized to obtain measures of the amplitude, direction and frequency of postural adjustments.

RESULTS AND DISCUSSION

Many variables were significantly related to age. While both VAR and TEX of COPY increased concurrent with aging, only VAR of COGY was significantly related to age. Therefore, although greater anterior/posterior VAR is evident, the COG is maintained over the base of support. This finding emphasizes the importance of ankle joint adjustments in the control of standing posture, particularly in the absence of vision. In addition, analysis of individual segment kinematics demonstrated that although VAR of segment motion increased with age (VAR head, shoulder, hip Y; head, hip, knee Z), TEX was not significantly related. The relation of aging with increases in VAR in the absence of a concurrent increase in TEX suggests a change in the postural control strategy with age which appears to be equally effective in maintaining upright stance. The nature of the control function may be better understood by evaluating these postural adjustments. The magnitude of the adjustments was measured from positive and negative deviations from the baseline, in each direction (X,Y,Z). The average magnitude was tested with a stepwise regression. The size of the upward vertical (Z) adjustments of the COG significantly ($p<.01$) increased with age as did the anterior/posterior (Y) and medial/lateral (X) adjustments measured by the COP. In order to more fully understand the changes in postural control strategy which were related to age, we divided our subjects into two categories (old 57-78, young 21-56; 11 old, 11 young). Additional aspects of the whole body adjustments were examined using factorial ANOVA methods and adjusting the alpha values accordingly. The number of adjustments of COPX and COPY were significantly ($p<.01$) greater in the older subjects although COG adjustments were not significantly different. Finally, a Repeated Measures design was used to compare first and last adjustments which also were found to differ significantly ($p<.005$) for COPX and COPY. These results suggest that frequent and irregular ankle joint adjustments may characterize a different postural control strategy. The older subjects appear to maintain a fairly rigid posture and utilize large and inconsistent adjustments to maintain stability.

CONCLUSIONS

1. Increasing variability of segment motion was strongly associated with aging. COP variability and total excursion and COG variability but not total excursion increased with age.
2. The older adults made significantly larger adjustments to vertical posture and this was the only COG adjustment related to age.
4. Increases in the number and magnitude of COP adjustments were consistently related to aging although TEX COG indicated comparable stability. The postural adjustments evident in the older subjects, particularly those indicated by COP, were irregular and varied significantly from the beginning of the trial to the end.

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ACKNOWLEDGEMENTS

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STUDING THE AERODYNAMICS OF LOW CENTRE OF GRAVITY JAVELINS

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In the 23rd Olympic Games held in Los Angeles, the International Amateur Athletic Federation resolved that the improved men's javelin would be used from April 1, 1984. The new javelin is called a low centre of gravity javelin because of shifting the centre of gravity 40mm to the point of the javelin. The author in this project experimented with two improved javelins in a wind tunnel and attempted to probe into a subject relevant to aerodynamic.

The experiment was carried out in China Aerodynamics Research and Development Centre (CARDC). The wind tunnel section area was 8m X 6m. A 15mm stick strain balance was installed within the javelin. The wind velocity was 24, 30, 34 m/s. The range of angle of release was -10° — 45° .

Analyzing the experiment gives:

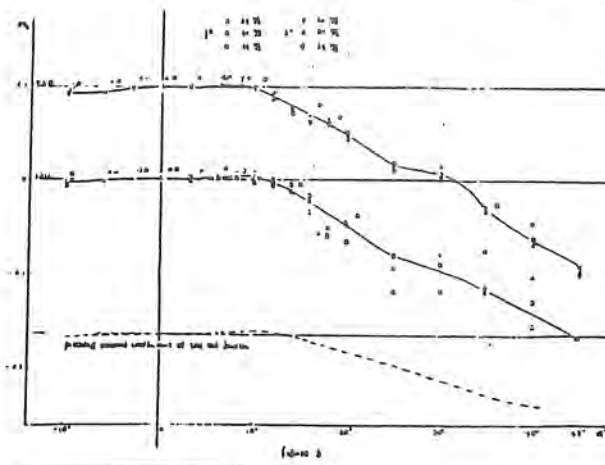
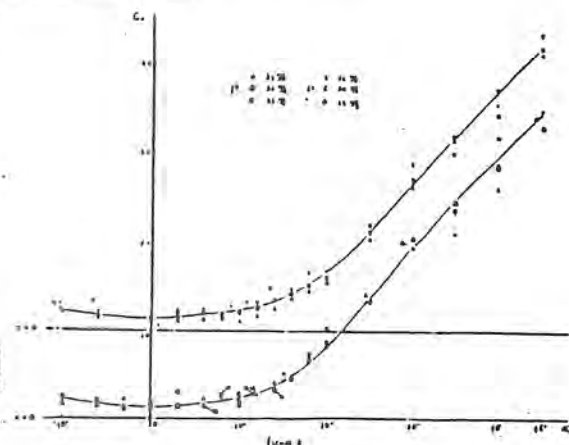
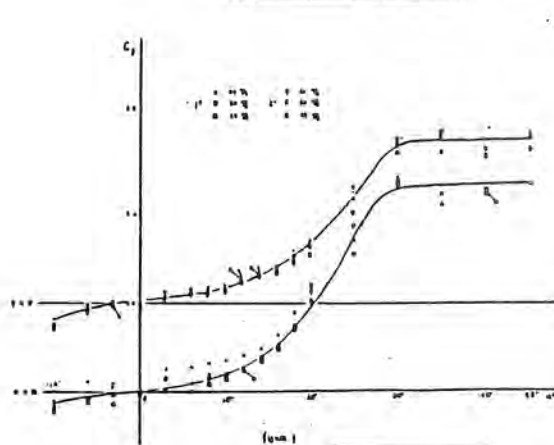
1. The drag affecting the improved javelin increases with a rise of the angle of release when the angle is within 45° , while the lift does not after the angle going beyond 30° .
2. The arm of force from the javelin's centre of gravity to the centre of air pressure is not well related to the initial speed but varies considerably with the variation of the angle of release.
3. The drag to lift ratio is either not well related to the initial speed. It increases with the rise of the angle of release. The javelin 1, $K_{\max} = 1.29$, the javelin 2, $K_{\max} = 1.06$. Both their angle of release are 25° . The javelin is thrown at 25° . It has a higher fly but a poorer horizontal speed. For this reason the athletic result could not be raised.
4. To bring the aerodynamic character into play, and to raise athletic results, a proper condition of throwing release is necessary. The author suggests to make a simulated throwing calculation that provides the theoretical basis for athletic

trains. It is then judged in athletic practice.

See table 1, place of drag to lift ratio and centre of pressure; figure 1, drag coefficient of the javelin; figure 2, lift coefficient of the javelin; figure 3, pitching moment coefficient of the javelin.

table 1

α°		-10°	-6°	-3°	0°	3°	6°	8°	10°	12°
1 st javelin	K	-0.64	-0.29	-0.14	0	0.29	0.47	0.68	0.84	1
	X_p	0.036	0.033	0	-0.05	0.033	0.065	0.015	0.011	-0.015
2 nd javelin	K	-1.10	-0.56	-0.21	0	0.36	0.50	0.58	0.71	0.84
	X_p	0.026	-0.01	-0.015	-0.035	0.03	0.067	0.018	-0.0067	-0.045
α°		14°	16°	18°	20°	25°	30°	35°	40°	45°
1 st javelin	K	1.09	1.13	1.16	1.19	1.29	1.18	0.95	0.79	0.68
	X_p	-0.037	-0.048	-0.052	-0.054	-0.047	-0.043	-0.051	-0.061	-0.075
2 nd javelin	K	0.84	0.93	0.98	1.00	1.06	1.03	0.83	0.68	0.58
	X_p	-0.074	-0.081	-0.08	-0.078	-0.069	-0.054	-0.071	-0.09	-0.10



A METHOD FOR COMPUTING THE THREE-DIMENSIONAL FORCES AND MOMENTS IN THE LOWER LIMB DURING GAIT

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INTRODUCTION

The forces and moments internal to the human body may not be directly measured. Therefore, models of the body are developed to estimate these reactions using the principles of rigid-body mechanics. The diversity of the human body creates a complex set of equations to be solved. Researchers often view the limb segments in two-dimensions or omit specific terms in the equations in order to simplify the solution process. The objective of this study was to formulate the complete three-dimensional Newton-Euler equations of motion for the lower limb and to solve them for the reaction forces and moments at the ankle, knee and hip joint linkage centers during stance phase of locomotion.

METHODS

The right lower limb was considered to be a series of three interconnected rigid bodies; the foot, the shank and the thigh. The complete three-dimensional Newton-Euler equations of motion were written for each segment, with respect to that segment's center of mass and principal axes, 1, 2 and 3, as follows:

$$\begin{aligned} m\bar{a}_1 &= m\bar{g}_1 + \bar{R}_1 + \bar{F}_1 \\ m\bar{a}_2 &= m\bar{g}_2 + \bar{R}_2 + \bar{F}_2 \\ m\bar{a}_3 &= m\bar{g}_3 + \bar{R}_3 + \bar{F}_3 \end{aligned} \quad (1)$$

$$\begin{aligned} \bar{M}_1 + \bar{T}_1 + (r_2 \bar{F}_3 - r_3 \bar{F}_2) + (p_2 \bar{R}_3 - p_3 \bar{R}_2) &= I_1 \dot{\bar{W}}_1 - \bar{W}_2 \bar{W}_3 (I_2 - I_3) \\ \bar{M}_2 + \bar{T}_2 + (r_3 \bar{F}_1 - r_1 \bar{F}_3) + (p_3 \bar{R}_1 - p_1 \bar{R}_3) &= I_2 \dot{\bar{W}}_2 - \bar{W}_3 \bar{W}_1 (I_3 - I_1) \\ \bar{M}_3 + \bar{T}_3 + (r_1 \bar{F}_2 - r_2 \bar{F}_1) + (p_1 \bar{R}_2 - p_2 \bar{R}_1) &= I_3 \dot{\bar{W}}_3 - \bar{W}_1 \bar{W}_2 (I_1 - I_2) \end{aligned} \quad (2)$$

where m is the mass of the segment, $\bar{a}$ is the acceleration of the center of mass, $\bar{g}$ is the gravity vector, $\bar{R}$ is the distal reaction force, $\bar{F}$ is the proximal reaction force, $\bar{M}$ is the distal reaction moment, $\bar{T}$ is the proximal reaction moment, $\bar{p}$ is the vector between the center of mass and the distal linkage center, $\bar{r}$ is the vector between the center of mass and the proximal linkage center, I_1 , I_2 and I_3 are the principal moments of inertia, $\dot{\bar{W}}$ is the angular acceleration and $\bar{W}$ is the angular velocity.

Three dimensional position data were collected for fourteen targets placed on the right lower limb and hip of four subjects during stance phase of walking gait (Figures 1, 2 and 3). Force plate data were simultaneously recorded and the corresponding center of pressure was computed for each instant of time. The three-dimensional position data were filtered and differentiated to provide linear velocities and

accelerations of each target. Total body weight, limb length and other anthropometric measurements were also recorded for each subject.

The location of the center of mass and the directions of the principal axes of each segment were determined from a review of the literature and expressed with respect to specific anatomical landmarks and anthropometric measurements. Segment masses were determined to be proportional to total body mass. The principal moments of inertia were computed from geometric models developed to reflect the mass distribution characteristics of each subject. A numerical method, based on the Method of Least Squares, was used to determine the angular velocity and acceleration of each rigid body segment. All vector quantities were transformed into the appropriate local segment coordinate system and the equations of motion were solved for the components of the force and moment acting at the proximal linkage center.

RESULTS AND CONCLUSIONS

The forces and moments computed at the ankle, knee and hip joint linkage centers compared favorably with similar results presented in the literature. The choice of coordinate system in which these reactions were expressed, (ie. local segment coordinate system vs. lab coordinate system), was found to have an effect on the presentation of the results. The contributions of each term in each equation was determined and it was concluded that, although some dynamic terms showed a minimal contribution at various moments in stance, a complete three-dimensional analysis was necessary to adequately describe the forces and moments in the lower limbs during locomotion.

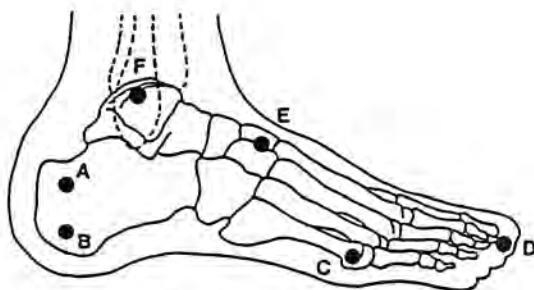


Figure 1.

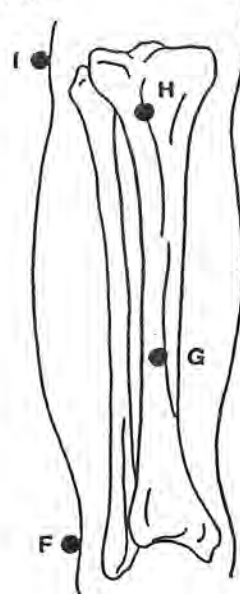


Figure 2.

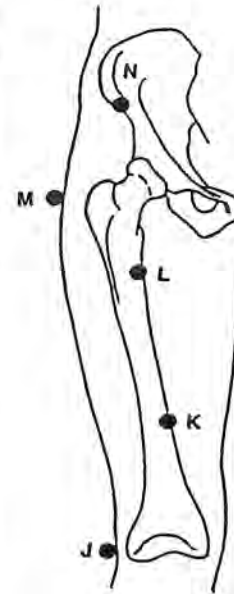


Figure 3.

Torsional movements of the foot during landing

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1 Introduction

In many sporting activities the risk of injury during landing after a jump is considered high, because large acting moments can force the foot into distortion. These moments are exerted on the foot due to high vertical and horizontal forces in combination with landing positions of the foot which are prone to injury (Stacoff, Kälén and Stüssi, 1986; Stüssi, Stacoff and Tiegermann, 1987). The risk of getting injured is increased even further if a torsionally stiff shoe sole forces the foot over the lateral border of the shoe (Segesser, Jenoure, Feinstein and Vogt-Sartori, 1986). The goal of this investigation was to check the hypothesis, whether the stiffness of the shoe sole can influence the torsional movement between forefoot and rearfoot and hence decrease the risk of injury.

2 Methods

In the testing condition the subjects had to land on a ramp sloped by 20 degrees which moves the forefoot into supination. This simulated a landing with large horizontal forces or a landing on an opponent's foot (Figure 1). The forefoot touched the ground first as it can be observed in landing during a sporting activity. The movements of the heel were only influenced by the construction of two different types of shoe soles (a) with low torsional stiffness, and (b) with high torsional stiffness. Low stiffness was achieved by grooves in the shoe sole. For comparison measurements under barefoot condition were also included. The movements were filmed from behind at 100 Hz and the angular displacements of the forefoot, rearfoot and lower leg were measured. The torsion angle was defined as the relative movement between forefoot and rearfoot about

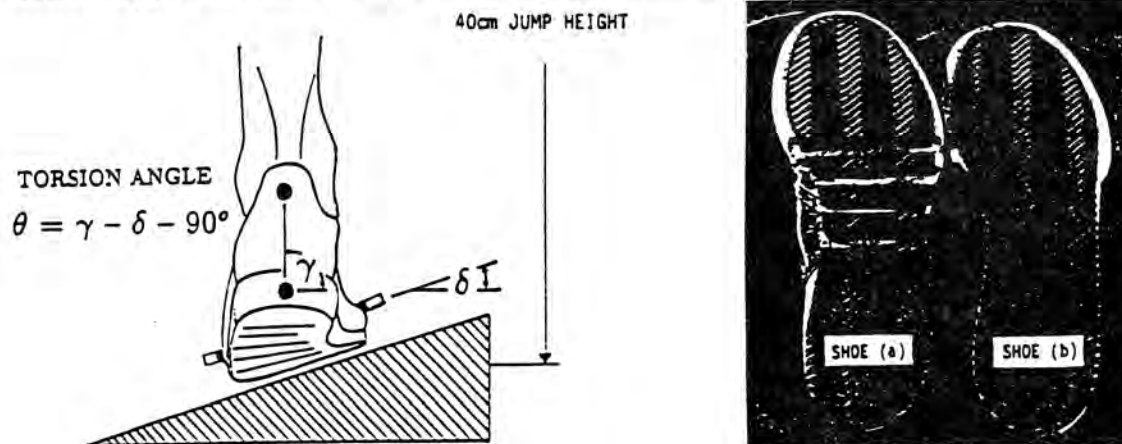


Figure 1: Set-up of the test, definition of angles and shoes used for the study with (a) low and (b) high torsional stiffness

the long axis of the foot. Three subjects participated in the study, repeating each test condition [barefoot, shoe (a), shoe (b)] ten times. For the statistics the H-test after Kruskal and Wallis was used.

3 Results

The results show (Figure 2) that under the given conditions the barefoot twists between forefoot and rearfoot up to 20 degrees. This movement is referred to as torsion and is described in the anatomical literature as movements in the Chopart's and Lisfranc's joints (Kapandij, 1970). With the torsionally stiff shoe only 10 degrees of torsion was measured, in contrast to the shoe with low torsional stiffness which was closer to the barefoot values. The angular displacement of the two shoes started to differ at about 20 ms after touchdown. After 70 ms the results of the two shoes differed significantly for all subjects with a $p=0.05$.

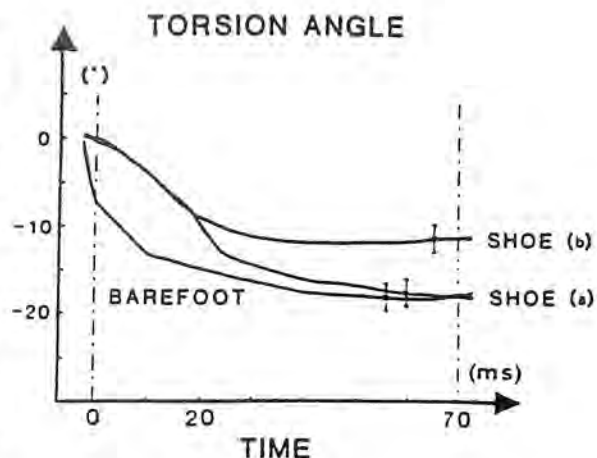


Figure 2: Mean change of torsion angle during landing, subject 1, std=1.5°, 10 repetitions

4 Discussion and conclusions

When landing on an uneven ground or with a large sideways component a torsional movement between forefoot and rearfoot takes place. This movement can be influenced by the shoe sole construction. A shoe with a grooved sole [shoe (a)] allows a torsional movement of the foot closer to the barefoot condition. As a consequence the rearfoot remains in a more vertical and stable position during landing compared to a shoe with no grooves [shoe (b)] and high torsional stiffness. Since the two shoe types start to differ after 20 ms they can influence the landing movement of the foot even before the lower leg muscles which are reported to tension at 70 ms after touchdown (Thonnard, Plaghki, Willems, 1986). It is concluded that in landing after a jump a shoe with a low torsional stiffness (i) increases the torsional movement of the foot which (ii) can help to reduce the risk of injury.

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COMPARISON OF PEAK AND CONSTANT ANGLE TORQUE-VELOCITY CURVES DURING ECCENTRIC AND CONCENTRIC ARM FLEXION IN HIGH STRENGTH AND LOW STRENGTH INDIVIDUALS

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There is a theoretical disagreement about the use of peak (PT) versus constant angle torque (CAT) to establish the torque-velocity curve (TVC). Because angle of PT significantly varies with velocity of muscular contraction and muscle length, prior studies have suggested that the TVC may be confounded by using PT (Thorstensson et al., 1976; Lesmes et al., 1978; Coyle et al., 1979) instead of CAT (Perrine & Edgerton, 1978; Gregor et al., 1979; Caiozzo et al., 1981). Studies using CAT as the criterion score reported "flattening" of the TVC near zero velocity, whereas studies employing PT reported TVCs that resembled the hyperbolic force-velocity relationship found in animal preparations. In addition, Yates and Kamon (1983) concluded that TVCs using PT were more sensitive to influences produced by muscle fiber type than TVC generated from CAT.

The purpose of the present study was to examine the effects of strength level on the TVC using PT and CAT at 90° and 150° during an arm flexion exercise. Forty college males (age 22.8 years, mass 83.7 kg, height 178.8 cm) were placed into either a high strength (HS, N=20) or low strength group (LS, N=20). Peak torque and CAT at 90° and 150° were measured at -2.09, -1.57, and -0.52 rad·s⁻¹ during eccentric (EC), and at 0.52, 1.57, 2.09, 6.28, and 7.85 rad·s⁻¹ during concentric (C) arm flexion with the Biodex computerized dynamometer (Shirley, NY, USA). A 3-factor analysis of variance with repeated measures revealed significant differences in the angle of PT during EC and C arm flexion (range 104.9° to 132.4°; $F=27.5$, $p<0.05$), with no significant differences between groups ($F=0.62$, $p>0.05$). However, there were significant between-group differences in PT, CAT at 90° and 150° during EC flexion ($F=18.5$, $p<0.05$) and during C flexion ($F=24.8$, $p<0.05$). There were also significant within group differences for EC PT at 3 velocities ($F=3.61$, $p<0.05$) and for C PT at 5 velocities ($F=141.0$, $p<0.05$). EC CAT ($F=92.3$, $p<0.05$) and C CAT ($F=69.6$, $p<0.05$) were also different within groups. Scheffe's post-hoc contrast revealed that the significant torque by velocity interaction ($F=8.8$, $p<0.05$) did not occur between TVC for PT and CAT, but between the 2 TVCs for CAT (Figure 1).

The above findings agree with the reports by Thorstensson et al. (1976), Lesmes et al. (1978), Coyle et al. (1979), and Yates and Kamon (1983) who inferred no substantial deformation in the shape of the TVCs when using PT compared to CAT. In addition, these authors suggested that the TVC using CAT would be more sensitive to individual differences in muscle fiber type than TVC using PT (e.g. Yates and Kamon, 1983). In the present study, there were significant differences between HS and LS in C and EC torque at every speed, but there was no interaction between groups and the TVCs using PT or CAT. Thus, the TVCs did reflect the large individual differences in muscular strength when subjects were categorized by strength level, but these differences in strength were independent of the method for establishing the TVCs.

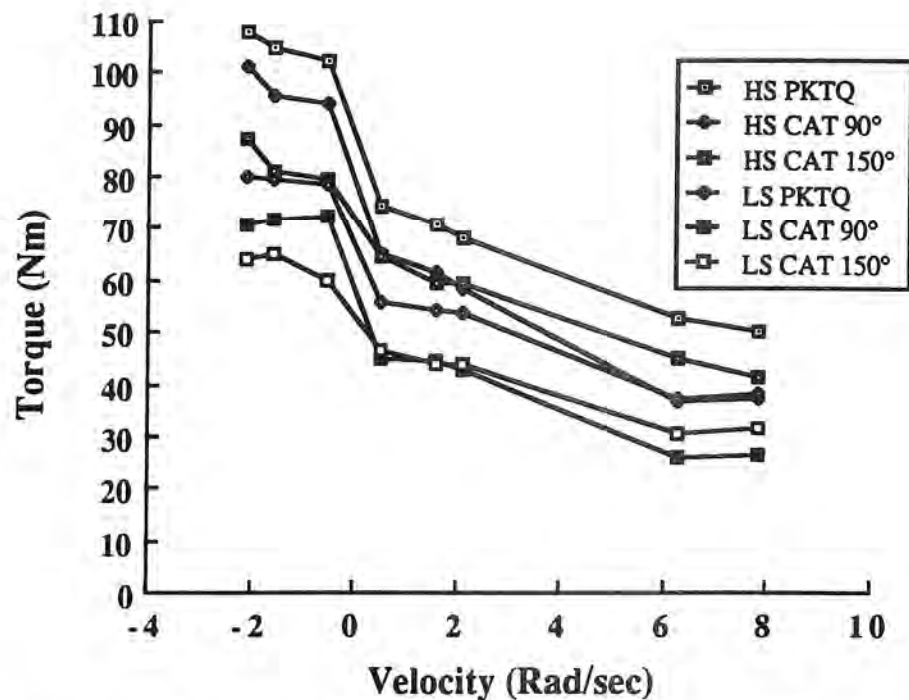


Figure 1. Comparison of peak (PKTQ) versus constant angle torque (CAT) at 90° and 150° during eccentric and concentric arm flexion exercise in high strength (HS, N = 20) and low strength (LS, N = 20) groups.

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EFFECTS OF FATIGUE INDUCED BY STRETCH-SHORTENING EXERCISE ON VERTICAL JUMP, MAXIMAL ISOMETRIC FORCE, AND PATELLAR REFLEX

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Prior studies have explored the fatigue effects of isometric (Enoka et al., 1980), isotonic (Kroll, 1974), and stretch-shortening cycle exercises (Gollhofer et al., 1987) on various measures of voluntary and reflex properties of human skeletal muscle. Considerable evidence supports the tenet that fatigue of skeletal muscle arises postjunctionally (Edwards, 1981) yet may to some extent be compensated by motoneuronal or other neural factors (Stephens and Taylor, 1972). However, very few studies have considered such compensatory mechanisms to be operant through post fatigue changes in the recruitment pattern of prime movers during a performance task (Greig et al., 1985; Nilsson et al., 1985).

The purpose of the present study was to examine the effects of fatigue induced by 50 depth-jumps (DJ, 0.6 m) on vertical jumping ability, maximal isometric knee extension force (MVC), and patellar reflex profile. Twelve college males (27.3 yrs, 176.9 cm, and 75.2 kg) reported for baseline testing on Day 1 and again on Day 2 before and after the 50 DJs. Three trials of 3 types of jumps: counter-movement (CMJ), squat (SJ), and DJ, were employed with hands on the hips to assess vertical jump height computed from flight time. All jumps were performed on a metal plate so that take-off and landing occurred in the same vertical plane. Three trials of 5 s contractions were used to measure MVC at 90° in the seated position. Rate of tension development (RTD_{slope}) was measured during 3 additional trials of MVC. Patellar reflex, using 10 trials, was elicited by dropping a reflex hammer from 60° to the patellar ligament. Surface EMG of the vastus lateralis (VL) and biceps femoris (BF) was monitored with preamplified electrodes (Motion Control Inc.) during all jumps and reflex measures. Day-to-day reliability (intraclass correlation) for the criterion measures ranged between $R = 0.88$ to 0.91 . Two-way ANOVA revealed no reduction in the height of the 50 DJs ($F = 0.69$, $p < 0.05$), but a 16% significant lengthening of contact time at the 50th compared to the 5th jump ($F = 1.99$, $p < 0.05$, Table 1.). Height of CMJ and DJ did not change after the 50 DJs ($p > 0.05$), but the duration of the EMG burst for VL and BF averaged 13.4% longer ($p < 0.05$). Conversely, height of SJ declined by 11.3% ($p < 0.05$). There was a 20.1% and 22.9% decrease in MVC and RTD_{slope} ($p < 0.05$). No significant changes occurred in reflex latency and duration of action potential (AP, $p > 0.05$), but reflex time, AP dual time, and AP amplitude increased by 6.4%, 6.1%, and 23.3% ($p < 0.05$).

The current findings suggest that despite a significant loss of voluntary force, performance in CMJ and DJ remained unaltered by an earlier onset (and longer duration) of the agonist and antagonist muscle groups. This longer duration of muscle activity may be related to a larger amplitude and/or a slower rate of muscle stretch during the eccentric phase that allowed for the generation of sufficient propulsive force after the 50 DJs. In addition, the increased peak-to-peak AP amplitude suggest an enhanced excitability of the patellar reflex. We conclude that to maintain performance in CMJ and DJ following 50 DJs, subjects compensated for the the loss of voluntary force by lengthening the activity duration of the prime movers and enhancing muscle spindle sensitivity.

Table 1. Effects of 50 depth vertical jumps on vertical jump, associated electromyographic patterns, MVC, and reflex profile (N = 12).

	Pre		Post		%Δ	t ^a
	Mean	±SD	Mean	±SD		
CMJ, cm	32.4	3.84	31.9	4.31	1.5	0.79
SJ, cm	22.1	4.41	19.6	3.94	11.3	2.32*
DJ, cm	34.8	4.12	34.1	5.21	2.0	0.07
CMJ						
Ag. duration ^b	524.8	83.48	610.7	101.04	16.4	2.22*
Ant. duration	470.2	78.31	512.7	118.00	9.0	2.20*
SJ						
Ag. duration	488.7	76.95	552.8	86.17	13.1	2.40*
Ant. duration	478.9	74.71	539.9	70.70	12.7	2.67*
DJ						
Ag. duration	591.1	114.64	674.1	123.72	14.0	2.58*
Ant. duration	560.1	119.79	638.3	136.47	14.0	2.70*
Contact time	413.6	66.75	467.9	75.59	13.1	3.40*
MVC, Kg	92.9	21.99	74.2	19.65	20.1	4.43*
RTD _{slope} , Kg/s	32.3	8.51	24.9	7.83	22.9	4.32*
Reflex latency, ms	20.49	1.052	20.36	1.196	0.6	0.77
Reflex time, ms	91.55	6.328	97.41	6.039	6.4	6.24*
Dual time, ms	21.95	1.589	20.61	1.731	6.1	3.29*
AP duration, ms ^c	9.41	2.816	9.61	2.562	2.1	0.35
AP ampl., μV	232.11	141.812	321.33	166.882	38.4	2.67*

^a Paired two-tailed t-test. ^b Duration of agonist and antagonist EMG activity, ms.

^c Duration of action potential. * p<0.05.

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CHANGES IN PEDALLING TECHNIQUE OF ELITE CYCLISTS WITH INCREASED WORKLOAD

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Introduction: While it has been reported that cyclists increase the effectiveness of their pedalling technique as workload increases (1,2), no data base exists for the changes in pedalling techniques of elite subjects under conditions similar to competition. Specifically, we tested three hypotheses with regard to elite cycling: (a) subjects selectively changed the pedal orientation; (b) subjects increased upward forces during the upstroke; and (c) subjects utilized a greater range of the crank cycle for force production.

Methods: A specially instrumented pedal (3) and other instrumentation on the bike recorded the position of the crank, the absolute angle of the pedal with respect to horizontal, and the components of the applied force oriented tangential and normal to the pedal surface. This pedal allowed the subject to use regular racing cleats and was mounted on the right side of an ergometer that was modified to simulate the subject's bike. Force records were collected (at 90 rpms) for 14 elite male cyclists (USCF category 1 and 2) at 4-6 workloads (single leg power outputs ranging from 92-280 W). Oxygen consumption was measured during the testing, and workload was calculated as a percent of VO_2 max. At each workload, an average force record was calculated for the subject from 5 consecutive crank revolutions (data filtered at 30 Hz with a second order lowpass Butterworth filter). Means were calculated at the lowest workload (range=56-68% of VO_2 max) and the highest workload that was not above their VO_2 max (range=83-100%). Individual responses to the increased workload were also analyzed.

Results: The results are discussed in response to the above hypotheses. (a) Comparing the means at the two workloads showed the subjects increased the pedal angle during the downstroke (Fig. 1). However, individual analysis of all workloads revealed that seven subjects showed no apparent tendency to change the pedal angle as workload increased, while only three subjects exhibited an obvious tendency. (b) The mean also showed ineffective downward normal forces during the upstroke at the low workload. At the high workload the normal force was mostly upward, but it did not produce significant positive torque about the crank (Fig. 2). Inspecting the individual records revealed cyclists who produced significant positive torque during the upstroke at the higher workloads also did so at the lower workloads. Therefore, generating significant positive torque during the upstroke to lower the demands during the downstroke was not a strategy adopted specifically at the higher workloads. (c) Horizontal and vertical components of the applied force were calculated with respect to a reference frame fixed to the bike (to negate the effect of different pedal orientations) and revealed the primary response to increased workload was to increase the vertical force. Analysis of the horizontal force at the high workload showed the subjects increased a forward directed force during most of the downstroke without changing a backwards directed force during the upstroke. Also, the peak value of the torque about the crank occurred slightly earlier at the high workload (fig. 2). The cyclists adapted to increasing workload predominantly by increasing force production (horizontal and vertical components) during the downstroke with no tendency to spread the force over a greater range of crank motion. At the high workload the subjects produced 96% of the total propulsive torque during the downstroke. Examination of individual response as workload increased revealed that six subjects kept the horizontal force relatively constant. Interestingly, these

same subjects were in the group that showed no tendency to change pedal orientation with workload.

Conclusion: Elite subjects showed two main strategies for adapting to increasing workload. Some subjects changed their pedal technique during the downstroke as a way to increase the horizontal component of the force as well as the vertical, while others tended to increase only the vertical component. At the higher workload, the subjects did not use a strategy of significantly increasing the production of propulsive torque during the upstroke.

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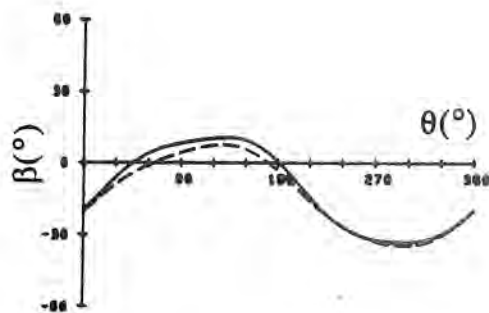


Figure 1. Plot of pedal angle (β) vs. crank angle (θ) for high (solid) and low (dashed) workload.

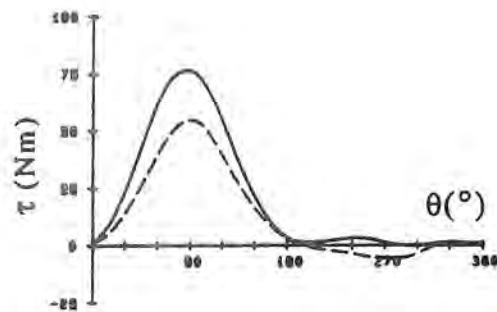


Figure 2. Plot of torque about the center of the crank (τ) vs. crank angle (θ) for high (solid) and low (dashed) workload.

TIME DELAY(EMD) BETWEEN THE EVOKED ELECTROMYOGRAM(EMG) AND THE EVOKED TORQUE

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INTRODUCTION

Komi, p.V. and others have reported about the time delay of EMD(Electric Mechanical Delay), which they defined as the difference time between the electric signal and the mechanical output wave form. EMD were measured as long as 40-60msec in the case of the movement of multi-joint, (total body movement). Grabiner, M.D. also reported the time delay between the surface EMG and the change of the joint angle as about 48msec on the total body movement. Authors have studied the evoked EMG with the electric stimulus, and have reported about the EMD between it and the torque. The value of EMD was only about 10 msec, and found 4 to 6 times difference between the total body movement and the single joint movement. It seems that the difference of EMD time value between former studies relies much on the kind of the mechanical output and sensors, and also influenced by solution and frequency response difference of EMG and output. For this reason, we intended to study the reason of the long or short change of EMD and difference between healthy subjects and patients by the accurate measurement and the experimental condition. The purpose of this study is also to clarify the dynamic mechanism of EMD.

METHOD

Fig.1 shows the diagram of the experimental system, and H-wave, M-wave, Supramaximal M-wave were recorded by the current stimulus. We used stimulus strength of 7-8mA in H-wave, 18-22mA in M-wave, 24-26mA in Supramaximal M-wave, and gave the stimulus by taking more than 1 minute of time interval. In addition to these, the voluntary muscular contraction were performed 5 times taking 3 minutes of time. The percutaneous current stimulus were given to the tibial nerve at the fossa poplitea. And then, contraction of the triceps surae and plantar flexion movement of the ankle joint were generated at the same time. The evoked EMG was generated from the soleus, the torque with the plantar flexion movement was measured by the Piezo-type torque sensor, which was attached to the back of the planta. In order to be sure accurate posture, the electrogoniometer was attached outside of the ankle joint. And then, the mechanical output was generated with the evoked EMG.

Subjects were tested about four healthy young men and six Myotonic Dystrophy patients, and we obtained their approval of applying the electric stimulus experiments. Three experimental conditions are contained as follows: (1)At rest (control experiment), (2)At three intensities of the electric current stimulus (H, M, and Supramaximal M waves), (3)At two kinds of strong or weak intensities at voluntary plantar flexion movement. By using these experimental conditions, the meaning of EMD and the muscular capacity and mechanism of mechanical output were discussed.

RESULT AND DISCUSSION

In the case of EMD measurement by the single stimulus at rest, averaged EMD values were recorded respectively about four healthy subjects 11.4(84)msec with H-wave,

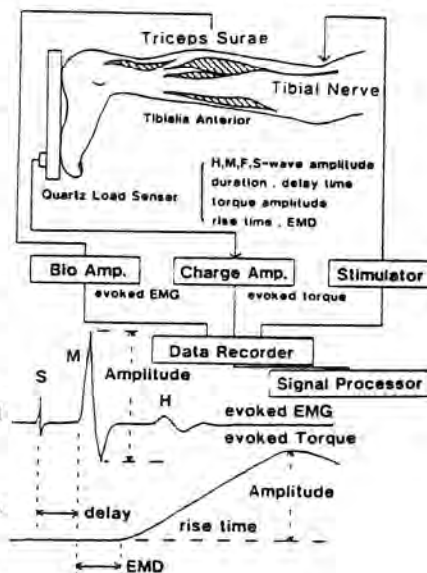


Fig.1 Experimental system of EMD

9.3(85)msec with M-wave, 8.6(80.5)msec with Supramaximal M-wave stimulus (the figure within the parenthesis shows the data for patients). Those values are shown in Fig.2 and 3. In the case of healthy subject, the delay time between H-wave and torque was shown with standard error of 6msec(18msec in patients). In the case of H-wave stimulus, because of facilitation in the spinal cord, the delay time is influenced shorter at all sorts of experimental conditions and body conditions. Then, EMD of H-wave changed at longer time. In the case of M-wave of Fig.3, the difference of the stimulus intensities influenced on EMD values. At increasing amplitude of 7mV, 7.5mV, and 8.0mV of electromyogram, there were occurred to be much differences at EMD values, and were observed a linear relation between the EMG amplitude and EMD, proportionally. During voluntary contraction of healthy, EMD is 2.1 times as long as that during the electric stimulus. In the case of electric stimulus, we could stimulate only triceps surae and only single charge of the motor unit. While, in the case of voluntary contraction, it is difficult to contract only the triceps surae and a single joint of the ankle. The nerve control of only one muscle is nonexistence in the joint movement. The motor unit of EMG at voluntary contractions is not special at voluntary excitation exactly. Changes of EMD might depend on the rising time of mechanical movement and muscular conditions. And appearance of torque are controlled by intensities of the muscular contraction or velocity of the muscular sliding movement. The sliding velocity of the muscle filament should depend on intensities of the muscular contraction, and perhaps it might be influenced by stimulus intensities and the dystrophy of the muscle filament.

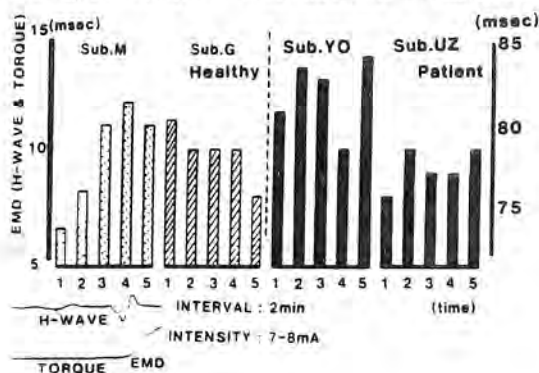


Fig.2 EMD at the H-wave stimulus among healthy(left) and patient(right)

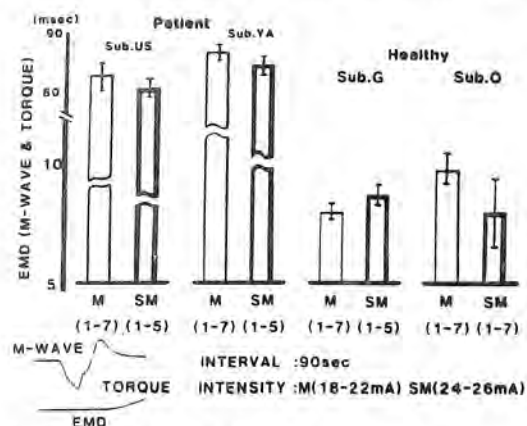


Fig.3 EMD at the M-wave stimulus among healthy(right) and patient(left)

SUMMARY

By adding percutaneous electric stimulation to the tibial nerve innervated on the triceps surae, the plantar flexion torque and the evoked EMG signals were recorded at the same time. Then, the time difference between the EMG and the torque were measured about healthy and patients quantitatively. This EMD was 11.4(84)msec with H-wave, 9.3(85)msec with M-wave and 8.6(80.5)msec with Supramaximal M-wave (parenthetic figures are about patients). And in the case of voluntary contraction, 28.5msec with weak tension, and 24.3 msec with strong tension. These EMD depended on the amplitude of evoked EMG, and it should be influenced to be shortened by various conditions of the joint, the motor units, and muscle filaments.

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STEP LENGTH, FREQUENCY, VELOCITY AND POWER RELATIONSHIPS IN RUNNING AND JUMPING

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INTRODUCTION. In human cyclic movement general tendency to increase velocity is to increase both step frequency and step length (e.g. Hoshikawa et al. 1973). The change in stepping pattern results from the angle action of the leg due to the muscular activity and force (e.g. Miyashita et al. 1971). The range of movement and angular velocity in cyclic motion increases with speed to some limit (e.g. Luhtanen and Komi 1978). The purpose of this study was to explain two major problems as follows:

1. the step length, frequency and velocity relationships in long jump, triple jump, 40 %, 60 %, 80 % submaximal, maximal, running, 3 step, 10 step without run up and tapping and
2. step length, frequency, velocity, force and power relationships in submaximal, maximal and supramaximal running.

METHODS. Subjects were thirty seven male track and field athletes representing sprint running ($n = 24$, $m = 76.3 \pm 6.9$ kg), jumping ($n = 11$, $m = 77.7 \pm 6.7$ kg) and decathlon ($n = 2$, $m = 79.0 \pm 4.2$ kg) events. For problem number one twenty four subjects ran, stepped and jumped on tartan type track in an indoor hall. Triple jumps ($n = 7$, $m = 73.7 \pm 4.3$ kg) were performed on a similar outdoor track. The performances were recorded as follows: long jump and running performances were filmed with a Locam 51 0003 camera at frame rate 100 Hz. Sanyo video camera, recorder and timer system (50 frames/sec) was used to record 10 step, 3 step and 30 consecutive tapping performances. A Bolex H 16 Reflex camera with frame frequency of 48 Hz was utilized in triple jump. For problem number two ($n = 11$, $m = 76.2 \pm 6.9$ kg) also the ground reaction forces were measured in running with 2 D force platform. Kinematic film analyses was employed to calculate the velocity of center of gravity, step length and frequency using film and video material. Average ground reaction forces were analyzed in the pushing phase in horizontal and vertical directions. The relationships between running velocity (v), step length (l) and step frequency (f) were applied using the formula

$$v = l \times f \quad (1)$$

The relationships between power (P), step characteristics and horizontal ground reaction forces (F) were analyzed using formula

$$P = F \times v = F \times l \times f. \quad (2)$$

RESULTS AND DISCUSSION. The step length and frequency relationships in long jump, triple jump, 3 and 10 step jumps without run up, maximal, submaximal running and tapping are shown in Figure 1. For the problem two in supramaximal running with a rubber rope horizontal towing system the running speed was

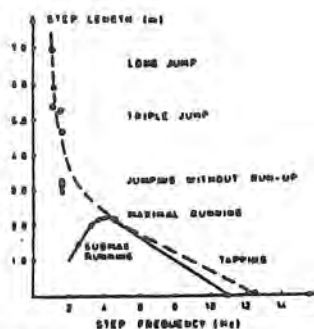


Figure 1. Step length and frequency characteristics running, jumping and tapping.

$10.72 \pm 0.37 \text{ ms}^{-1}$, step length $2.21 \pm 0.14 \text{ m}$, step frequency $4.87 \pm 0.24 \text{ Hz}$ and average horizontal pushing force $358 \pm 67 \text{ N}$. In maximal running the average maximal velocity, step length, frequency and horizontal pushing force were $9.95 \pm 0.37 \text{ ms}^{-1}$, $2.19 \pm 0.14 \text{ m}$, $4.55 \pm 0.21 \text{ Hz}$ and $381 \pm 77 \text{ N}$, respectively. At a submaximal speed of $5.08 \pm 0.41 \text{ ms}^{-1}$ was the step length $1.66 \pm 0.14 \text{ m}$, step frequency $3.07 \pm 0.11 \text{ Hz}$ and horizontal pushing force $221 \pm 84 \text{ N}$.

For the problem number one it was found that in sprinters the step length used in maximal sprinting was on average $38 \pm 7 \%$ related to the length in their long jump. The sprinters used in maximal running step frequency which was also $38 \pm 7 \%$ of the maximal tapping frequency. The step frequencies in 3 and 10 step were on average 38% of the frequency in maximal sprinting. When the sprinters were jumping 10 steps they used 150% of the step length and 57% of the step frequency as compared to the step length and frequency in maximal sprinting. For the problem number two the power evaluation showed that according to the formula 2 the power output remained constant (50 Wkg^{-1}) in maximal and supramaximal running although step length and frequency increased in supramaximal running. When the running speed decreased from the maximal value to 51% the power output decreased to 29% , step length to 76% , step frequency to 67% and horizontal pushing force to 58% .

In conclusion, individual step length frequency curve could be applied for the training process of sprinters knowing that the increment of running velocity at submaximal velocities comes with the longer step length and at maximal and supramaximal velocities mainly through step frequency. If the individual maximal step length, frequency and horizontal pushing forces and the actual values at the maximal running speed are known in testing conditions then an evaluation of resources in step length, frequency and pushing force could be done for building up the training program of sprinters.

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INTERACTION BETWEEN SWIMMING TECHNIQUE AND PERFORMANCE CAPACITY IN SWIMMING.

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INTRODUCTION. The efficiency of swimming technique can be best described by examining the behavior of the relationships between mean velocity (V), stroke rate (SR), and stroke length (SL) during swimming (East, 1970; Craig & Pendergast, 1979). The swimmers representing different fitness profiles have been noted to differ not only in competitive performance but also in the effectiveness of stroking, i.e. the best swimmers perform with longest SL (East, 1970; Craig & Pendergast, 1979; Craig & al., 1985). Costill & al. (1985) concluded that the product of SL and V (Stroke Index; SI) could be used to describe the efficiency of stroke technique and the best performers would thus demonstrate the highest SI. It was the aim of this study, therefore, to examine the interaction between stroke technique and performance capacity during a fitness test in free style swimming.

METHODS. Three groups of male competitive swimmers (15 swimmers in each group) served as subjects. The mean values for age were 19.4 ± 2.9 , 17.4 ± 1.9 and 15.7 ± 1.3 years, for height 183.5 ± 5.6 , 177.7 ± 5.5 and 177.4 ± 5.1 cm, and for weight 72.7 ± 6.2 , 67.5 ± 5.3 and 69.8 ± 9.1 kg in the groups I, II and III, respectively. They were selected according to their attendance in physical conditioning tests during the course of regular training. They represented different fitness categories so that the average time for 100 m free style swimming was 55.03 ± 1.3 s in group I, 57.72 ± 1.43 s in group II, and 60.49 ± 1.44 s in group III. The subjects performed an incremental type of n*100 m swim test starting with a very low speed and increasing the speed step by step until exhaustion. V, SR, SL and SI were measured to describe the stroke technique during the swims. Blood samples were drawn after each of the swims to analyze the blood lactate concentration (BLa). The conventional statistics consisted of means and standard deviations for the parameters. T-statistics with paired observations was used to test the statistical difference between the studied groups.

RESULTS. The examination of V demonstrated that group I obtained the highest and group III the lowest values in all intensities ($p < .001$). On the other hand BLa was lowest in group I and highest in group III swimmers at submaximum V levels while no difference was found in the maximum of BLa. Both SL and SI showed statistically different ($p < .001$) values between the groups. The maximum of SL were 2.557 ± 0.179 , 2.426 ± 0.254 and 2.293 ± 0.128 m in groups I, II and III, respectively. When the point of maximum SL was reached, BLa started to increase above it's initial level. However, SI reached maximum in V which corresponded to highly different BLa levels (2-11 mmol)

group I swimmers demonstrating the highest SI values (3.695 ± 0.315 , 3.339 ± 0.286 and 3.134 ± 0.336). In SR comparisons the best swimmers could perform with highest V at certain SR levels as compared with their inferior counterparts.

DISCUSSION. The present BLA vs V relationships demonstrate the well known effects of endurance training on physical fitness. On the other hand the present results showed that the physiological responses could be estimated by examining the values of SL and SI. One might thus suggest that these variables could be applied to describe the effectiveness of swimming technique during the course of the present type of physiological test (Keskinen & Komi, 1989). Furthermore, the finding that V at maximum SL is closely the same as the so-called anaerobic threshold according to the BLA vs V curves (see also Weiss & al., 1988 and Keskinen & al., 1989) suggest that during an incremental loading the development of lactic acidosis leads to parallel development of local muscle fatigue as shown by the decrease in SL. While the observed changes in SL were similar in all of the examined groups, they can be considered as positive effects of endurance training on performance in swimmers. The finding that the V of maximum SI corresponded to highly different BLA levels might lead one to suggest that the different specialization of the swimmers between sprint and distance events might be demonstrated by SI concept.

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WAVE ANALYSIS ON SURFACE OSCILLATION SIGNALS FROM TWITCH CONTRACTION.

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PURPOSE

The dynamic mechanics of the muscle consists of three components and is shown in Fig.1 : elasticity, viscosity and contractility, and which perform the base function of the muscle contractive pattern (muscle length, muscle tension, and contractive velocity). It is difficult to observe the dynamic components non-invasively from outside of the skin surface. However the identification of the muscle contractive components has become possible by analyzing complex oscillation waves generated on the surface of the skin. The purpose of this study was to find the elastic, viscous and contractile components by the measure of the decrease rate of the wave form and of the frequency of oscillation, and by the amplitude of evoked EMG.

METHOD

Subjects were five healthy men, and five patients of MD disease (Myotonic Dystrophy). They kept walking and the triceps surae showed atrophy-involuntional phenomena. H-wave was evoked with 4-6mA, M-wave with 18-22mA, and supramaximal M-wave with over 24mA by adding an electric current stimulus percutaneously at the knee fossa (poplitea). Furthermore, muscle contraction were practiced with voluntary condition. The surface oscillation which is transmitted to the skin surface of the muscle was measured with the Piezo-electric-sensor (manufactured by TEAC). Also, the planter flexible force was recorded with Washer-type tension sensor (manufactured by Kistler). Under of contractions, the oscillation waves were superimposing averaged at 10 times according to each stimulus condition, and the frequency analysis was calculated by the measure of the decrease rate and FFT (Fast Fourier Transform) used with analyzer of the Nikolei-4094 type computer. The place in which sensor was installed are shown in Fig.2.

○ Decrease rate

$$\tan \delta = \Delta / \pi$$

$$\Delta = nT$$

$$n = 1/T \log(a_0 / a)$$

a : amplitude of the first wave

a₀ : amplitude of the last wave

T : frequency of the last wave

TT : the decrease time of wave form

○ Fourier Transform

$$F(f) = \int_{-\infty}^{+\infty} \chi(t) e^{-2\pi f t} dt$$

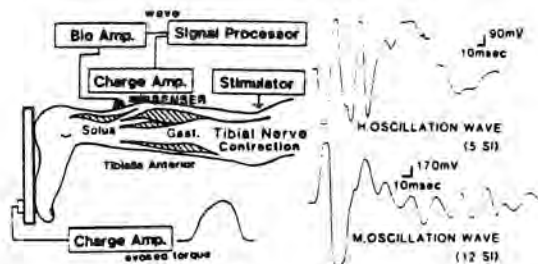


Fig.2 Experimental chart and a example of oscillation waves.

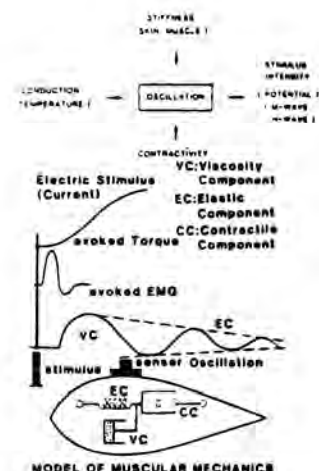


Fig.1 Model of muscle mechanics and oscillation sources.

RESULTS AND DISCUSSION

Surface oscillation changed remarkably in proportion to the amplitude of evoked EMG and the strength of the electric stimulus, and after about 10msec on appearance of evoked EMG, evoked torque was generated. The amplitude of torque was also generated in proportion to the intensity of the muscle contractive potential. Surface skin oscillation waves also changed in proportional to stiffness of the muscle. In healthy men, there was observed a longer oscillation wave that has damping time of 1,020 msec. But averaged damping time of healthy men was 332.8msec. Sustained oscillation was recorded to be notable in the case of MD patients, and their average time was 2,106.0msec. Because their central nerve control should be reduced to relaxation from the stress, and the antagonistic muscle contraction and inhibitory nerve control had remarkable inferior in the case of patients to healthy men. As a result of analysis, the oscillation waves were patterned a look like sine-wave during twitch contraction. Then, the contractive waves were estimated with the elastic component, viscous component and their contractility within the muscle physically. The function level of the muscle had influenced on these oscillation waves, and the mean of the decrease rate which are shown at Fig.3, were calculated as follows: 0.20 (H-wave), 0.14 (M-wave), 0.08 (SM-wave), and 0.05 (voluntary) in healthy men, and 0.09 (H-wave), 0.27 (M-wave), 0.05 (SM-wave), and 0.19 (voluntary) in patients. Those figures showed a quite remarkable difference among subjects of healthy and patients. Then, these shows the function difference of the elastic component in the muscle, and the sliding velocity of muscle filaments clearly occurred as the contractive oscillation of the surface. A result of FFT analysis about the oscillation wave, was shown in Fig.4. The PPF1 (the first peak frequency) appeared at 12Hz in healthy men, and the PPF2 (the second peak frequency) was obtained at the vicinity of 24Hz with high power (density). While in patients, only the PPF2 was shown the lower frequency, at that a tendency of muscle dystrophy would be assumed to be inferior responses of function. This would show the function reduction of the viscous component in the muscle.

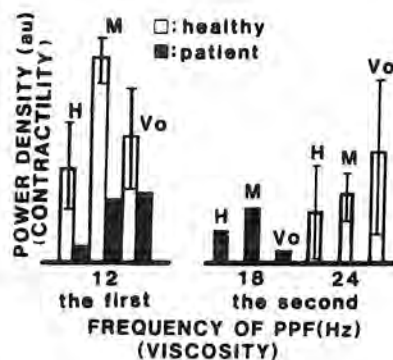


Fig.4 Comparison between healthy and patient with a result of FFT analysis about the oscillation waves.

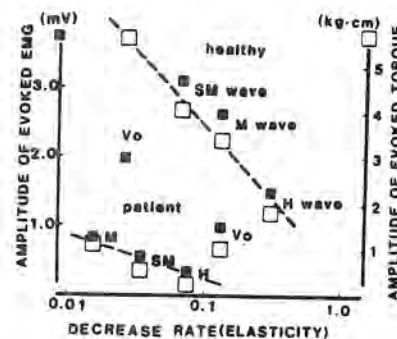


Fig.3 Comparison between healthy and patient with the decrease rate of oscillation waves on the surface skin.

SUMMARY

In wave analysis on surface oscillation signals from twitch contraction, the change of the muscular mechanics and its function difference of their components have been possible to be estimated with the decrease rate and FFT, non-invasively.

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MATHEMATICAL MODEL AND ANALYSIS OF PATELLAR DISLOCATION

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INTRODUCTION

Recurrent dislocation of the patella generally produces high stresses on the patellofemoral joint surfaces, and frequently causes severe degenerative changes in their articular cartilages. In the literature, attention has been directed mainly to the surgical treatment but its multifactorial etiology has not been fully discussed. Regarding the pathomechanics of patellar dislocation, few authors have described them as associated with the joint surface configurations, and the extensor mechanism (1,2). As for the joint surface configurations, Brattstrom (3) has stated that a clear connection exists between the size of the lateral femoral condyle and the occurrence of the dislocation of the patella. However, the relationship between the extensor mechanism and the patellar dislocation remains unclear.

In the present study, influences of insertion and strength of each extensor element on the patellar dislocation were investigated through our 2-dimensional mathematical model analysis in combination with experimental design theory.

2-DIMENSIONAL HORIZONTAL MODEL OF THE KNEE JOINT

Our simplified 2-dimensional model is given in Fig. 1, which describes horizontal sections of the knee joint in 30 degrees of flexion. The elements F1 (lateral fascia), F4 (patellar tendon) and F6 (medial fascia) are symbolized by three springs. The elements F2 (vastus lateralis), F3 (vastus intermedius) and F5 (vastus medialis) are present with three strings pulled by the force of F2, F3 and F5, respectively. The curvature sections of the patella and the femur are expressed by the spline* functions using X-ray photographic data. From both the geometric and kinematic equilibrium conditions, we may compose 7 equations. Then, solving these equations, we may introduce 7 variables; position (x, y) and attitude (θ) of the patella with respect to the femur, contact positions (X1, X2) and forces (P1, P2) of lateral and medial sides.

Among the above, we determined the absolute value of the lateral contact position (X1) as the index of the patellar dislocation. We also paid attention to the contact forces (P1, P2) from the standpoint of articular damage.

SIMULATIONAL ANALYSIS USING THE EXPERIMENTAL DESIGN THEORY

Simulation about the normal knee was performed. The values of spring coefficients of fascias and patellar tendon were quoted from the literature (4, 5). The tension ratio among three muscles was set as: F2/F3/F5:2/3/2.5(6). As for the dimensional values, actually measured data were used for input data. Then, a series of simulations were repeated under the parametric alteration of input data with respect to

tension ratio of muscles, spring coefficients, inserting locations. Three parametric alterations of every factor was tried, namely, force and spring coefficient; null (rupture), normal value, double of normal value. Inserting location; normal location, right and left shift by 3 mm (patellar side), 5 mm (femur side). The L3(27) orthogonal matrix from the Theory of Experimental Design (7) was adopted to our simulation and the method of variance analysis was employed for data analysis.

CONCLUSION

In the present study, we performed mathematical simulation to identify the influences of insertion and strength of each extensor element on patellar dislocation using a 2-dimensional model, and obtained the following results:

1. Regardless of the inserting location in the patellar and the femoral sides, the insertions of the patellar tendon has strong influence on patella dislocation and for lateral and medial contact forces.
2. Tensile force of the vastus intermedialis strongly influences the patella dislocation and lateral contact force.
3. Medial contact force is influenced mainly by tensile force of the vastus medialis.
4. Fascia rupture scarcely makes lateral stability of the patella fall down.

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* Function to quantify unusual, free curvatur.

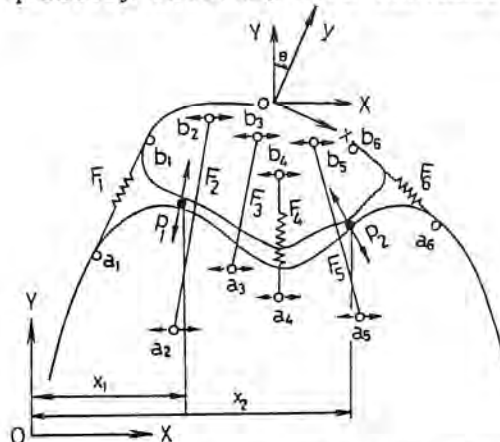


Fig. 1 Top view of our 2-dimensional knee model at 30 degrees flexion.

IMPACT CHARACTERISTICS OF SOCCER BALLS VARYING IN SIZE AND DESIGN

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Numerous authors have discussed injuries in soccer that result from the impact of the ball against the head (2, 5, 6). In an attempt to accommodate players of all ages, as well as the indoor version of soccer, manufacturers produce balls that vary considerably in size and design. It has been speculated that such variations may influence factors that determine injury. Previous research has provided some insight into this by establishing a theoretical model that illustrates the effect of ball impact on head accelerations (4), by documenting the magnitudes of ball impact, and by examining the effect of factors such as moisture and inflation pressure (1, 3). However, little is known about the specific influence of size and design factors on the characteristics of impact. Thus, the purpose of this study was to examine selected impact characteristics of soccer balls that varied in both size and design.

METHODS

Fifty-eight new soccer balls, representing three standard sizes and two unique designs (indoor and headlite) were selected for testing. Ball impact was assessed by repeatedly dropping each ball onto a force platform such that the impact velocity was approximately 18 m/sec. All balls were tested dry and at the same inflation pressure and temperature. Data were recorded initially on FM tape, and later A/D converted at 6 khz. The data were further reduced to provide measures of four characteristics of impact: peak impact force (PF), adjusted rise time to peak force (RT), total impulse (TI), and impact duration (ID).

RESULTS/DISCUSSION

A one-way ANOVA for each variable, followed by Scheffe' post-hoc tests, provided evidence of numerous significant differences between the five categories of balls (Table 1). As anticipated, increased ball mass was associated with significant increases in PF. The mean PF of the size #5 balls exceeded that of the #4 balls by 26%, the #3 balls by 35%, and the headlite balls by 47%. PF influences the degree of both tissue compression and head acceleration that occurs during impact. The data indicate that decreases in ball size may substantially reduce the amount of loading that takes place, thus decreasing the likelihood of injury. This may be partially offset, however, by the finding that the larger balls tended to have longer RT values (adjusted for peak force). RT reflects the potential rate of loading and, in combination with peak force, influences the probability of injury.

Relative to the specific influence of impact on head acceleration, as ball size decreased TI decreased. This occurred in spite of the fact that the ID values were greater for the smaller balls. The lowest TI values were associated with the headlite balls, and were 47% below that of the size #5 balls. This indicates that variability in ball size/type significantly influences the potential of the ball for inducing those injuries associated with impact loading. In comparing the size #5 outdoor balls with the indoor balls, the differences in both PF and TI can best be attributed to corresponding differences in mass. However, the mean RT of the indoor balls was greater than that of the outdoor balls. This may be a function of enhanced compliance of the external surface of the indoor ball. Collectively, the results indicate that ball size and design substantially influence important characteristics of impact. The data also provide support for the position expressed by others that ball impact may produce the potential for head injury. Further, as illustrated by the response of the headlite ball, it appears to be possible to produce a ball that substantially modifies those factors most likely to be associated with injury.

Table 1 - Ball Characteristics

Ball Size/Type	#5	#4	#3	Headlite #4	Indoor #5
Sample Size (n)	32	8	9	3	6
Mean Mass (grams)	425	353	314	237	422
Mean Peak Force (N)	778.45	584.75	508.67	414.67	702.67
Mean Rise Time (ms)	4.16	3.70	3.36	3.33	4.24
Mean Impulse (N.s)	10.33	9.01	7.89	5.39	10.04
Mean Duration (ms)	24.59	27.90	27.56	27.32	25.00

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COMPARATIVE ANALYSIS OF DECATHLETE AND ELITE JAVELIN THROWERS.
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The primary purpose of this study was to compare selected kinematic parameters of decathlete javelin throwers to elite javelin throwers. A secondary purpose was to provide information pertaining to the relationships between throwing distance and the mechanical variables selected for analyzing decathlete javelin performance. Sixteen decathletes competing in the 1988 U.S. Olympic Track and Field Trials were videotaped with a Panasonic AG-160 camcorder positioned perpendicular to the javelin runway. The high-speed (1/1000) shutter function was used to minimize blur of the video image. A two-dimensional analysis of the best performance for each decathlete was completed using a PEAK Video/Computer Motion Measurement system. The data were then compared with corresponding values of elite male (N=6) and female (N=8) javelin throwers (reported by Whiting, Gregor, and Halushka) collected at the 1987 TAC National Championship.

The parameters selected for comparison were distance of the throw; release velocity; attitude, release, and attack angles; and the length of the last step. The attitude angle was defined as the angle of the javelin with respect to the right horizontal. The angle of release was the angle of the velocity vector of the javelin with respect to the right horizontal. The angle of attack was defined as the difference between the attitude angle and the angle of release. The length of the last step was measured from the toe of the back foot to the toe of the forward foot. Means and standard deviations for the selected kinematic parameters are as follows for decathletes, elite males, and elite females:

<u>PARAMETER</u>	<u>DECATHLETES</u>	<u>ELITE MALES</u>	<u>ELITE FEMALES</u>
Distance (m)	60.95 ± 5.09	78.02 ± 3.65	56.84 ± 3.31
Velocity at Release (m/s)	24.23 ± 4.19	29.70 ± 1.52	24.73 ± 2.02
Attitude Angle (deg)	40.85 ± 5.20	36.35 ± 4.88	41.55 ± 3.01
Release Angle (deg)	35.20 ± 2.80	36.48 ± 2.03	37.85 ± 2.57
Attack Angle (deg)	5.64 ± 4.55	-0.15 ± 4.53	3.70 ± 2.90
Last Step Length (m)	1.68 ± 0.20	1.71 ± 0.12	1.39 ± 0.12

In comparison, the decathletes generated lower release velocities than both the elite males and females. The release angle was also less for decathletes than the elite throwers. Greater attack angles and step lengths were found for the decathletes than for either the male or female elites. Attitude angles of the decathlete throws were less than those of the elite female javelin throwers, but greater than those of the male elite throwers. Comparison of throw distances between male and female javelin throwers was not considered to be appropriate because of the difference in the properties of the javelins. The elite males threw an average of approximately 17 meters farther than the decathlete throwers.

The correlation analysis was performed for informational purposes only. Since no statistical inference was intended, only correlation coefficients are provided. The relationships between distance of the throw with the mechanical variables selected for analysis are provided in the following table:

<u>PARAMETER</u>	<u>CORRELATION</u>
Velocity of release	.47
Attitude angle	-.20
Release angle	-.03
Attack angle	-.21
Last step length	.05
Height at release	.23
Distance behind foul line	.64

Of particular interest is the highest relationship found between distance of the throw with distance of the thrower behind the foul line. The practical significance of this correlation is somewhat lessened from an inspection of the magnitudes of the raw data for these two variables. For the decathletes with the three longest throws (including the meet record), they also exhibited the three greatest distances behind the foul line.

Regression Equations for Estimating the Length of Six Biarticular Muscle-Tendon Complexes of the Lower Extremity as a Function of Hip & Knee Flexion Angles D. Hawkins, Department of Biomedical Engineering and M.L. Hull, Department of Mechanical Engineering, University of California Davis, CA 95616

INTRODUCTION

When studying human movement, it is often of interest to determine the relative contribution that specific muscles make to generate a given motion. An active single joint muscle will lengthen or shorten, and hence do negative or positive work, in direct relation to the movement of the joint that it crosses. The situation for two joint muscles is more complex since muscle length depends on the position of both joints. The objective of this study was to develop a method for quickly estimating muscle-tendon lengths for six biarticular muscles crossing the knee and hip joints based on kinematic information and simple anthropometric parameters.

METHODS

A computer simulation approach was used to examine the relation between hip flexion angle, knee flexion angle, and muscle-tendon lengths. Lower extremity segment lengths measured from six male subjects were utilized as input data to computer software which simulated various joint center locations and hence various hip and knee angles. Joint center locations were simulated to occur in the sagittal plane with muscle origin-insertions located in 3-dimensional space. For each simulated body position, muscle attachment sites were located in local coordinate reference frames and then transformed into a global inertial reference frame. Non-dimensional muscle-tendon origin-insertion coordinate locations for the six biarticular muscles examined (biceps femoris, rectus femoris, semimembranosus, semitendinosus, sartorius, and gracilis) were obtained from Brand et.al. (1982). Anthropometric measurements obtained from the six subjects were used to scale the muscle origin-insertion data. The range of angles simulated was 60 to 180 degrees in 15 degree increments for both the hip and knee angles (180 degrees represents full extension). Three of the muscles examined (biceps femoris, sartorius, and semitendinosus) wrap around bony structures as they cross the knee and hence do not act in a straight line from origin to insertion. For these muscles, "effective" insertion sites were located at the point where the muscles cross the bony structures. For each muscle the distance from origin to insertion was determined using a straight line approximation. The resulting muscle-tendon lengths for each subject were normalized with respect to the subject's thigh length. Normalized muscle length data were examined statistically using polynomial multiple regression analysis.

RESULTS & DISCUSSION

Contour plots for two of the six muscles examined, taken from one subject, are illustrated in Figs. 1 and 2. The biceps femoris contour plot (see Fig. 1) is representative of the other

hamstrings muscles (i.e. semitendinosus, semimembranosus). Lengths of the hamstrings muscles are more sensitive to changes in hip angle than knee angle. The hamstrings are longest during full hip flexion combined with full knee extension and shortest when the hip is fully extended and the knee is fully flexed. The rectus femoris contour plot (see Fig. 2) shows that this muscle is more sensitive to knee angle changes when the knee is extended and the hip is flexed, and more sensitive to hip angle changes when the hip is extended and the knee is flexed. It is equally sensitive to both angle changes otherwise. This muscle is longest and shortest for hip and knee angles giving the shortest and longest hamstrings lengths respectively.

Each normalized muscle-tendon length was expressed as a function of the hip and knee angles in the following way:

$$\text{Length} = C_0 + C_1 * \theta + C_2 * \beta + C_3 * \theta^2 + C_4 * \beta^2$$

where θ , β are hip and knee angles respectively and C_0 , C_1 , C_2 , C_3 , C_4 are regression coefficients. Regression coefficients and the coefficients of multiple correlation (R), for the six muscles examined are given in Table 1. Since all correlation coefficients are greater than 0.94, these regression equations may be used to accurately determine muscle-tendon lengths.

CONCLUSIONS

The above equations serve to confidently estimate the length of six biarticular muscles for a specific individual provided joint kinematics and subject thigh length are known. Muscle-tendon lengths may in turn be used in conjunction with EMG data to estimate the relative work produced by each of these muscles in performing a given task. Such information is vital if we are to understand the role that these two joint muscles play during normal and pathological movements.

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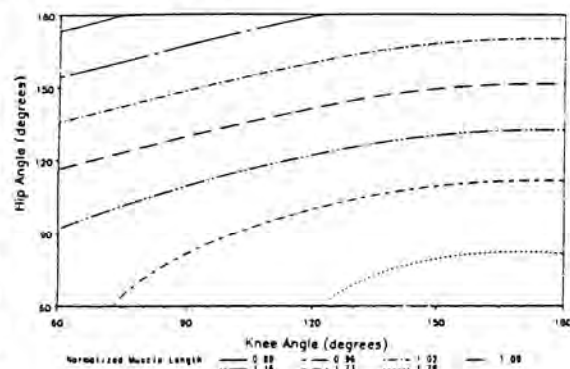


Fig. 1: Contour plot for the biceps femoris.

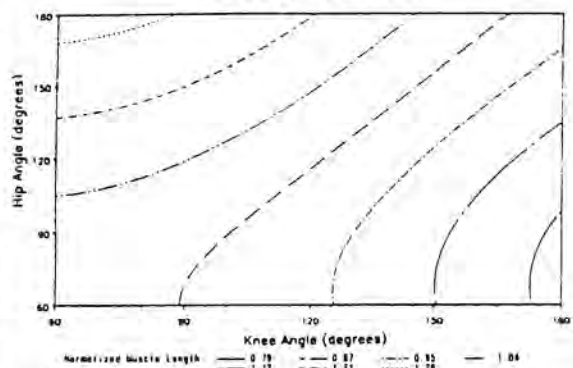


Fig. 2: Contour plot for the rectus femoris

Normalized length	C0	C1	C2	C3	C4	R
Biceps femoris	1.010	-7.50E-4	4.26E-3	-7.87E-6	-1.09E-5	0.98
Semimembranosus	0.968	-1.02E-3	3.15E-3	-6.34E-6	-3.84E-6	0.98
Semitendinosus	1.039	-3.18E-4	1.91E-3	-9.80E-6	-5.12E-7	0.97
Rectus femoris	0.946	2.48E-4	2.43E-3	7.04E-6	-1.95E-5	0.95
Sartorius	0.369	2.72E-3	2.82E-3	-9.25E-7	-6.29E-6	0.98
Gracilis	1.050	-6.23E-3	2.34E-3	2.29E-5	-1.50E-6	0.94

Table 1: Multiple regression coefficients and correlation.

A Study of Muscle Mechanics Through Modeling & Computer Simulation D. Hawkins, Department of Biomedical Engineering and M.L. Hull, Department of Mechanical Engineering, University of California Davis, CA 95616

INTRODUCTION

Human movement is the result of the body's response to a multitude of physical and volitive factors. To describe human movement requires an understanding of the various levels of the neuromuscular network including the central nervous system, the peripheral nerves, and the muscles. In this paper attention is directed to the muscles. In particular, a comprehensive muscle model is outlined and computer simulation results are presented.

Efforts to model the dynamic characteristics of muscle are not new. A number of muscle models have been derived with varying designs and complexities [Huxley, 1969, Zahalak et.al., 1976, Hatze, 1981, Woittiez et.al., 1984]. However, no one model gives a simple and complete formulation which brings together our present knowledge of muscle form and function. Many models include a few important factors while neglecting others of equal importance. A few models involve so many parameters that they are impractical to use. It is the object of the work described herein to bring our present knowledge of muscle structure and function together in a single descriptive formulation that can be used to predict muscle force in-vivo.

METHODS

A comprehensive model has been derived that describes the dynamics of isolated muscle tissue. This model is based at the cellular level with the muscle fibres being the fundamental unit. To complement this model, computer simulation software has been developed to study various factors related to muscle function (i.e. muscle compositional, neurological, and architectural factors). The model includes factors for fatigue and force enhancement as well, but these factors have not yet been incorporated into the simulation software. A brief overview of the muscle model derivation as utilized in the simulation software is given as Equation 1.

$$F_m = \sum_{i=1}^3 [n_{a_i} * F_{f_i}] * \cos(B) \quad (1)$$

where: F_m - resultant muscle force

i - fibre type identifier

= 1 for slow twitch oxidative (SO) fibres

= 2 for fast twitch oxidative glycolytic (FOG) fibres

= 3 for fast twitch glycolytic (FG) fibres

n_{a_i} - represents the number of activated ' i ' fibres

= function of(activation level, number of ' i ' fibres in muscle)

F_{f_i} - the force generated by an active ' i ' fibre

= function of(fibre type, muscle length, contraction velocity)

B - muscle fibres angle of pennation

RESULTS & DISCUSSION

Sample computer simulation results utilizing the above model are shown in Figs. 1 and 2. In Fig. 1, the effects of muscle architectural variations on muscle dynamics are illustrated. Figure 2 illustrates the effects that muscle composition has on force and velocity characteristics. Results from other simulations are not included due to space limitations. Figure 1 reveals that a muscle having longer fibres (i.e. more sarcomeres in series) has the ability to shorten faster than a similar muscle with shorter fibres. A muscle with a greater number of fibres in parallel (i.e. greater cross-sectional area) can produce more force than a muscle with fewer fibres in parallel, if equally activated. Figure 2 shows that a muscle consisting of predominately fast twitch fibres can generate greater shortening velocities per load than a muscle, of similar cross-sectional area, consisting of slow twitch fibres.

CONCLUSIONS

The results presented here demonstrate the usefulness of modeling and simulation techniques for studying muscle dynamics. This approach allows many different factors related to muscle function to be studied as well as their interactions. Such studies may give clues as to the site of neuromuscular disorders and reasons for outstanding physical capabilities.

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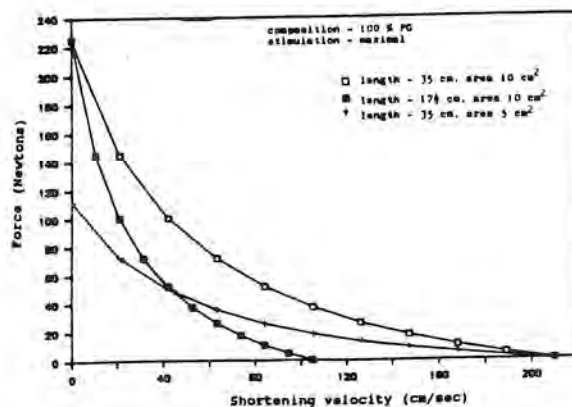


Fig. 1: Muscle architecture & contraction characteristics.

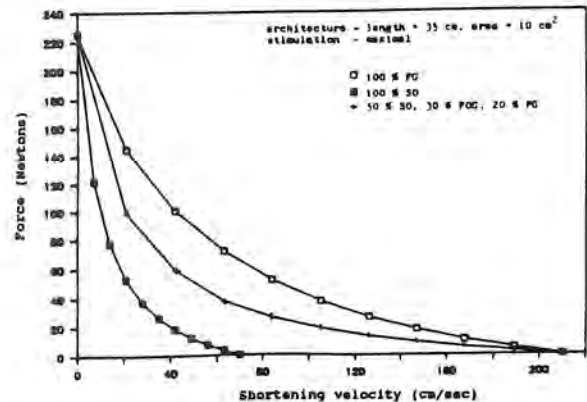


Fig. 2: Muscle composition & contraction characteristics.

**SPECTRAL ANALYSIS OF LOWER BODY
KINETICS DURING SELECTED ACTIVITIES**

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Eighteen normal male subjects with an average age, height, and weight of 27.3 years, 177.6 cm, and 756.6 N respectively were analyzed at 200 hz for lower body forces during four activities: Level walking, the first step up on a three step riser, the last step down on a three step riser, and during bouncing on their toes. A Kistler forceplate was used to determine the ground reaction forces (GRF) and center of pressure, while 3D joint coordinates were obtained via a Motion Analysis VP310 video processor connected to a Sun 3/260 mini computer which supported the Motion Analysis EV3D software package.

Utilizing the synchronized forceplate and joint center locations a program written by the investigator in Fortran 77 calculated the lower body bone forces parallel and normal to the long axis of the bone, (addition of the proximal and distal segment forces with respect to segment angle). The moment about the segment center of gravity, GRF, and compression and shear segment forces were calculated for one cycle, (i.e. for walking, a cycle was from right foot touching the forceplate until the right foot touched the ground again), then standardized for the subjects body weight (BW). The kinetic information for each step cycle was then discrete fourier analyzed (DFA) to determine the frequency content. The frequencies of the first harmonic varied slightly within each activity, but when rounded to the nearest hundredth, all values within a single activity were the same. Table 1 shows the rounded off frequency of the first harmonic and average standard deviation (SD) for each activity analyzed. Table 2 shows the order of the harmonics which contribute the majority of power to the spectrum for each activity and segment.

ACTIVITY	1ST HARMONIC (Hz)	MEAN SD
BOUNCE	1.90	0.485
WALK	0.81	0.045
UP STEPS	0.78	0.065
DOWN STEPS	0.75	0.064

Table 1
First Harmonic for each Activity

Except during bouncing, when the segments and ground kinetics were standardized to maximums of one and one for time and force, the plots of ground and segment kinetics for all eighteen subjects followed virtually the same patterns for the respective segment and activity. Walking and walking up steps

	WALKING	UP STEPS	DOWN STEPS	BOUNCING
Fz	1,3	1,3	1,3,4,2	1
Fy	1,2	1,2	1,2	1
Foot Compression	1,3	1,3	1,2,4,3	1
Foot Shear	1,3	1,3	1,2,4,3	1
Foot Moment	1,3	1,3	1,3	1
Shank Compression	1,3	1,3	1,3,2,4	1
Shank Shear	1,2,3	1,2,3	1,3,4	1
Shank Moment	1	1	1	1
Thigh Compression	1,3	1,3	1,3,4	1
Thigh Shear	1,2	1,2,3	1,2	1
Thigh Moment	Virtually nothing, even at 1st harmonic			

Table 2
Harmonics in Order of Contribution to Power of Spectrum

seemed to almost be mirror images of each other except for the hip shear force walking up steps which did not have as much positive shear. Also, even though the GRF traces were similar, the compression and shear forces found in the foot followed two distinct patterns with respect to the subject striking the forceplate heel, or toe first.

All activities showed the greatest forces in compression of the shank. Walking had a mean maximum value of 2.122 BW (SD=0.193) and occurred at 15% of the cycle. Walking up steps had a mean maximum of 2.341 BW (SD=0.226) at 16% of cycle. Walking down steps showed a mean maximum force at 87% of cycle equal to 2.946 BW (SD=0.407) and bouncing obtained its maximum force at two different percents of the cycle, 25% and 50%. The forces were 1.029 (SD=0.016) and 1.22 (SD=0.285) respectively.

Maximum forces and moments for other segments did not necessarily fall in the same percent cycle as that found for the maximum shank compression forces. Generally the maximum forces were around 15% or 50% for walking and walking up steps, 25% or 50% for bouncing, and 57% or 90% for walking down steps. In walking and walking up steps, 15% was about the time when the knee stopped flexion as the upper body moved over the the lower body. Fifty percent was just before toe off when the weight was all on the ball of the foot. During bouncing, 25% was on the way up and 50% was just at the top. All weight was on the ball of the foot in both cases. The cycle walking down steps was defined differently than that for walking and walking up steps. This is why the percents seem so different. Fifty-seven percent of the down steps cycle was just after the initial contact with the forceplate and 90% was during the initial pushoff phase. Ninety percent of cycle corresponds close to the 50% cycle when walking and walking up steps.

BIOMECHANICAL CHARACTERISTICS OF PROPULSIVE TORQUE
GENERATION IN TWO GROUPS OF ELITE CYCLISTS

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INTRODUCTION: The purpose of this project was to examine the biomechanical and physiological factors that affect the performance of elite cyclists during a 40 km time trial. Specifically, this paper focuses on examining the pedalling strategies used by elite cyclists to generate the propulsive torque about the center of the crank (used to increase or maintain the speed of the cyclist) during a pedal revolution.

SUBJECTS: The subjects were 14 male cyclists (United States Cycling Federation categories 1, 2 and 3) and they were divided into two groups based on their most recent time in a 40 km time trial: Group 1, $t=53.6 \text{ min} \pm 1.6 \text{ min}$; Group 2, $t= 58.7 \text{ min} \pm 1.5 \text{ min}$.

METHODS: A pedal dynamometer (1) and other instrumentation were attached to the right side of an ergometer that had been modified to simulate a racing bicycle. The following data were then collected (sampling rate = 550 Hz) for 10 consecutive pedal revolutions: (1) θ , the angle of the crank with respect to the vertical; (2) ψ , the orientation angle of the pedal; (3) F_T and F_N , the tangential and normal components of the force applied by the cyclist to the pedal, respectively. The force data were filtered, the five revolutions nearest 90 RPMs were averaged, and the following measures were then computed: (1) F_R , the net force applied to the pedal ($F_T + F_N$); (2) F_R , the magnitude of F_R ; (3) F_H and F_V , the horizontal and vertical components of F_R ; (4) T , the torque created by F_R about the center of the crank; (5) F_E , the magnitude of the component of F_R perpendicular to the crank arm and that also created a propulsive torque on the crank. An index of effectiveness (IE) was computed as the area under the F_E versus θ curve divided by the area under the F_R versus θ curve (see Figs. 2 & 3).

Testing was performed over two days. On Day 1, VO_2 and the data listed above were collected while the subjects cycled (at 90 RPMs) for five minutes at exercise levels that corresponded approximately to 50%, 60%, 70%, 80% and 90% of their VO_2MAX . On Day 2, the subjects rode a laboratory performance test that was designed to simulate a 40 km time trial (2). The object was for each cyclist to perform as much work as possible in one hour while riding at their preferred RPMs. VO_2 was also measured at selected intervals during the performance test, and the average VO_2 during the hour was determined. Comparisons between the groups were then made using the data collected from the pedal and ergometer on Day 1 at the exercise level that was nearest the average VO_2 of the cyclist during the performance test. The data were averaged across the subjects in each group and Student t tests were used to compare the groups on the various measures.

RESULTS: The results indicated that Group 1 had a higher absolute VO_2 during the performance ride (Group 1 = $4.59 \pm 0.25 \text{ l/min}$; Group 2 = $4.17 \pm 0.34 \text{ l/min}$; $p < 0.05$), and that the cyclists in Group 1 were able to maintain an higher average power output on the ergometer during the performance ride (Group 1 = $349 \text{ W} \pm 22 \text{ W}$; Group 2 = $313 \text{ W} \pm 28 \text{ W}$; $p < 0.05$). As all subjects pedalled near 90 RPMs ($90.1 \text{ RPMs} \pm 1.4 \text{ RPMs}$), the differences in the power outputs were evident in the values and the patterns of the T versus θ curves of the groups (Fig. 1). As shown in Figure 1, the cyclists in both groups created almost all of the propulsive (positive) torque during the downstroke (between 0° and 180°), with the peak propulsive torque (Group 1 = $77.6 \text{ Nm} \pm 7.7 \text{ Nm}$; Group 2 = $66.6 \text{ Nm} \pm 11.8 \text{ Nm}$; $p < 0.05$) occurring near 90° (Group 1 = $86.5^\circ \pm 8.4^\circ$; Group 2 = $85.5^\circ \pm 4.7^\circ$). Surprisingly, when compared over the entire pedal revolution the subjects of Group 2 were more effective than the cyclists in Group 1 (Group 1: $\text{IE} = 62.9\% \pm 4.7\%$; Group 2:

IE = $68.2\% \pm 7.4\%$; $p < 0.10$), but the groups demonstrated similar effectiveness when only the downstroke was examined (Group 1: IE = $70.8\% \pm 3.3\%$; Group 2 = $73.8\% \pm 5.5\%$; $p < 0.05$ -- see Figs. 2 & 3). When the plots of F_V were examined for each subject, it was apparent that the cyclists in Group 1 made large downward vertical forces on the pedal during the downstroke and negligible forces on the pedal during the upstroke. By comparison, the cyclists in Group 2 made smaller vertical forces on the pedal during the downstroke, but they did attempt to pull up on the pedal during the upstroke (i.e., create an upward vertical force on the pedal between 180° and 360°). The pattern of F_H was much more variable among the cyclists, but the subjects in Group 1 tended to apply larger forward directed forces to the pedal during the downstroke.

CONCLUSION: The cyclists who performed better during a 40 km time trial created significantly larger propulsive values for T during the downstroke, primarily by applying large downward vertical forces to the pedal at this time. Only the cyclists with the slower times in a 40 km time trial attempted to make upwardly directed vertical forces on the pedal during the upstroke. Therefore, it seems that the better performers do not attempt to create a relatively constant propulsive torque over the entire pedal revolution.

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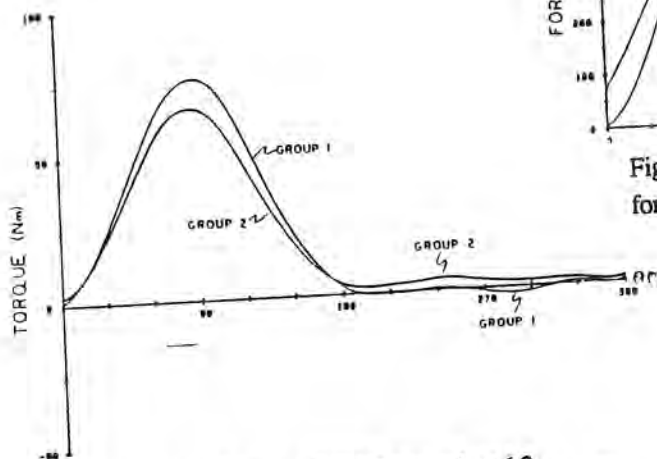


Figure 1. Plots of T versus θ for Groups 1 and 2. A positive torque was a propulsive torque. $\theta = 0^\circ$ or 360° when the crank arm was pointing vertically upward, and $\theta = 180^\circ$ when the crank was pointing vertically downward.

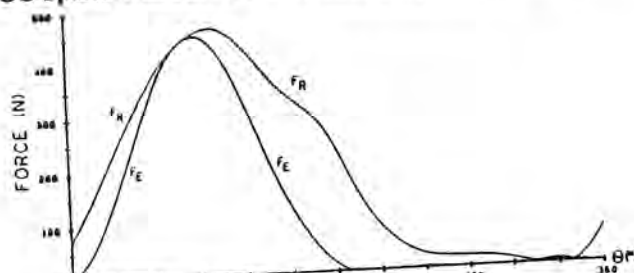


Figure 2. Plots of F_E versus θ and F_R versus θ for Group 1.

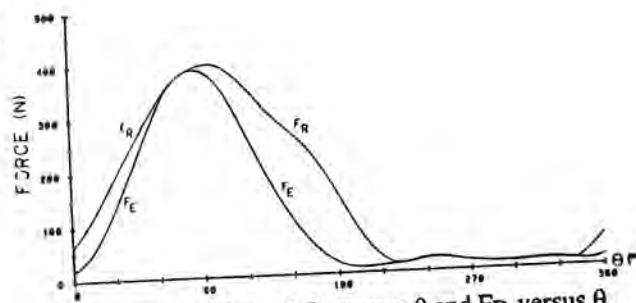


Figure 3. Plots of F_E versus θ and F_R versus θ for Group 2.

LOWER EXTREMITY MOMENTS OF FORCE DURING THREE GAIT CONDITIONS

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The examination of lower extremity joint moments of force (JMF) during locomotion has been suggested as a method to more fully understand mechanisms of movement (Winter, 1983). JMFs during walking and running at various speed have been examined independently by several researchers (Simpson, 1987; Winter, 1983; Robertson & Taunton, 1982) however, limited information is available describing the transition speed between walking and running. Descriptive information on the same population across speeds could be useful in better understanding the mechanisms of human gait. In addition, little JMF data has been presented for female performances. The purpose of the study was to evaluate and compare the lower extremity joint moments of force produced during the support phase for walking, running and a transition pace for female subjects.

METHODOLOGY

Three female subjects performed 10 trials at each of three speed conditions: run (R) = 12.9 +/- 0.8 km/h; walk (W) = 6.5 +/- 0.5 km/h; transition (T) = maximum voluntary walking speed. Simultaneous force (500 Hz) and sagittal plane film records (100 fps) were collected during right foot support on the force platform. Required anthropometric measurements were obtained via a modified Hanavan technique (Miller & Morrison, 1975). Film data were digitized, optimally filtered, and derivatives calculated (Hatze, 1980; Biolib, 1977). Support phase ground reaction force (GRF) data were appropriately scaled and center of pressure (CP) values calculated. The CP data were then adjusted to the spatial reference system of the film. GRF, CP, and film data were interpolated to 200 equally spaced intervals for all gait conditions using a cubic spline routine. JMFs for the ankle, knee, and hip joints were calculated and an ensemble JMF curve created for each 10 trial subject-condition.

RESULTS AND DISCUSSION

Group mean ensemble curve values and standard deviation values are presented in Table 1. The mean values represent trends apparent for all subjects although the magnitudes of the differences are not totally reflected by the standard deviations due to the small sample size. Evaluation of mean maximum joint effects collapsed across speeds indicated that knee extension was the dominant action (180.3 Nm) followed by first maximum hip flexion (151.5 Nm). Maximum ankle flexion (27.6 Nm) and knee flexion (58.0 Nm) exhibited the smallest values. Temporal changes between flexor and extensor moments and time of JMF peaks were generally more stable across the subjects compared to the JMF magnitudes.

It was hypothesized that maximum flexor and extensor JMF values would follow changes in speed. This trend was apparent for both maximum flexor and extensor moments. Collapsing across joints for each condition resulted in average maximum flexor moments of 85.1, 63.9 and 55.6 Nm for the R, T and W conditions, respectively. The average maximum extensor moments were an average of 1.7 times greater than flexor moments (149.0, 107.7 and 94.5 Nm). Average flexor and extensor values followed the patterns of the maximum moment values.

Overall variability was greatest during T, as might be expected due to the relative novelty of the task. Maximum flexor and extensor moments during T deviated 4.5, 62.3; 53.5, 28.6; and 61.6, 18.5%, from the mean values for the ankle, knee and hip joints, respectively. The results of this study are consistent with previous research. Winter (1983) reported that the primary function of the knee is to absorb energy. Knee extensor moments were predominant in the current investigation. Greatest extensor moments were observed during R indicating speed dependence as previously indicated by Simpson (1987).

Table 1. Group Mean Ensemble Joint Moment Curve Characteristics.

Table 1. Group Mean Ensemble Joint Moments						
	RUN		TRANSITION		WALK	
	Flexor	Extensor	Flexor	Extensor	Flexor	Extensor
ANKLE JOINT						
Max Value (Nm)	38.5/20.4	102.9/24.2	27.6/17.2	96.0/4.4	16.7/6.6	105.2/7.3
Time Max (% S)	12.9/2.2	60.5/5.0	12.8/2.7	77.1/1.5	10.6/1.6	77.9/1.5
Avg Value (Nm)	22.6/13.4	58.0/12.4	13.5/9.4	41.9/4.0	7.5/3.8	46.0/6.2
Transition (% S)	28.4/11.3		35.0/10.1		24.9/6.1	
KNEE JOINT						
Max Value (Nm)	50.5/15.2	286.4/66.7	49.5/26.5	149.7/42.8	73.9/3.3	104.8/16.6
Time Max (% S)	79.9/1.2	33.4/5.6	72.5/1.9	18.6/1.0	77.6/2.3	18.9/1.1
Avg Value (Nm)	27.4/8.2	158.8/41.7	27.2/3.8	51.2/13.8	35.9/1.0	42.4/7.0
Transition (% S)	69.5/3.5		54.4/6.7		47.1/2.9	
HIP JOINT						
Max1 Value (Nm)	196.4/10.4	70.6/26.9	155.8/28.9	120.9/74.5	102.2/26.5	93.2/28.5
Time Max1 (% S)	24.4/1.2	9.3/1.7	18.2/1.2	3.7/0.2	18.4/0.8	3.0/0.7
Avg Max1 (Nm)	96.3/9.1	28.0/11.2	57.3/12.1	68.3/29.4	36.9/9.3	50.7/15.8
Max2 Value (Nm)	55.1/21.4	136.0/28.3	22.8/14.4	64.2/4.9	29.4/5.6	74.6/22.1
Time Max2 (%S)	99.8/0.2	73.9/0.8	99.0/0.0	81.3/0.7	98.2/1.9	74.5/11.2
Avg Max2 (Nm)	32.3/14.3	79.5/18.5	25.2/0.0	49.3/4.0	18.9/4.1	37.2/8.3
Transition1 (% S)	13.8/2.2		7.5/1.5		7.2/0.6	
Transition2 (% S)	54.4/1.6		53.2/5.4		49.2/4.5	
Transition3 (% S)	91.0/2.4		92.0/3.9		92.0/2.0	

Note: Mean is followed by standard deviation; % S = percent of support phase.

ACKNOWLEDGEMENT

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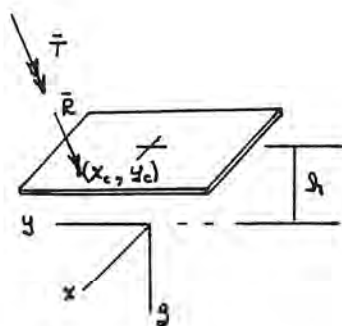
Analysis of Ground Reaction Torque in Walking and Running

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Current force platforms measure the three components of the ground force between the foot and the platform and the three components of the moment of this force about the instrumentation center of the platform. The measured resultant force, $\bar{R}$, and the moment, $\bar{M}_O$, may be resolved into a unique wrench or screw axis comprised of the resultant force and a parallel torque vector, T . The intercept of this screw axis with the surface of the platform (x_c, y_c) defines the force concentration point on the foot (see Figure 1). Previous methods define a center of pressure (x_{cop}, y_{cop}) to be the centroid of the vertical force distribution. It was assumed that the resultant force and an equivalent moment about the z axis, $\bar{M}_e^z$, acted through that point (see Figure 2). The wrench torque is always equal or less in magnitude to this moment and does not parallel it.

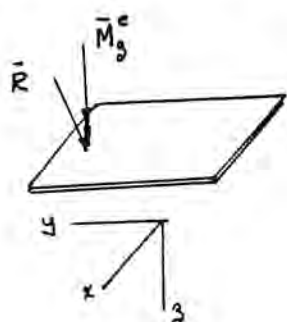
The ground reaction torque has received minimal attention in studying gait during walking and running and is of importance in evaluating pathological gait and understanding overuse injuries. During walking, the ground reaction torque is external (internal action of the proximal segment upon the distal segment) at heel contact and internal during forefoot contact and toe-off. These torques are shown in Figure 3 for double stance phase for left foot toe-off and right foot heel contact. Also shown are the appropriate braking and propulsion forces. At this point in gait the body has completed counterclockwise pelvic rotation and these torques and the couple due to the braking and propulsion forces provide the clockwise moment to advance the left leg. The dynamic inertia forces on the swing leg are minimal during walking.

During running, with only single foot contact (see Figure 4), these torques reverse in direction during heel contact and toe-off as compared to walking. As running speed increases, the clockwise ground reaction torque at heel strike controls the unbalance of angular momentum between the upper body and pelvis. At slower speeds the counterclockwise ground reaction force during propulsion (forefoot) becomes more evident and aids in producing the required angular momentum to rotate the right limb. Experimental tests of a subject walking from 0.8 to 3.2 m/s and running from 2.5 to 4.9 m/s verify these directions of the ground reaction torque. Torque magnitudes varied from 2 to 12 Nm during these tests and similar tests in pathological gait trials showed pattern and magnitude variations of these torques.



$$\begin{aligned}\bar{R} &= F_x \hat{i} + F_y \hat{j} + F_z \hat{k} & \hat{i}_R &= \bar{R} / |\bar{R}| \\ \bar{T} &= (\bar{M}_O \cdot \hat{i}_R) \hat{i}_R = [(M_{Oz} F_z + M_{Oy} F_y + M_{Ox} F_x) / |\bar{R}|] \hat{i}_R \\ x_c &= [-h F_x R^2 - M_{Oy} (R^2 - F_y^2) + M_{Ox} F_x F_y + M_{Oz} F_y F_z] / (F_z R^2) \\ y_c &= [-h F_y R^2 + M_{Ox} (R^2 - F_x^2) - M_{Oy} F_x F_y - M_{Oz} F_x F_z] / (F_z R^2)\end{aligned}$$

Figure 1. Screw Axis



$$\begin{aligned}\bar{M}_O^e &= \bar{M}_{Oz} - (F_y x_{cop} - F_x y_{cop}) \hat{k} = [(M_{Oz} F_z + M_{Oy} F_y + M_{Ox} F_x) / F_z] \hat{k} \\ x_{cop} &= -M_{Oy} / F_z & y_{cop} &= M_{Ox} / F_z\end{aligned}$$

Figure 2. Centroid of Vertical Forces

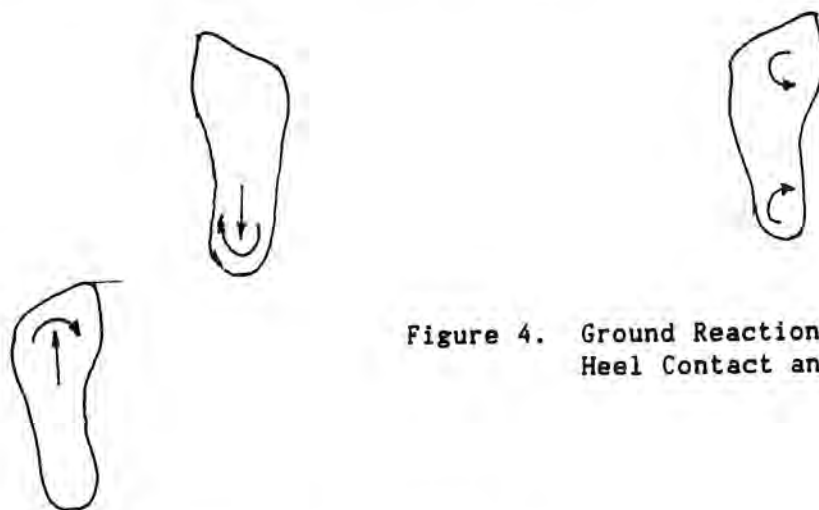


Figure 3. Ground Reaction Torques
During Double Stance Walking

Figure 4. Ground Reaction Torque
Heel Contact and Toe-Off Running

Supported, in part, by gift from Brooks Shoe Company, Inc.

Internal Structure Loading in the Knee Joint: A Biomechanical
Application of Optimization Theory^{†*}

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A nonlinear static optimization problem, minimizing the summation of the muscle stresses cubed, was formulated to solve the redundant biomechanical problem for the knee joint during stance phase of running. Thirty-seven muscles, thirteen ligamentous structures and joint contact forces were included in the three-dimensional model. Muscle origins and insertions were calculated from data published by Brand, et al. (1). Six variations of the biomechanical model of the knee were examined: I. Three-dimensional model, with moving joint contact and ligamentous contribution; II. three-dimensional model, with moving joint contact, no ligamentous contribution; III. hinge joint model, no ligaments; IV. moving joint contact, muscles satisfying only the flexion-extension moment; V. hinge joint model, muscles satisfying only the flexion-extension moment, and VI. three-dimensional model, no ligamentous contribution and solution tolerances relaxed so only 99% of the total joint moment was satisfied. The joint surfaces were represented two-dimensionally as suggested by Moeinzadeh (2) and the joint contact was assumed to be at a single point moving along a line midway between the medial and lateral epicondyles.

This problem was solved at discrete time intervals in stance phase of gait for a female subject, 16 years of age and weighing 100 pounds. Kinetic and kinematic inputs to the model such as ground reaction forces, linear and angular accelerations, and joint angles were calculated from three-dimensional film and force plate data collected as the subject ran barefoot across the force plate.

In this model, the ligament contribution to the moment at the knee depends solely on the relative orientation of the tibia and femur. For flexion angles between 0° and 50° , the inherent internal rotation of the tibia associated with the screw home mechanism of the knee was assumed to be a function of the fifth root of the angle of flexion (3). For most ligaments, this model predicts the greatest loading near full extension with structures becoming slack between 30° to 50° flexion. The constitutive equation for the ligaments was taken as the function of the square of ligament strain.

A cost function was chosen to maximize the endurance time (4). All ligament and muscle forces were constrained to be non-negative and an upper bound of 1 MPa was placed on the muscle stress. A generalized reduced gradient algorithm was used to find the solution to the problem.

For this model, the ligament contributions were found to be minimal. Simply increasing the tolerances on the equation of motion constraints without the ligaments by 1% (Case VI) gave results which were comparable to the full three-dimensional model (Case I). However, treating the knee as a simple hinged joint greatly affected the results. This may be seen by the comparison of Cases I and III in Figure 1. Of even greater importance in some muscles is neglecting the abduction-adduction and

internal-external moments. This is most obvious in the case of the lateral gastrocnemius comparing Cases I and III to Cases IV and V in Figure 2.

Although the models were solved at successive discrete time intervals, the continuity of the results indicates a useful solution without dynamic optimization.

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FOOTNOTES

⁺Supported, in part, by NSF Engineering Creativity Award and Brooks Shoe Company, Inc.

[#]Partial fulfillment of M.S. degree requirements of first author.

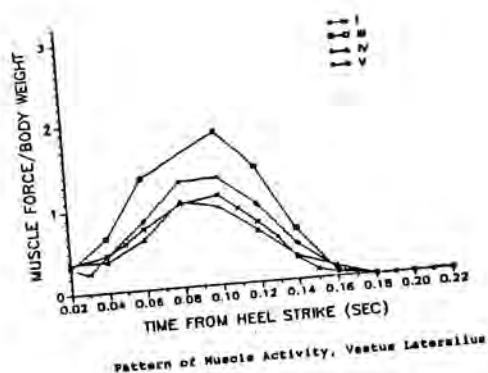


Figure 1

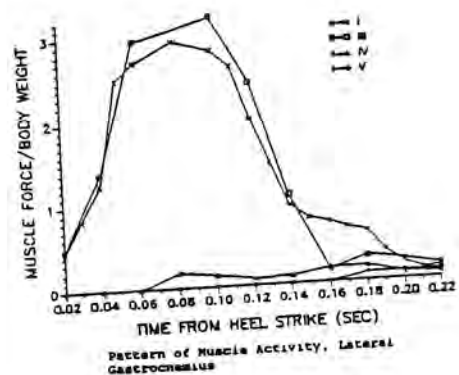


Figure 2

The Effect of Lower Extremity Inertia Manipulation on Walking Economy and Mechanics

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INTRODUCTION

It has been suggested that a potential source of interindividual variation in economy (i.e., aerobic demand) is differences in the distribution of mass among the segments of the body (Cavanagh & Kram, 1985; Myers & Stuedel, 1985). For example, lower extremity loading produces greater increases in aerobic demand of running than similar loading of the trunk, and mass distributed more distally on the extremities results in higher aerobic demands than those placed more proximally (Martin, 1985; Myers & Stuedel, 1985). This suggests that economy may be especially sensitive to mass distribution or moment of inertia manipulations of the lower extremity. Using the moment of inertia of the leg about a transverse axis through the hip (I_{hip}) as a descriptor of this mass distribution factor, the purpose of this study was to determine if matched increases in the moment of inertia achieved via different loading configurations produced similar changes in the aerobic demand and mechanics of walking.

METHODS

Fifteen men ($\bar{X}_{\text{age}} = 27.1 \pm 4.8$ yr, $\bar{X}_{\text{mass}} = 74.6 \pm 8.1$ kg, $\bar{X}_{\text{ht}} = 176.7 \pm 7.6$ cm) completed two test sessions: an orientation session in which subjects were acclimated to treadmill walking at the experimental speed of 1.69 m/s and a primary data collection session in which the subjects completed both treadmill and overground walking under five randomly ordered load conditions. These included a baseline condition (i.e., no added load), and loads of either 0.175 or 0.350 kg added to each foot near the segment center of mass. For the two remaining conditions, loads were added to the shanks. The goal for these loads was to match as closely as possible the increases in the average I_{hip} produced by the foot loads. Based on preliminary analysis of the subjects' walking patterns, it was predicted that loads of 1.13 and 2.25 kg positioned 25% of shank length below the knee would closely approximate the desired I_{hip} manipulation. Addition of load to the thighs was prohibitive because of the high masses that would have been required to produce the desired I_{hip}.

Aerobic demand under each load condition was determined from the last two minutes of a 6 min walk (1.69 m/s) on a motorized treadmill. The subject then completed 10 overground walking trials for each load at the same speed ($\pm 3\%$). Selected temporal and kinematic descriptors of the subjects' walking patterns for two of the 10 trials were quantified from a sagittal view recorded at 50 frames per second using standard high speed cinematography procedures.

RESULTS

Table 1 and Figure 1 summarize observed changes in temporal and kinematic descriptors of subject walking patterns and in aerobic demand expressed relative to total mass (i.e., body mass + load mass), respectively.

TABLE 1. Means for selected temporal and kinematic descriptors.

Variable	No Load	Foot Loading		Shank Loading	
		Light	Heavy	Light	Heavy
Average I_{hip} ($kg \cdot m^2$) ^{1,2}	2.68	2.91	3.24	3.00	3.32
Stride length (m) ³	1.727	1.722	1.743	1.725	1.712
Support time (ms)	645	650	654	650	650
Swing time (ms) ¹	391	395	403	392	392
Peak knee veloc. (m/s) ^{1,2}	3.32	3.31	3.28	3.30	3.21
Peak ankle veloc. (m/s) ^{1,3}	4.70	4.56	4.46	4.75	4.76
Peak thigh veloc. (rad/s) ¹	3.44	3.47	3.36	3.41	3.31
Peak knee veloc. (rad/s) ^{1,3}	5.57	5.74	5.80	5.51	5.39

Statistically signif. ($p < 0.05$) effects for ¹position, ²load, ³interaction

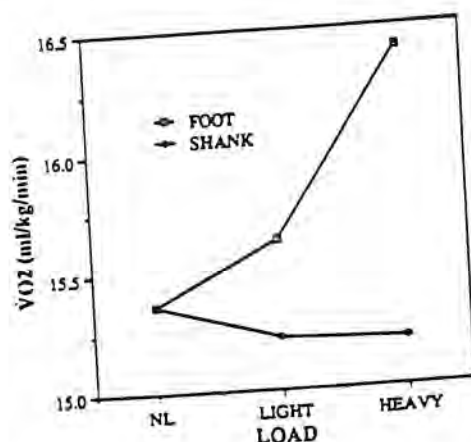


Figure 1. The effect of lower extremity loading on economy.

the results are clearly in contrast with those reported by Taylor et al. (1974) who concluded that the effect of differences in limb mass distribution on economy were too small to measure. However, it is clear that I_{hip} alone does not account for the observed economy variations since similar I_{hip} increases for shank and foot loading positions did not result in similar increases in aerobic demand. These results suggest that accelerating and decelerating more distally weighted limbs appears to be considerably more costly than for more proximally weighted extremities, while gait mechanics are affected only modestly by this factor.

DISCUSSION

Results of the analysis demonstrated clearly that shank and foot loading produced differential effects on walking economy, despite similar increases in I_{hip} for the two loading positions. While aerobic demand declined slightly under the shank loading, it increased substantially with loading of the feet. Similarly, gait mechanics were affected differently by foot and shank loading, although all adjustments were rather small. These results generally agree with those reported previously by Martin (1985) and Myers and Steudel (1985) who showed that more distal loading of the extremities resulted in greater aerobic demand. Conversely,

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KINETIC ANALYSIS OF THE SERVING ARM DURING THE PERFORMANCE OF THE TENNIS SERVE

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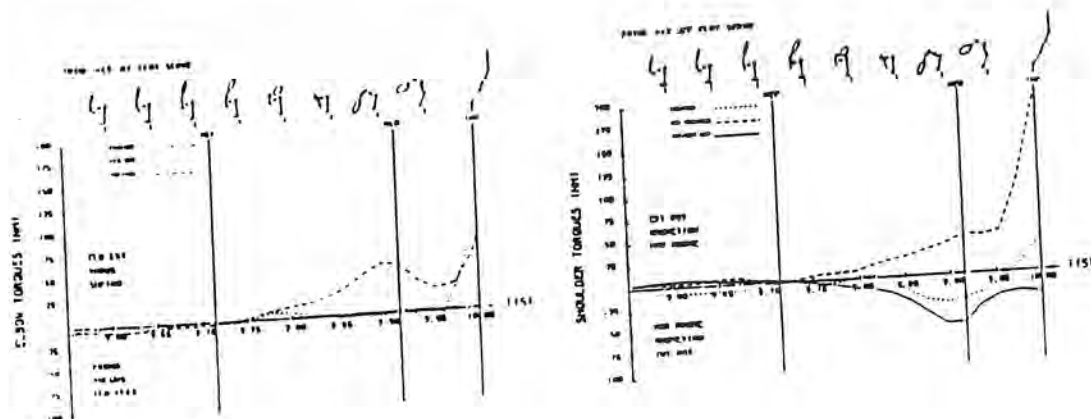
In the sport of tennis, the serve is the most important offensive weapon used by the players. A successful serve depends on the velocity of the ball after impact which in turn is determined by the velocity of the racket prior to impact. The magnitude of the linear velocity of the racket prior to impact is a factor of the forces transmitted to the racket via the player's serving arm. A great deal of the analyses available on the tennis serve dealt with the kinematic parameters or were performed using two-dimensional (2D) cinematography (3,4,5,6). A few studies have been performed using three-dimensional (3D) cinematography (1,7,8), but they were limited to kinematic analyses.

Six male, right-handed, intercollegiate tennis players were asked to perform several trials of a hard-driven flat serve while they were filmed using the Direct Linear Transformation (DLT) of 3D cinematography. The serve with the greatest ball velocity for each subject was analyzed until the time of impact. 3D coordinate data of 21 body landmarks and five racket points were obtained and smoothed using quintic spline-functions. The events of maximal elbow flexion (MEF), maximal external rotation (MER), and impact (IMP $t=10.000$ s) were defined in order to facilitate the description of the patterns of motion. Rigid body kinematics and the inverse dynamics approach were used to calculate the joint torque and force at the respective joints. The serving arm was modeled as a four-link kinetic chain (racket, hand, forearm, and upper arm). The joint force and torque values for each of the joints were calculated in terms of a reference frame defined on the ground and later transformed to anatomically relevant, non-inertial reference frames defined at the shoulder and elbow joints.

The force and torque data exhibited similar patterns of motions for all subjects. During the backswing phase from $t=9.500$ s to $t=9.750$ s small joint forces and torques were utilized at the shoulder and elbow joint (± 0.064) and the beginning of the forward swing (acceleration phase). At this position, the upper arm was about shoulder level and the forearm, hand, and racket were beginning to go behind the back. The trunk began to produce a counterclockwise rotation about its longitudinal axis and an upward and forward motion produced by the leg drive. The motions of the trunk caused external rotation, abduction, and horizontal adduction of the upper-arm, mainly created by the inertial lag of the forearm, hand, and racket. During this period, starting at MEF, the shoulder began to produce internal rotation, abduction, and horizontal adduction torques. At the elbow, there was a varus torque present. These torques increased steadily, with the internal rotation torque achieving its maximal value of -50.80 Nm prior to the instant of MER ($t=9.938$ s ± 0.02). At the instant of MER, there was an internal rotation torque (-21.96 Nm ± 11.3), an abduction torque (-43.03 Nm ± 11.5), and a horizontal adduction torque (39.27 Nm ± 15.3) at the shoulder. The elbow joint had a varus torque of 43.32 Nm ± 11.8 .

The adduction torque began to decrease, becoming an adduction torque until impact ($30.33 \text{ Nm} \pm 30.7$). The internal rotation torque decreased slightly, but just near 0.025 s prior to impact ($t=10.000\text{s}$) it began to increase again until contact ($-33.02 \text{ Nm} \pm 21.1$). The horizontal adduction torque decreased slightly, but approximately 0.050 s prior to impact there was a rapid increase which continued until impact ($164.85 \text{ Nm} \pm 62.6$). At the elbow joint, the varus torque slightly decreased until $t=9.975 \text{ s}$ then it began a rapid increase until impact ($74.16 \text{ Nm} \pm 32.1$). Also, around $t=9.975 \text{ s}$, there was a rapid extension torque continuing until impact. This extension torque was present in four of the subjects. Two subjects showed a decrease in the extension torque which became a flexion torque just before impact. These two subjects had the highest ball and racket velocities.

In conclusion, during the early stages of the serving motion (backswing) small torques and forces are utilized to position the serving arm at the beginning of the forward swing (acceleration phase). The acceleration phase is mainly produced by the internal rotation and the horizontal adduction torques at the shoulder and by the rapid extension torque at the elbow joint. A large varus torque at the elbow joint was present throughout the forward swing.



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A New Transducer Technology for Measurement of Plantar Pressures

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There have been many technologies and methods developed for recording pressures. Some of these include optoelectronic (Betts et al., 1980), resistive strain gauges (Soames et al., 1982), capacitive (Nicol and Hennig, 1978), and piezoelectric (Hennig et al., 1982). All of these methods have their limitations which include: patient targeting; subject encumbrance; single step data; complex hardware and software for signal amplification and processing; and dynamic operation only. We are utilizing a new technology that eliminates most of the limitations outlined above. The Force Sensing Resistor (FSR) was developed by Interlink Electronics, of Santa Barbara, CA. The device has some unique characteristics and, in particular, a change in resistance with applied force. We have developed a modified version of the FSR which consists of an electrically conductive polymer matrix manufactured by Interlink covering a flexible printed circuit board layout. Figures 1 and 2 depict two layout designs and their very different output results. An excitation of 5 volts was applied to the transducers. One complete cycle of loading and unloading are presented. The output voltage versus applied pressure is shown with each transducer. It can be seen from the output curves that the device has a non-linear output. This characteristic may be undesirable in some situations but, in our application, it can be accommodated. Knowledge of this non-linearity has allowed software to be developed that has appropriate linearization algorithms. The effects of different element layouts can be seen in the respective outputs. A usable range of 3 to 4 MPa is possible with transducer 2. Hysteresis is not evident which greatly simplifies processing. The output of the device over several cycles of loading and unloading shows that there is little if any time delay in the output of the transducers. Cyclic loading tests have shown that the output is consistent and stable over thousands of cycles. Figure 3 shows a block diagram of the design to measure plantar pressures. A very thin (0.6 mm) 32 element

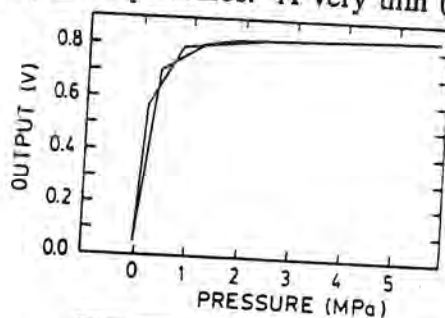


Figure 1

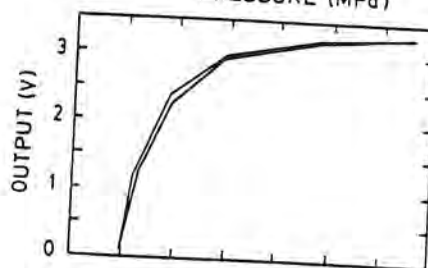


Figure 2

transducer array has been constructed using a flexible printed circuit material and conventional processing techniques. A small backpack worn around the waist contains a battery power supply along with the appropriate multiplexing and switching circuitry. Analog data can be fed via umbilical wiring or eight channel telemetry to an analog to digital converter. Processing of the raw signal (no amplification required) is done on an IBM PC or compatible with the appropriate software. Meaningful output can be available almost immediately in many forms. We believe that this new method of measuring pressures has significant advantages over previously used methods. One of the major advantages is the elegant operation of the system: amplifiers and data filtering are not required. This directly leads to the low cost, versatility, and adaptability of the system. The usable range of the transducers is also very high in terms of recording plantar pressures, and printed circuit technology is well established. The thickness and weight offer no hindrance to normal subject movement. Users are able to adapt the system quite easily to their analog to digital systems via hard wiring or telemetry. These advantages make the system a valuable tool in the study of gait and many other applications where pressure data are required.

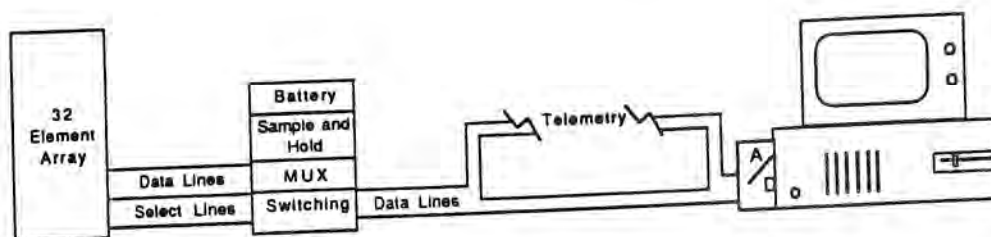


Figure 3

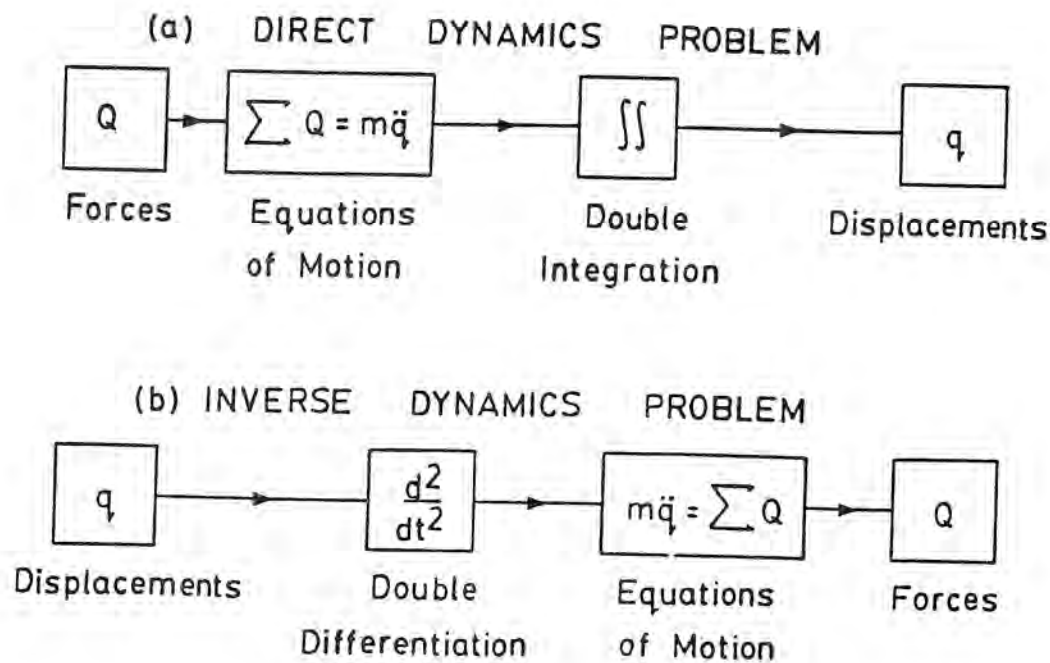
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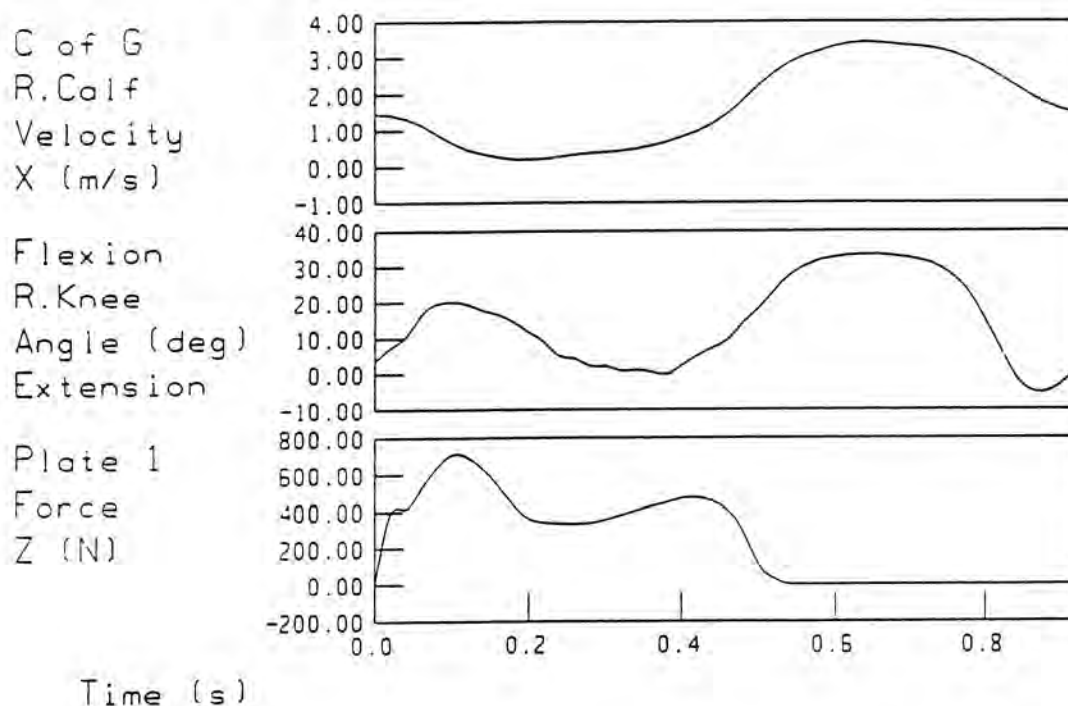
In Search of the Homunculus

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When we think about the way in which the human body walks, the analogy of a marionette springs to mind. Perhaps the puppeteer who pulls the strings and controls our movements is an homunculus, the supreme commander of our locomotor program. This point is illustrated in a rather humorous but revealing way by Inman et al. (1981). The purpose of this paper is to create a structural framework that enables us to understand the way that gait analysis should be performed and thus how all the different processes are integrated. We are proposing that the gait analyst should try to move as far back in the movement chain as possible, so that he or she approaches the *causes* of the walking pattern, as opposed to studying only the *effects*. As pointed out by Vaughan et al. (1982), there are essentially two types of problems in rigid body mechanics. The first is the Direct Dynamics Problem in which the forcing functions applied to a mechanical system are known and the objectives are to determine the resulting motion of the system. This is the approach the homunculus pursues. The second is the Inverse Dynamics Problem in which the motion of the mechanical system is completely specified and the objective is to find the forcing functions causing that motion. This is the approach that the gait analyst pursues. We have developed a software package that enables those who have only an IBM PC (or compatible) computer to perform



detailed 3D gait analysis (Vaughan et al., 1989). There are four input data files: electromyography; anthropometry; 3D segment displacements; and 3D ground reaction forces. The program GAITMATH is used to derive another six files: body segment parameters; joint segment orientations; segment center of gravity kinematics; anatomical joint angles as well as segment angular velocities and accelerations; and dynamic forces and moments at the joints. There are over 240 variables which can be viewed with the program GAITPLOT. This powerful, interactive environment allows both the novice and experienced gait analyst to explore the various interconnections between the neural, muscular, skeletal and external systems. In particular, such a flexible environment (only one of literally millions of plots is illustrated here) can play a pivotal role in helping to elucidate the cause-effect relationships in human gait.



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ASSESSMENT OF KNEE JOINT POSITION SENSE FOLLOWING SURGICAL RECONSTRUCTION FOR ANTERIOR CRUCIATE LIGAMENT INSUFFICIENCY

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INTRODUCTION. One of the most important aspects of proprioception is joint position sense, particularly following anterior cruciate ligament (ACL) trauma and/or surgery. In the absence of adequate peripheral feedback regarding limb position in space, the ACL-insufficient knee may be placed at increased risk of injury during simple locomotor tasks such as walking and descending stairs. The purpose of this study was to assess joint position sense (JPS) in the post ACL-reconstruction and contralateral normal knees of 48 male and female subjects at two postoperative intervals (intermediate = 24 to 40 months; long term = 41 to 101 months).

METHODS AND MATERIALS. The ability to reproduce the position in which a joint has previously been placed has commonly been used to evaluate proprioception (2, 3, 4, 5). The subjects who participated in this study ranged in age from 18 to 49 years (mean = 23.7 yrs), and were evaluated at an average of 4.0 years (minimum = 2.0 years) following autogenous ACL-reconstructive surgery. The subjects were seated on a Cybex II dynamometer (Cybex Division, Lumex Corp., Ronkonkoma, NY), and constrained in a manner similar to that utilized to perform knee extension/flexion exercise. Subjects were blindfolded and tested on their ability to actively reproduce five passive knee positions. Limb position was determined from the digital output from the dynamometer's internal goniometer interfaced with the system's dedicated computer. Starting at a position of 90 deg flexion, the examiner passively extended the subject's leg by pulling upward and forward on the dynamometer lever arm at a rate of 10-15 deg/sec. The leg was extended to one of five random positions, to target angles of 15, 20, 25, 30, and 35 deg. This process was repeated once for each of the five positions, with both limbs being evaluated. A Latin squares design was employed to control for learning effects. The limb was held at the target angle for approximately 3 sec, then returned to the starting position. Subjects were then asked to actively move the limb to the previously placed position, giving a verbal indication when they perceived they had reached the target angle. The system described permitted real time monitoring of joint position to the nearest degree. An analog signal was plotted on a X-Y recorder to provide additional information, e. g., velocity of movement, target angle overshoot or undershoot. Mean algebraic error and mean absolute error values were determined.

RESULTS. Mean absolute error for the ACL-reconstructed knees was 5.4 ± 3.2 deg compared to 5.2 ± 2.7 deg for the contralateral normal knees. Paired t-tests revealed no significant differences in JPS between the ACL-reconstructed and normal knees at any of the five positions tested ($p > .15$) [Table 1]. Analyses of variance revealed no significant differences in mean algebraic or mean absolute error between the intermediate and long term postoperative groups. No clear pattern or strategy for reproduction of passive positioning emerged, as an average 47.5% of the subjects "overshot" the five target angles, 45.8% "undershot" the target angles, with an average of 6.7% of the subjects being "on target" (zero error).

TABLE 1. Contralateral Mean Absolute Error in Knee Joint Position Sense (in degrees).

KNEE TARGET ANGLES					
KNEE:	15 deg	20 deg	25 deg	30 deg	35 deg
POST-SURGICAL	5.58	5.92	4.98	4.73	5.35
NORMAL CONTRA-LATERAL	4.71	5.58	4.44	5.29	5.38

DISCUSSION. The results of the indicated no significant differences in knee joint position sense between the ACL-reconstructed and normal contralateral limbs. These findings are similar to those of Barrack et al. (2) and Grigg et al. (3), who reported that knee and hip JPS, respectively, was not altered by total knee arthroplasty or by total hip replacement. Confounding these results are recent studies by Newberg (4) and Barrack, Skinner, and Buckley (1), who observed significant differences in knee joint proprioception an average of 15 months after ACL reconstruction in the former case, and in knees with acute and chronic ACL insufficiency in the latter case. Several possible explanations exist regarding the factors responsible for the contralateral limb JPS symmetry observed in the current study group: (a) knee JPS was not disrupted by the ACL injury or subsequent surgical reconstruction, or (b) the postoperative interval required to regain symmetrical JPS was less than the two year minimum follow-up period designated in this study. The most plausible explanation, supported by the findings by Newberg (4) and Barrack et al. (1), is that knee joint position sense is disrupted in varying degrees by both the ACL lesion and autogenous ACL reconstructive procedures, but that the mechanoreceptors compensate by creating an "ensemble" of proprioceptive information (6) to offset any JPS loss prior to the two year minimum employed in this study. Additional prospective studies are required to investigate the temporal relationship between ACL injury, ACL reconstructive surgery, and knee joint position sense.

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GROUND REACTION FORCE SYMMETRY IN THE GAIT OF OSTEOARTHRITIC WOMEN Kathleen M. Knutzen, Lorraine R. Brilla, Kevin Quinn, Vauhn Wittman, Dept. of Physical Education, Health and Recreation, Western Washington Univ., Bellingham, WA 98225

Alterations in gait associated with the incidence of osteoarthritis (OA) have been reported by numerous investigators. Comparing osteoarthritic subjects with normals, investigators have reported that arthritic subjects walk with a reduced cadence, a reduced stride length, they spend more time in both stance and swing phases, they produce both lower vertical and anteroposterior forces, they produce greater medial forces at contact, and move through lesser range of motions at the knee and hip (1,2,3,5,6). Studies comparing the gait of normals and OA subjects have been hampered by the noted discrepancy in gait velocity between the two groups which has led to unresolved questions concerning the actual influence of the pathology on gait parameters. The purpose of the present investigation was to examine ground reaction force differences between limbs in order to assess the presence or absence of symmetry in the OA subject.

Subjects. Ten women, (mean age=68.7 $\pm$ 4.6 yr.) who had documented incidence of OA in the lower extremity, volunteered as subjects. Seven subjects reported OA in one or both hips, 1 subject reported OA in the knee, and 2 subjects reported a combination of hip and knee OA. The physical characteristics of the subjects were within normal ranges for height ($M=158.8\text{cm}$ $\pm$ 5.4) and weight ($M=72.2\text{kg}$ $\pm$ 13.8).

Methods. Ground reaction force data were sampled (200 Hz) utilizing an AMTI Force Platform System mounted within a 30 meter lane. Following the collection of subject history and physical characteristics, the subjects practiced walking through the experimental area until contact could be made with the platform with an uninterrupted gait cycle. Each subject was allowed to walk at a pace which they deemed comfortable, however, once a pace had been established, the subjects were instructed to maintain a similar pace throughout the test. The walking velocity was monitored over a six meter area using a photoelectric timing system. Five trials were collected for each limb.

The force platform data representing the three force components were scaled and normalized for body weight. Fourteen vertical, 13 anteroposterior, and 16 mediolateral variables were identified for each trial, and mean values were computed across the five trials. Limb comparisons were made between the limb determined to be most affected with the arthritis (AL) compared with the limb least affected or not affected with arthritis (UL). This was determined through the medical history questionnaire or through personal communication with each subject. Eight subjects reported the left limb to be the limb most restricted by osteoarthritis and

two subjects indicated the right limb to be the most affected. The statistical comparison between limbs was conducted through the application of a repeated measures ANOVA. Statistical significance was indicated by a value of $p < .05$.

Results. The walking velocity for the affected limb trials ($M=1.00 \text{ s} \pm .132 \text{ m/s}$) did not differ significantly from the trials for the unaffected limb ($M=1.02 \text{ s} \pm .157 \text{ m/s}$) as determined by a dependent t-test ($p < .05$). The subjects generally walked at slow speeds ranging from .87 to 1.27 m/s.

Vertical ground reaction forces did not differ between AL and UL conditions. The first maximum force was achieved at 14.3 and 14.8% of the support phase for AL and UL, respectively. The value of the first maximum forces were the same (8.4 N/kg) and these forces were considerably less than the second maximum forces (AL=10.9 N/kg, UL=10.8 N/kg). The time to attainment of the second maximum vertical force (AL=66.1%, UL=66.1%) and the total impulse (AL=5.4 N s, UL=5.8 N s) were also similar between limbs.

There were no significant differences in the anteroposterior ground reaction force variables. Maximum braking forces were the same for both AL and UL (1.6 N/kg) and the time to transition to propulsion varied minimally between limbs (AL=52.3%, UL=52.1%). Similarly, maximum propelling forces (AL=1.6 N/kg, UL=1.7 N/kg), braking impulses (AL=.255 N s, UL=.248 N s), and propelling impulses (AL=.290 N s, UL=.294 N s) did not significantly vary between limbs.

The performance of the two limbs varied significantly in the mediolateral direction. These changes were most predominant in the early stages of footfall. For the first 30% of the support phase, the affected limb produced a significantly greater negative impulse (AL=.013 N s, UL=.009 N s), a greater algebraic impulse (AL=.010 N s, UL=.005 N s), a greater average force (AL=.043 N/kg, UL=.022 N/kg), and greater total excursions (AL=.504 N, UL=.026 N). This pattern repeated itself throughout the total support period, although the comparisons were not significantly different in the middle portions of the support phase (30-60%). Additionally, comparisons made for 0-100% of the support phase yielded significant differences between the limbs for all of the aforementioned variables. In conclusion, the mediolateral ground reaction forces were the only asymmetrical components present in the osteoarthritic gait.

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Calculating Ideal Bat Weight Using the Muscle Force-Velocity Relationship

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Why are professional baseball players "corking" their bats (Gutman, 1988; Kaat, 1988)? The answer may lie in the results of two Physics studies of bat-ball collisions that concluded the *optimal bat weight* is 15 ounces (Kirkpatrick, 1963; Brancazio, 1987). However a 15 ounce solid wood bat would only be 15 inches long. So perhaps the professionals are corking their bats to make them lighter. This conclusion might also explain why ball players choke up on the bat; choking up makes the bat effectively shorter, moves the center of mass closer to the hands thereby reducing the moment of inertia, and in essence makes the bat act like a lighter bat. This conclusion could also help explain the great popularity of aluminum bats. The manufacturers can make them lighter while maintaining the same length and width. However, both of these Physics studies were limited by their explicit assumptions. Kirkpatrick assumed that the optimal bat was the one that required the least kinetic energy. And Brancazio's calculations explicitly assumed that the "batter generates a fixed quantity of energy in a swing," independent of the bat weight. Let us now extend these calculations by allowing the amount of energy imparted to the bat by the batter to depend on bat weight.

Physiologists have long known that muscle speed decreases with increasing load (Fenn and Marsh, 1935; Hill, 1938; Wilke, 1950). This is why bicycles have gears: so the rider can adjust muscle speed to the maximum energy transfer point while bicycle speed varies greatly. This force-velocity relationship shows that the kinetic energy put into a swing will be zero when the bat weight is zero, and also when the bat is so heavy that the speed is reduced to zero. The *maximum-energy-transfer bat weight* will be somewhere in between. However, this maximum-energy-transfer bat weight does not tell us the bat weight that will make the ball go the farthest. To calculate this weight we must couple the muscle force-velocity relationship to the equations for conservation of momentum. Then we can solve the resulting equations to find the bat weight that would allow a batter to impart the maximum speed to the ball. This would, of course, make a potential home run go the farthest, and give a ground ball the maximum likelihood of getting through the infield. We call this weight the *maximum-ball-speed bat weight*.

We used professional and amateur baseball players to correlate the two most significant parameters of the swing of a baseball bat: bat weight and maximum bat speed. Our instrument for measuring bat speed, the Bat Chooser†, has two vertical light beams each with an associated light detector (similar to an elevator door electric eye). The subjects swing a bat so that the center of gravity of the bat passes through the light beams. The computer records the time between the interruptions of the light beams. Knowing the distance between the light beams and the time required for the bat to travel that distance, the computer calculates the speed of the bat's center of gravity for each swing. Each subject swung six bats (ranging in weight from 14 to 49 oz) through the light beams. Each bat was swung five times. The order of presentation was randomized.

For the data for the 30 swings, we fit three different, physiologically realistic (Agarwal and Gottlieb, 1984), equations: a straight line $y = Ax + B$, a hyperbola $(x + A)(y + B) = C$, and an exponential $y = Ae^{-Bx} + C$. We chose the equation that gave the best fit. Next we substituted this relationship into the ball speed equation derived from the conservation of momentum equations, took the derivative with respect to the bat weight, set this equal to zero, and solved for the maximum-ball-speed bat weight. For one subject this was 39 oz. However, the ball speed curve is pretty flat between 30 and 40 oz. There is only a 1.1% difference in ball speed between the 31 oz bat this player normally used and a 39 oz bat. Evidently the greater control permitted by the 31 oz bat outweighs the 1.1% increase in speed that could be achieved

† Bat Chooser is a trademark of Bahill Intelligent Computer Systems, patents are pending.

with the 39 oz bat.

We do not believe that the maximum-ball-speed bat weight is the best bat weight for any player. A lighter bat will give a player better control and more accuracy. So obviously a trade-off must be made between maximum speed and controllability. Because the ball speed curve is so flat around the point of the maximum-ball-speed bat weight, we believe there is little advantage to using a bat as heavy as the maximum-ball-speed bat weight. Therefore, we have arbitrarily defined the *ideal bat weight* to be the weight where the ball speed curve drops 1% below the speed at the maximum-ball-speed bat weight. Using this criterion, the ideal bat weight for this subject is 32 oz. We believe this gives a reasonable trade-off between distance and accuracy. Of course this is subjective and each player might want to weigh the two factors differently. But at least this gives a quantitative basis for discussion.

The ideal bat weight varies from person to person. When we used our instrument on eight adults (four engineers, and four members of the University of Arizona baseball team) for an 80 mph pitch, the ideal bat weight ranged between 27 and 32 oz. When our analysis was applied to three boys 8, 10 and 11 years old for a 40 mph pitch (which is the typical pitch speed for this age group), the ideal bat weights were found to be 18, 13, and 17 oz. They normally used 23, 23, and 26 oz bats. The lightest Little League approved bat that we have seen is 20 oz. When these calculations were applied to Bahill playing slow pitch softball (pitch speed 18 mph), the ideal bat weight was found to be 20 oz. The lightest legal softball bat that we have seen is 28 oz. This implies that Little League and softball players cannot yet buy bats that are light enough for them.

As previously noted we fit three different equations to our force-velocity data: straight lines, hyperbolas, and exponentials. In half of our 18 experiments (including all those involving the University of Arizona baseball players) all three curves yielded excellent fits with the straight lines being slightly (less than 1%) better. The other half of our experiments had larger standard deviations, so the fits were not as good. For these the hyperbolic and exponential curves fit the data much better than the straight lines. However, most of these hyperbolic fits were not typical Hill equations, because the curves did not intersect both axes. We conclude that for whole intact human beings the muscle force *does* depend on muscle speed, and in general hyperbolic and exponential equations fit the data best.

The physics of bat-ball collisions predict an *optimal bat weight* of 15 oz. The physiology of the muscle force-velocity relationship showed that our star subject could put the most energy into a swing with a 50 oz bat, i.e. the *maximum-energy-transfer bat weight* was 50 oz. Coupling physics to physiology showed his *maximum-ball-speed bat weight* to be 39 oz. Finally trade-offs between maximum ball speed and controllability showed that his *ideal bat weight* was 32 oz. These experiments explain why most adult batters use bats in the 30 to 34 oz range, they explain the variability in human choice of bat weight, they suggest that amateurs should use much lighter bats, and they suggest that there is an ideal bat weight for each person.

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EARLY BONE STRUCTURE AND FUNCTIONAL RECONSTRUCTIONS AT THE END OF LONG BONES AFTER TUMOR REMOVAL

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Tumor at the end of long bone treated by curettage and bone grafting is easier to recur. Complications following tumor resection and prosthesis are frequently encountered, which is not indicated in younger patients. An immediate biological and mechanical reconstruction of bone structure is recommended as follows. After tumor removal, the remaining bone shell and cavity are treated by soaking in 75% alcohol for 30 minutes or by freezing and heating for 3-5 times. Autogenous ilium and fibula are used for grafting, the ilium block is placed under the subarticular bone, one end of the fibula is inserted into the intramedullary cavity of normal fragment and another end is put against the ilium block, which are fixed with a L-TCP (Trapezoidal compression plate). The structure of bone grafts and metal device are required to agree with the principle of triangular truss. Both biological and mechanical frames are then established. The gaps between the bone and metal are filled with bone cement. Finally a normal contour of original bone will restore. The patient is permitted to move the joint immediately or 3 weeks later, and to walk without support 3-6 months postoperatively. The results of clinical angiography and disulphine blue infusion in animal experiment showed that the bone grafts were alive with normal blood supply.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

QUANTIFICATION OF NEUROMUSCULAR-PROPAGATION FAILURE IN SINGLE MOTOR UNITS OF CATS DURING A FATIGUE TEST

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Among the peripheral mechanisms that may contribute to muscle fatigue are failure of the action potential to invade all axonal branches, failure of transmission across the neuromuscular junction, and decreased excitability of the sarcolemma. In most nerve-muscle preparations, it is difficult to distinguish between these mechanisms. Rather, many investigators monitor the net effect of the three mechanisms, often referred to as "*neuromuscular-propagation failure*", by stimulating a nerve or a small bundle of axons and measuring the elicited muscle compound action potential. Despite the widespread use of this technique, there is disagreement over which feature of the muscle action potential is the most appropriate index of neuromuscular-propagation failure. Some investigators have argued that it is sufficient to measure the amplitude of the muscle action potential (Bellemare and Garzaniti, 1988; Stephens and Taylor, 1972) while others have suggested that area is the more appropriate measure (Bigland-Ritchie, Kukulka, Lippold, and Woods, 1982). The purpose of this study was to determine whether it is necessary to use action-potential area to describe the changes in the evoked compound muscle action potential or whether the measurement of action-potential amplitude is sufficient.

Experiments were performed on adult cats (1.7-4.6 kg) that were deeply anaesthetized with sodium pentobarbital. The hindlimb was denervated except for the nerve innervating the tibialis posterior muscle. A low compliance line served to attach the tendon of the muscle in series to a force transducer. The L6-S1 ventral roots were transected proximally and stimulated individually while recording whole-muscle force. Those ventral roots whose stimulation resulted in detectable force were divided into 30-40 fine filaments. These filaments were activated repetitively (2.5 Hz) with rectangular-wave electrical pulses (0.1 ms duration; 2.5-3.0 times threshold) by a monopolar, stainless-steel electrode (anode) referred to a similar electrode embedded in the back musculature. A single motor unit was considered to be isolated when a unitary action potential appeared in the averaged neurogram and when the profile or magnitude of both the force and EMG responses showed no dependence on stimulus strengths above threshold. Whole-muscle and motor-unit forces were recorded using transducers capable of accommodating forces in the 0.1 mN-10 N range. EMG activity was measured with a spring-mounted, uninsulated, stainless-steel wire that was inserted along the intramuscular tendon in a distal-to-proximal direction or by a silver-ball electrode that was placed in contact with the visible surface of the muscle.

The data reported here concern changes in the EMG of single motor units (n=45) during a standard fatigue test. The fatigue test comprised intermittent stimulation of the ventral root filaments with 330-ms trains of 13 stimuli (40 Hz) that were delivered once each second for 360 s. Measurements were averaged for the 13 waveforms of each train. The EMG, which represented the compound action potential from several muscle fibers of single motor units, was characterized by the measurement of amplitude, duration, area (integral of the absolute value), and a "shape factor" (k) that related area (A) to amplitude (a) and duration (d):

$$A = kad$$

$$k = A/ad$$

The shape factor was a coefficient of proportionality that always had a value less than one. The purpose was tested by determining whether or not k remained constant during the fatigue test. If it did remain constant, then area was not an independent parameter but rather it was proportional to the product of amplitude and duration.

Table 1. Initial absolute and subsequent relative values for the parameters used to describe the evoked EMG during the fatigue test.

		S (n = 16)	FR (n = 13)	F(int.) (n = 3)	FF (n = 13)
Amplitude					
Initial	(μ V)	293 $\pm$ 392	178 $\pm$ 252	200 $\pm$ 93	667 $\pm$ 1242
120 s	(%)	94 $\pm$ 9	140 $\pm$ 26	92 $\pm$ 25	68 $\pm$ 24
360 s	(%)	90 $\pm$ 17	78 $\pm$ 21	47 $\pm$ 17	13 $\pm$ 10
Duration					
Initial	(ms)	2.99 $\pm$ 0.62	2.86 $\pm$ 0.79	3.32 $\pm$ 0.27	3.05 $\pm$ 0.68
120 s	(%)	113 $\pm$ 12	109 $\pm$ 10	133 $\pm$ 26	169 $\pm$ 41
360 s	(%)	116 $\pm$ 15	138 $\pm$ 58	129 $\pm$ 52	189 $\pm$ 61
Area					
Initial	(μ V $\cdot$ ms)	210 $\pm$ 283	125 $\pm$ 179	160 $\pm$ 94	392 $\pm$ 701
120 s	(%)	105 $\pm$ 7	106 $\pm$ 10	132 $\pm$ 18	134 $\pm$ 37
360 s	(%)	104 $\pm$ 16	93 $\pm$ 25	65 $\pm$ 23	29 $\pm$ 20
Shape Factor					
Initial		0.25 $\pm$ 0.03	0.25 $\pm$ 0.03	0.24 $\pm$ 0.05	0.23 $\pm$ 0.04
120 s	(%)	101 $\pm$ 6	101 $\pm$ 10	109 $\pm$ 8	125 $\pm$ 32
360 s	(%)	102 $\pm$ 6	108 $\pm$ 19	115 $\pm$ 18	143 $\pm$ 51

Values are mean $\pm$ SD

Motor units were classified with conventional criteria (Burke, Levine, Tsairis, and Zajac, 1973) into four categories (type S, n=16; type FR, n=13; type F(int.), n=3; type FF, n=13). The degree of change in EMG varied with the fatigability of the motor unit; the most fatigable motor units (i.e., type FF) had the greatest declines in amplitude and area and the greatest increase in duration. Although the most fatigable units also had the greatest increase in k, the variability was so great that the change in k was not significant. This observation led us to conclude that k, the coefficient of proportionality, was constant for all four motor unit types during the 360-s fatigue test and that the EMG could have been adequately represented by the measure of amplitude.

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An Analysis of Trabecular Bone Micro-Mechanics using Homogenization Theory with Comparison to Experimental Results

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Introduction: The microstructural analysis of trabecular bone (1,5,8,10,13,16) is important for understanding bone remodeling around implants (4), normal equilibrium bone remodeling (3,7), pathologic bone fractures in osteoporosis (11), as well as other areas of skeletal biology. A new approach to the analysis of trabecular bone micro-mechanics using homogenization theory combined with morphological analysis is presented. A model of trabecular bone is developed within the definitions of homogenization theory that matches the morphological characteristics of trabecular bone. Here, the predictions of the procedure are compared to a standard finite element analysis and to experimentally determined trabecular bone stiffness.

Background: In homogenization theory, the total mechanical behavior of a microstructured material is represented by an asymptotic expansion of the field variable. It assumes the microstructure is periodic such that one representative micro model is analyzed for each specific microstructure. The field variable expansion for a displacement based formulation is: $u_i^e = u_{0i}(x_i) + \epsilon u_{1i}(x_i, y_i) + \epsilon^2 u_{2i}(x_i, y_i) + \dots$ where u_i^e is the total displacement, u_{0i} is the global displacement, u_{1i} and u_{2i} are the local perturbations of the displacement, x_i is the global coordinate, y_i is the local coordinate and ϵ is the ratio of the characteristic size of the microstructure to the size of the global body. When the expansion is substituted into the virtual displacement form of the equilibrium equations, the following equations result. Note that these represent analysis at the global level [1], the local level [2], and the coupling between the two [3,4,5].

$$\int_{\Omega} \left(\frac{1}{Y} \int_Y [E_{ijkl} - E_{ijpm} \frac{\partial \chi_p^{kl}}{\partial y_m}] \frac{\partial u_{0k}}{\partial x_l} \frac{\partial v_i}{\partial x_j} dY \right) d\Omega = \int_{\Omega} [f_i \delta y] v_i d\Omega + \int_{\Gamma} t_i v_i d\Gamma \quad [1]$$

$$\int_Y E_{ijpm} \frac{\partial \chi_p^{kl}}{\partial y_m} \frac{\partial v_i}{\partial y_j} dY = \int_Y E_{ijkl} \frac{\partial v_i}{\partial y_j} dY \quad [2]$$

$$u_{1i} = -\chi_i^{kl} \frac{\partial u_{0k}}{\partial x_l} \quad [3]$$

$$D_{ijkl} = \frac{1}{Y} \int_Y [E_{ijkl} - E_{ijpm} \frac{\partial \chi_p^{kl}}{\partial y_m}] dY \quad [4]$$

$$e_{0ij} = e_{ij}^{mac} - \frac{1}{2} \left(\frac{\partial \chi_i^{kl}}{\partial y_j} + \frac{\partial \chi_j^{kl}}{\partial y_i} \right) e_{kl}^{mac} \quad [5]$$

E_{ijkl} is the elastic tensor of the tissue, D_{ijkl} is the apparent elastic tensor, e_{0ij} is the micro strain and e_{ij}^{mac} is the macro strain. For more details about homogenization theory and its application see (2,6,15).

Methods: Two open cell shapes were used, a strut model and a spherical void model, as shown in Fig. 1. Cells were developed which matched trabecular bone morphology from different metaphyseal regions. A finite element analysis of equation [2] is used to solve for $\frac{\partial \chi_p^{kl}}{\partial y_m}$. These functions are then used to calculate the apparent elastic tensor in equation [4].

Comparison to Previous Micro-Model: A replica of the Beaupre and Hayes (1) spherical void model was analyzed by the homogenization method with the results $E = 1.38$ GPA, $G = 0.21$ GPA, and $\nu = 0.08$. Beaupre and Hayes predicted $E = 1.85$ GPA, $G = 0.45$ GPA, $\nu = 0.08$. These differences are consistent with the upper bounding characteristics of standard displacement based finite element methods.

Modulus Volume Fraction Dependence: For both models, volume fraction modulus relationships were determined over a range of angles with respect to the principal material direction. The 0° and 45° relationships are plotted in Fig. 2. These relationships change significantly depending on the direction about the principal material axes. This reflects experimental evidence that density modulus dependence varies with testing direction and anisotropy (14).

Comparison to Specific Bone Specimens: Finally, specific micro-models were derived to match the average morphology of specific specimens of trabecular bone. The tissue modulus was based on recent morphology of specific specimens and selected to be 5 GPA (9,12). The average of the three orthogonal moduli predicted from the homogenization analysis were compared to the average of the three experimental orthogonal moduli, as shown in Fig. 3. A fairly good agreement was found for these initial cells.

Conclusions: A new approach for the analysis of trabecular bone micro-mechanics has been developed using homogenization theory and micro-models based on trabecular bone morphology. These initial models provide insight into the dependence of trabecular bone stiffness on volume fraction. Clearly, more sophisticated cells must be developed which better reflect trabecular bone architecture. The use of homogenization theory based on morphological parameters is a promising technique for understanding trabecular bone micro-mechanics.



Fig. 1. Strut Model (left) and Spherical Void Model (right).

Metaphyseal Region	Experimental Modulus (MPa)	Predicted Modulus - Strut (MPa)	Percent Difference	Predicted Modulus - Sphere (MPa)	Percent Difference
Prox Tibia	103.0	184.3	+79		
Prox Tibia	177.7	237.7	+33		
Prox Tibia	153.3	97.8	-40		
Prox Tibia	370.7	295.8	-20		
Prox Tibia	212.7	463.6	+117		
Prox Hum	46.0	53.4	+16	94.8	+106
Prox Hum	286.7	294.7	+3	442.6	+54
Prox Hum	65.3	158.4	+142	189.8	+189
Prox Hum	293.3	297.1	+1	428.4	+46
Prox Hum	380.7	255.9	-33	395.8	+6
Iliac Crest	208.3	414.3	+98		
Iliac Crest	50.7	126.1	+148		
Lum Spine	59.0	271.2	+359		
Dist Femur	170.7	163.0	-8		
Dist Femur	304.0	417.8	+37		

Fig. 3. Predicted mean moduli from homogenization compared to experimental mean moduli.

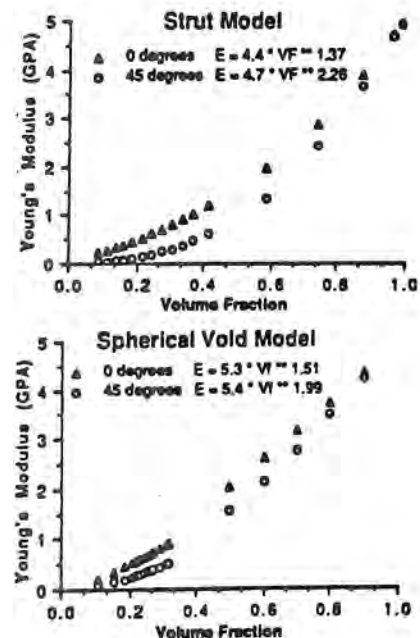


Fig. 2. Volume Fraction Modulus relations at 0° and 45° for the Strut Model (top) and the Spherical Void Model (bottom).

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TORSION AND REARFOOT STABILITY IN RUNNING

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1 Introduction

Excessive pronatory and supinatory movements of the rearfoot are known to be one of the major causes for pain and injuries in running (Segesser and Nigg, 1980, Clarke et al, 1984, Nigg et al, 1986). In order to control rearfoot motion, running shoes are supplied with a number of different functional elements to stabilize the rearfoot. From the mechanical point of view, however, the rearfoot movement depends not only on the shoe construction of the heel but also on the forefoot movement during ground contact and therefore on the connection between forefoot and rearfoot (torsional stiffness). The goal of this study was to measure the forefoot and the rearfoot movements while running a curve and to determine the relation between forefoot and rearfoot movements for a barefoot and a shoe condition.

2 Methods

Ten subjects were asked to run barefoot and with a common running shoe around a left and a right curve with a radius of 6.5m (with heel-toe contact). Hereby the ground contact of the left foot was filmed from behind at 100 Hz. The angular displacements of the forefoot (ϕ) and the rearfoot (γ) were analyzed. The relative movement between forefoot and rearfoot projected into the frontal plane was defined as torsion and calculated as follows: $\theta = \gamma - \phi - 90^\circ$ (figure 1).

3 Results

The results for the left foot while running the left curve demonstrated that the angular displacement of the forefoot was very similar for both the barefoot and the shoe condition (around 25°). The rearfoot movement, however, was different: In the barefoot condition an angular displacement towards pronation of about 10° was measured. With the shoe the angular displacement increased up to 25° . As a consequence the torsion angle was for the barefoot condition around 15° , but for the shoe condition it was close to 0° .

The results for the left foot while running the right curve were different. Now the subjects performed barefoot a torsional movement already before touchdown, such that neither the forefoot nor the rearfoot underwent any large angular displacement during ground contact. With the shoe this torsional movement before touchdown was not found so that immediately after touchdown forefoot and rearfoot were forced into supination (angular displacement of around 10°).

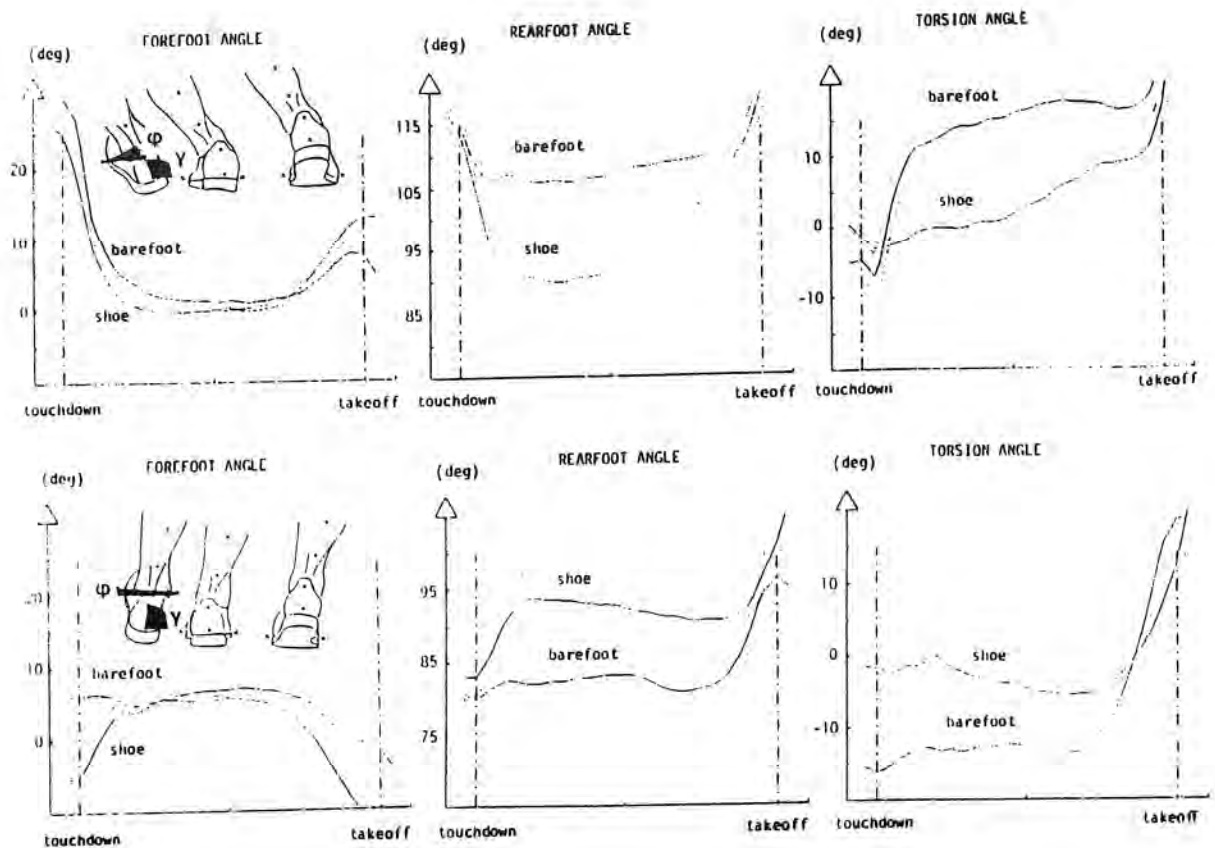


Figure 1: Angular displacements of the left foot while running a left curve (upper) and a right curve (lower). Statistics: normalized curves, $n=10$, standard error).

4 Discussion and conclusion

For both running conditions, left and right curve, the rearfoot pronated or supinated more with shoes compared to barefoot. The reason may be that the anatomical movement at the Chopart's and Lisfranc's joints (Kapandji, 1970) is reduced by the construction of the shoe. In other words, the high torsional stiffness of the shoe restrained the torsional movement of the foot. It is concluded that a sport shoe should be characterized by a lower torsional stiffness in order to allow the torsional movement of the foot and hence a reduction of excessive rearfoot movements.

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TECHNIQUES USED BY ELITE LONG JUMPERS IN PREPARATION FOR TAKEOFF

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During the final strides of the approach in the long jump, the athlete (a) adjusts the lengths of the remaining strides to correct for previous errors in striding (Lee, Lishman and Thomson, 1982; Hay, 1988); (b) adjusts the position of the body in preparation for takeoff (Nixdorf and Bruggeman, 1983); and (c) attempts to increase horizontal velocity to arrive at the board with the maximum that can be controlled during the takeoff (Hay, Miller and Canterna, 1986).

The purposes of this study were: (a) to obtain descriptive data on selected characteristics of the positions adopted by elite female and male long jumpers during the final strides of the approach, and (b) to determine the relationships among these selected characteristics and the horizontal velocity of the athlete's G at touchdown, the vertical velocity of the athlete's G at takeoff and the official distance of the jump.

METHODS

Trials by 26 female and 20 male long jumpers were selected for analysis. Parameters describing the location of G relative to the toe of the support foot were determined for touchdown (TD), mid-support (MS) and takeoff (TO) for each of the last four support phases of the approach — Figure 1.

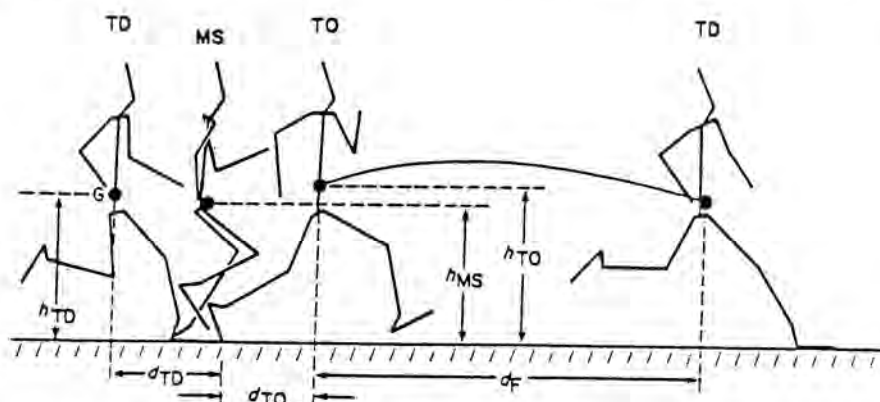


Figure 1: Parameters used to characterize position of subject

RESULTS

Heights of Center of Gravity. There were no significant differences among the mean values of h_{TD} , h_{MS} or h_{TO} for the support phases of the

fourth-last, third-last and second-last strides -- Figure 2. With one exception, these heights were significantly less for the support phase of the last stride.

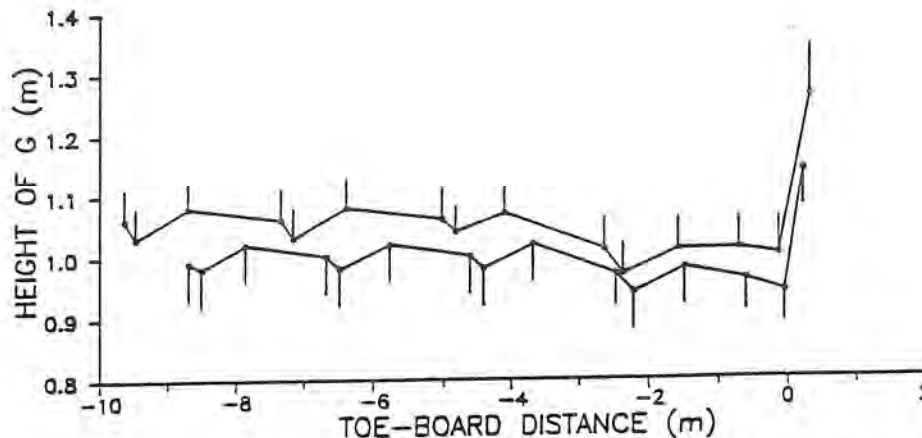


Figure 2. Means and S.D. for h_{TD} , h_{MS} and h_{TO} for last five support phases for males (top) and females (bottom).

Touchdown, Takeoff and Flight Distances. There were no significant differences among the mean values of d_{TD} , d_{TO} or d_F for the fourth-last third-last and second-last strides. d_{TD} for the last stride was significantly greater in magnitude than for the previous three support phases; and d_{TD} for the jump was significantly greater than for the previous four support phases. d_F for the last stride was significantly shorter than for any of the previous four support phases.

Position Variables and Distance. Three of the position variables -- $d_{TO(4L)}$, $d_{TD(J)}$ and $h_{TO(J)}$ -- were significantly correlated with the distance of the jump (stature partialled out). These results indicated that the greater $d_{TO(4L)}$, the greater $d_{TD(J)}$ and the greater $h_{TO(J)}$, the greater the distance of the jump.

Position Variables, Velocity and Distance. These position variables were significantly related to the distance of the jump by virtue of their relationships with the horizontal velocity of the approach and/or the vertical velocity at takeoff into the jump. Considered alone, none were influential in determining the distance of the jump.

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**DETERMINATION OF THE OPTIMUM CUTOFF FREQUENCY IN
THE DIGITAL FILTER DATA SMOOTHING PROCEDURE**

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INTRODUCTION

Various mathematical techniques have been employed to smooth raw data to reduce the effect of filming and digitizing errors on estimates of velocity and acceleration. Winter (1974) advocated a second-order recursive digital filter procedure for smoothing film analysis data. Although this digital filter has been suggested as the best among the data smoothing techniques used in sport biomechanics studies (Li and Yu, 1983) and is widely used in biomechanics studies at present, the problem of determination of the cutoff frequency has not been solved. A wrong determination of the cutoff frequency can result in grossly over- or under-smoothed velocity and acceleration estimates. The purpose of this study was to develop a method for objective determination of the cutoff frequency of the digital filter.

METHOD

Vertical coordinate-time data for a freely falling body were used as the standard data. Different sets of computer generated random numbers were used as random errors and added to the standard data to generate different sets of artificial raw data. The mean absolute value of each set of random errors was controlled.

The artificial raw data were smoothed by the digital filter at different cutoff frequencies. The filtered outputs were evaluated by the error energy between the filtered data and standard data to determine the optimum cutoff frequency for each set of raw data.

Harmonic analysis was then used to analyze the characteristics of the signal and error for each set of raw data. In the harmonic analysis, the highest harmonic of the raw data was defined as the harmonic at which the raw data were best fitted by Fourier series. Harmonic amplitudes were normalized as percentages of the fundamental harmonic amplitude of each set of raw data. Based on the results of the harmonic analysis, independent variables were selected and added to sampling frequency to develop regression models.

RESULTS AND DISCUSSION

Two regression equations were developed. The optimum cutoff frequency (F_c) for a set of raw data can be estimated either from the sampling frequency (F_s) and the normalized harmonic amplitude before the overlap harmonic (NHA) using the relationship:

$$F_c = (0.1358 F_s^{1/2} + 1.6401 NHA^{-1/8} - 1.0432)^2$$

or from only the sampling frequency using the following formula:

$$F_c = (1.4845 + 0.1532 F_s^{1/2})^2$$

These two regression equations were used to determine the cutoff frequency for angular displacement data provided by Pezzack (1977). The cutoff frequencies of this set of data estimated from these two equations were 8.61 Hz and 6.6 Hz, respectively. The cutoff frequency determined by the first equation was very close to the 9 Hz used by Pezzack (1977). In general, the filtered outputs with the cutoff frequencies determined by either of the two equations provided accurate estimates of the true angular acceleration function. There were, however, some over-filtering effects on the peaks of the estimated acceleration curves.

The findings of this study were: (1) the optimum cutoff frequency of digital filter is influenced by the sampling frequency. Over 77% of the variation of the square root of the optimum cutoff frequency is due to the sample frequency; (2) the normalized harmonic amplitude before the overlap harmonic can be used as an indicator of the characteristics of the error signal. About 9% of the variation of the square root of the optimum cutoff frequency can be explained by the normalized harmonic amplitude before the overlap harmonic.

CONCLUSIONS

The results of statistical analysis and application of the regression equations have demonstrated that the optimum cutoff frequency of the digital filter can be estimated from the sampling frequency and the normalized harmonic amplitude of the signal before the overlap harmonic. Further studies are recommended to provide more complete explanation of the variation in the optimum frequency in digital filter procedure.

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Effects of foot orthotics upon the ankle and knee mechanical alignment

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Introduction

Although a number of anatomical factors must be considered in the diagnosis and evaluation of lower extremities, it seems that many problems of the ankle and knee joints are related to the foot structures and their function during the support phase of the gait cycle (Bates et al., 1979). This would suggest that there is a complex interaction between lower extremity injuries and anatomical factors, especially considering the importance of the lower limbs kinetic chain. Particularly, excessive foot pronation has been speculated to be a cause of ankle and knee problems (Hunt, 1985).

Often, foot orthotic devices are utilized to correct or prevent certain lower extremity mechanical misalignments (Bates et al., 1979). Specifically, a medial wedge, placed under the heel in the inner sole, can correct the excessive pronation of the subtalar joint. Hunt (1985) demonstrated the importance of the medial wedge in the correction of the knee valgum, measured by the Q-angle defined as the knee angle in the frontal plane. The Q-angle seems then to be an important kinematic reference in the evaluation of knee syndromes, such as patellofemoral syndrome (PFS). Furthermore, it has been demonstrated that a 2-D kinematic analysis, provided by an automated video analysis system, can provide valuable information on the effects of orthotic devices during static and dynamic activities (Cairns et al., 1988).

Therefore, the purpose of this study was to further assess the effects of foot orthotic devices on selected aspects of the foot and leg walking mechanics, particularly the pronation angle of the subtalar joint and the knee valgum, or Q-angle, by examining a small group of subject jogging on a regular basis. This study also allowed to validate the use of an automated 2-D video analysis system in the evaluation of orthotic device effects on ankle and knee joints mechanical alignment.

Methodology

Five male subject, jogging on a regular basis and asymptomatic to lower extremity problems, participated in this study. All subjects had to first stand and then walk on a treadmill at a constant speed identical for all (1.1 m/s). While standing or walking on the treadmill, subjects were filmed using two CCD video cameras connected to an automated 2-D video analysis system (Motion Analysis Corp.). Pronation (posterior view) and Q (anterior view) angles were derived by securing reflective tape markers (2,5 cm in diameter) on anatomical landmarks (midline of the shank, achilles' tendon, and heel, anterior superior iliac spine, patella, and tuberosity of the tibia). After videotaping the subjects, the video signal was processed through a video processor (VP-110, Motion Analysis Corp.) in order to derive the outlines of the six markers. These outlines were digitized (60Hz) using a Sun microsystem (Sun 2/120). The ExpertVision™ software was used to compute the centroids from which path data were obtained. A Tukey window algorithm was utilized to smooth the path data after which the bearings and angular displacement were calculated. Each trial consisted of 0.66 s of data (40 frames) representing the support phase during the dynamic trials.

Three conditions were examined: (1) without medial wedge, (2) with a 1/8 inch medial cork wedge, and (3) with a 1/4 inch medial cork wedge. Four selected parameters were quantified for both legs under all conditions during static and dynamic trials: static pronation angle (SP = mean over .66 s static stance), peak dynamic pronation angle (PDP), static Q-angle (SQ =

mean over .66 s static stance) and peak dynamic Q-angle (PDQ). All subjects wore the same shoe model during testing (Asics Tiger Ultimate II).

Results

Mean and standard deviation for all parameters are presented in Table 1.

TABLE 1

Mean (and standard deviation) values of all parameters (Parm) measured during the support phase under all conditions for both legs.

	Right leg			Left leg		
Parm	no wedge	1/8 in	1/4 in	no wedge	1/8 in	1/4 in
SP	6.22 (4.26)	6.32 (3.39)	3.50 (3.69)	5.96 (2.42)	5.44 (0.94)	3.14 (1.96)
PDP	8.70 (2.67)	8.23 (3.61)	6.35 (2.21)	8.71 (1.18)	7.47 (0.86)	6.50 (1.56)
SQ	10.03 (10.12)	6.72 (7.37)	6.91 (8.50)	8.72 (7.42)	6.52 (5.75)	6.95 (8.40)
PDQ	10.39 (8.51)	11.30 (6.21)	10.84 (7.58)	9.37 (6.29)	9.58 (6.34)	9.97 (6.70)

Interpretation and Discussion

The results indicate that the pronation angle, in static and dynamic trials, decreases with the application of a medial wedge increasing in thickness. The Q-angle decreased in static trials with the application of a wedge compared to the no wedge condition, whereas, there is no Q-angle differences between conditions in the dynamic trials. That could be explained by the very short heel ground contact period during gait. These results also suggest that a medial wedge only, placed as an orthotic device, does not seem to be sufficient to promote an effect on the knee valgum, especially in dynamic trials, and that mid- or fore-foot corrections must be added. The pronation angle, as was expected, decreases linearly with the application of thicker wedges. Therefore, correcting for over-pronation might not always result in Q-angle adjustments and, thus, if one wishes to modify the lower limb mechanical alignment, both the pronation and Q angles should be evaluated dynamically, and not only statically.

Finally, it appears that the utilisation of an automated 2-D video analysis system is an efficient mean to analyse and quantify slight angle variations of the lower limb joints. Hence, the utilization of such a kinematic analysis system for the evaluation of mechanical alignment of the ankle and knee is warranted, and could be a valuable tool in the diagnosis of pathological syndromes such as PFS.

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THE DEVELOPMENT OF A COMPUTER SYSTEM FOR STUDYING
CONSECUTIVE VERTICAL JUMPS.

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The Jump Platform PS-65, an accurate electronic apparatus was developed for measuring the height of vertical jumps by the flight time registered. It consists on a aluminum frame with infra-red lights built in. The electronic unit captures the pulses when the lights are interrupted or not and sends the information to an Apple computer that calculates the height of the jumps. A maximum of 512 vertical jumps can be performed inside the platform and a graphic showing a curve of number of the jump and height is displayed immediately.

A sample of 92 male physical education students was tested. It was applied two single jump tests (counter-movement jump and static jump) and the consecutive maximal vertical jump test during one minute. Anthropometrical data were taken. In all tests, it was not permitted the movement of the arms by keeping the hands fixed on the waist.

A correlation matrix was done with the 19 variables studied and a multiple analysis test was then applied to the 9 most significant variables, considering the inclination degree of the n° of the jump and height curve as the dependent variable.

The results showed three distinct groups :

- The group A presented a lower inclination of the curve (4.81 SD-3.05 degrees) representing a maintenance of the performance during the one minute test. Also, this group had regular and high results for the two single jump tests and elastic index and performed more jumps during the one minute test (81 SD-11 jumps). This group is formed by those subjects with high level of anaerobic endurance.
- The group C presented a greater inclination of the curve (30.09 SD-5.16 degrees) representing a decrease of the jump performance during the one minute test. Also, this group had higher results for the two single jump tests and elastic index and performed less jumps during the one minute test (44 SD-6 jumps). This group refers to subjects with a high level of explosive power.
- The group B showed intermediate results for all variables.

The results were ratified when national teams were tested. The sprint runners and high jumpers showed a typical curve as the group C. The long runners presented results as the group A. The absolute results were all higher than the sample studied as it was composed by students without any specific training program. The volleyball team presented an interesting result. Although all athletes were training with the same program, different shapes of curves were observed leading the coach to realize which athletes were more conditioned for explosive power and those that were more prepared for the maintenance of the jump movement (in case of "rally" for example).

For concluding, the Jump Platform PS-65 showed to be an accurate instrument for measuring the height of vertical jumps. The validation was done with filming procedures showing no significant difference at a 99.5% level. Although a force plate can be used for that purpose, the Jump Platform is less expensive and can be easily moved.

The study of vertical jumps during a period of time revealed to be an acceptable method for the evaluation of explosive power and anaerobic endurance in terms of jump performance.

For future studies, the Jump Platform PS-65 is now being improved for registering the vertical forces. This way the study can be extended for the kinetics of the vertical jump and its relation to the shape of the jump height curve.

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Kinematic and Temporal Characteristics Applied to Success:
Goals in Field Hockey

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Introduction

According to the current playing rules a goal can be scored only if struck from within the striking circle. A potential goal situation could occur directly from game play or from a penalty corner, long hit, or penalty stroke "dead ball" situation. Should goals be occurring systematically from specific game play or dead ball situations? What are the specific characteristics of a successful situation? This study applied kinematic and temporal parameters to technique, tactics, and game strategy. Success may be due to efficiency or may be the result of some other situational factors such as greater force or greater accuracy. Therefore, a look into the variables associated with success, taken from film data collected during actual game and tournament play at the highest (Olympic) level, was considered to be of benefit to researchers and individuals involved in the game of field hockey and perhaps in other sports in the future.

Materials and Methods

Results of this investigation were based on situational analysis of seven hundred and forty-four tape recorded circle play situations. One hundred and two circle play instances were randomly selected from the data and analyzed temporally and kinematically. The mechanical analysis was made using the method of cinematography. Two motor-driven 16 mm cameras (Photosonics 121, 100 frames/s verified by camera-internal timing-light pulses) equipped with 12 millimeter lenses were used for the mechanical analysis (Western Michigan University). Estimated distance, (time), and direction-angles were computed using the x, y coordinate system. The coordinates were obtained from paired films through the use of a pin-registered projector (Vanguard M160) and digitizing system (Numonics 1220). All digitizing and data collection was made possible through the USOC Biomechanics Laboratory. Distribution, variability, and central tendency scores of situational circle play occurrences and temporal and kinematic variables were calculated relative to four phases of potential success (a goal); these were (1) action into the circle, (2) offensive results, (3) offensive results: follow-up play, and (4) defensive maneuvers. Rank order correlation coefficients were calculated to summarize the strength of associations and relationships between variables. A regression analysis and significance at the 0.01 level were computed to determine the influence on goals scored as opposed to "chance" goals.

Results Applied

Situational circle play and goals scored related to level of significance involved 447 cases, men, and 297 cases, women. Of the 744 cases recorded, 50 goals were scored (6.7 per cent). Initial offensive play for men resulted in 21 goals, and 8 goals resulted from follow-up play; for a total of 29 goals (6.5 per cent). Success in goal scoring occurred 21 times for the women (7.0 per cent of the time); this included 15 goals on initial offensive play options and 6 goals on follow-up offensive play. Goals scored initially by means of the penalty corner was highly significant for women. The penalty stroke situation, a high -potential scoring situation, occurred 3 times for men and three times for women. The men tallied one goal and the women zero goals. No significance was recorded. A direct correlation coefficient of 0.92 was recorded between penalty corner action into the circle and offensive reception for both men and women. Offensive reception showed no correlation to goals

scored initially or on follow-up shots. The men demonstrated a level of significance when play continued in the circle with carry and pass options within the circle. Secondary shot/goal attempts showed no level of significance. Defensive play, tackling or interceptions, showed no significance. The game of field hockey continues to be a game of chance with three exceptions: (1) goals were scored by the women off the penalty corner, (2) the reception on the penalty corner had a high correlation for men and women, and (3) follow-up carry and pass play options for the men were significant.

Technique Data showed that the men mechanically moved their center of gravity into the penalty stroke (flick); the women executed the stroke moving away from the desired line of flight. Defensive techniques appeared to be more consistent and less variable, relative to direction, velocity, and versatility. Defensive players and goalkeepers did not indicate a weak side as the variables were reviewed. Offensive players demonstrated excellent speed but were lacked maneuverability or change of direction. On the penalty corner the data demonstrated inability to accurately push the ball out and receive the ball. The hit out needs to be executed within 1.0 seconds and received in 0.45 seconds; for a goal to be scored, the entire penalty corner sequence must be executed in less than 2.0 seconds (Klatt, 1960). Shot/goal velocity traveled a maximum of 31.6 meters/second; an average time for the shot was 0.4 seconds.

Tactics Second offensive players were available in support of the ball carrier on carry-in and pass-in play but were not utilized. Variables indicated that the ball carrier was pressured. Space and time appeared to be of a high premium. On carry-in play from the offensive top left side of the circle showed defenders marking closely; as play developed down the right side, the defenders and goalkeeper interchanged responsibilities. Carry-into the circle (continuous ball control) resulted in "manufactured" fouls. A second carry by the initial ball carrier demonstrated 112.45 degrees change in direction back into the marking defender (result: penalty corner). A second play revealed that the ball carrier played the ball in one direction and ran directly into the defender (result: good "no call"). Time and direction is essential in circle play off the pass-in situation. Time means space; the strength and speed of today's players seemed to shrink the space inside the circle. Second offensive players or support players were available minimally.

Game Strategy Goals were scored on initial play into the circle, (generally, a "fast break" situation); once play established itself in the circle, a dead ball situation usually occurred. It appeared as though play needed to be re-developed OUTSIDE the circle (as might occur in basketball). The inability to develop goals from field play was demonstrated through ineffective technique, tactics, and strategy. The size of the circle restricted development of offensive play; success in team maneuvers within the circle was further limited by the speed and range of the defensive players. The lack of field play scoring seemed to encourage teams to force dead ball situations. If this continues to be true, penalty corner "teams" will need to develop more creative strategy as well as effective individual technique. Consideration may need to be given to increasing the size of the circle, goal cage, or both, for more success.

ENERGY AND POWER CONSIDERATIONS IN SLOW WALKING.

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In recent years mechanical energy and joint power analyses have been reported in the assessment and treatment of pathological gait^{1,2}. A major problem in exploiting these methods has been the lack of normal data relating to the very slow speeds that are characteristic of many pathologies. The purposes of this study were to calculate the mechanical energy costs of walking for normal older subjects walking at slow speeds and to determine if energy conservation was linked to the speed of walking. Further, the investigators wished to determine if the power patterns accomplishing the work of walking were different from those previously reported for normal speeds.

METHOD

Eight subjects between the ages of 54 and 69 years were prepared with reflective markers over major joints and certain other body landmarks. Each was trained to walk over a force platform while three trials at each of three speeds were recorded on cine film: very slow (0.36 m/s), slow (0.62 m/s) and free speed (1.1 m/s). A 2-dimensional link segment model was used in the biomechanical analyses, performed according to the methods of Winter³ to yield mechanical energies and powers. The product of the net joint moments and angular velocities gave joint powers and the integrals of the positive and negative portions of the curves yielded work values. Net powers for each limb were calculated as the sum of the ankle, knee and hip over the cycle. Left leg data were approximated by shifting the point of initial contact to coincide with the instant of heel strike of the opposite side. A "combined" curve estimating the total power of the two sides of the body was obtained by adding the two together.

RESULTS AND DISCUSSION

The mechanical energy costs varied greatly with the speed. The data were best fit by the following equation:

$$E_m = 0.929 - 0.485 V + 0.674 V^2$$

where E_m was normalized energy cost per meter of distance walked (J/kg.m) and V was velocity (m/s). The proportion of energy conservation increased from 54% at very slow speeds to 65% at free speed, attributable to increases in the between-segment energy exchanges from 15% at very slow speeds to 25% at free speeds. Surprisingly there were no significant variations in within-segment conservation.

The major bursts of positive and negative work were of the same general pattern as those previously reported.⁵ The ankle plantarflexors produced 48% and 53% of the positive work at

very slow and slow speeds respectively, in contrast to normal estimates of 66% calculated from previous reports.⁴ The hip produced 37% and 38% at these speeds. The sum of ankle, knee and hip curves is shown in Fig. 1. Events 5 and 7 were the result of the ankle and hip generations while 3 was caused by knee extensors in stance. Events 9 and 1 were due to hip extensors late in swing and early in stance. The curves combining both sides of the body (Fig. 2) were noticeably different at different speeds. At slow speeds four generative bursts per cycle were evident; two were the result of event 5 augmented by events 9 and 1 and the other two were from 7 augmented by 3 of the other side. During very slow walking only the two major positive bursts were evident, and they occur later in the cycle. These results provide a small normal base with which to compare results of biomechanical analyses of subjects with motor pathologies.

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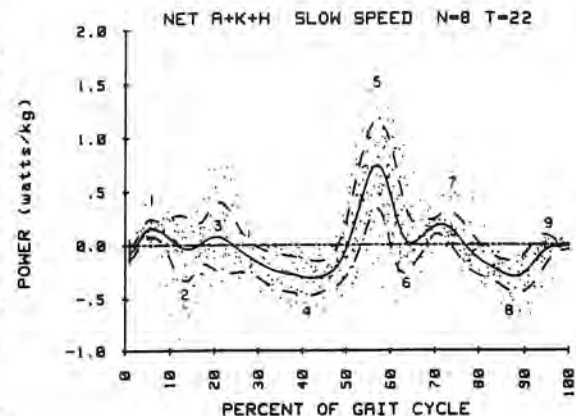


Fig.1. Net power curve of one limb for slow walking: sum of ankle, knee and hip powers. See text.

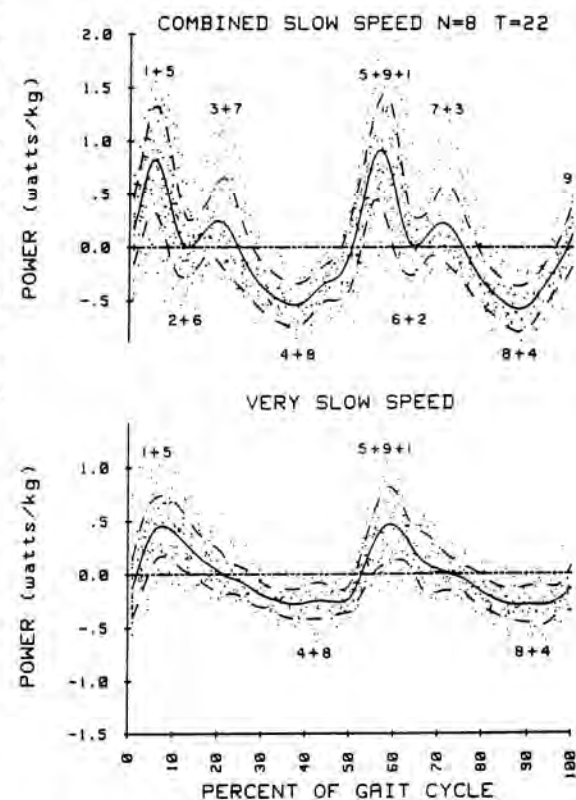


Fig.2. Combined power curves: sum of net power curves of both sides of the body for slow and very slow walking. Note absence of effects of event 3+7 at very slow speeds.

Technique Comparisons Between Hard and Soft Landings of Young Female Gymnasts

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In 1986, the United States Gymnastic Federation (USGF) registered more than 130,000 female gymnasts. Women's gymnastics has become one of the fastest growing sports in the United States with more children participating in a variety of programs at earlier ages. Increased participation leads to greater exposure of athletes to potential injury (McAuley et al., 1987). Injuries caused by single impact trauma as well as those caused by overuse are increasing in frequency among young children involved in organized sports (Micheli, 1983). Repetitive microtrauma in a sport with frequent high-impact loading conditions on the skeletal system may increase the chances of injury or disturb normal growth (Caine & Lindner, 1985). Because of the limited available research on the technique of landing impacts, this investigation was designed to determine if the landing technique of young female gymnasts could be altered to reduce peak impact forces encountered and therefore possibly reduce the chances for potential injury.

Methodology

The subjects, who volunteered for this investigation were 30 female gymnasts (age = 123.0 ± 11.8 months; ht = 139.0 ± 9.6 cm; mass = 34.2 ± 12.4 kg) attending the Texas Woman's University gymnastic camp in Denton, Texas during the summer of 1988. A gymnastic parallel bar, suspended between two volleyball standards equipped with adjustable T-bars, was positioned above a Kistler forceplate. Each subject performed a landing pretest from a height of 25 and 30 cm, followed by a period of practice, and a posttest at each height. All subjects were tested without shoes. Film records were obtained with a 16mm Locam high-speed camera operating at 100 frames per second. Each subject was filmed from the sagittal view of initial rest in the hanging position to a final stable position after landing. Seven segmental endpoints were digitized and then analyzed with software written by Richards and Wilkerson (1984). The raw data were smoothed with a second-order low-pass digital filter set at 6 Hz (Winter, 1982). Force data were sampled at 600 Hz.

Results

Peak vertical forces were expressed in raw data form (N) or as a ratio to body mass. The means and standard deviations for the pretest and posttest are presented in Table 1. A two-way analysis of variance (ANOVA) with repeated measures revealed significant differences between pre- and posttests at both heights for peak vertical forces ($F(1,116) = 98.26, p < .05$) and for body weights ($F(1,116) = 113.98, p < .05$). Subjects decreased their peak vertical forces from pre- to posttest when a session of several landings was practiced between tests. Instructions were given to land softer after the pretest. The computer served as a feedback instrument to display landing forces to each subject. A two-way ANOVA with repeated measures found a significant difference between conditions for impulse ($F(1,116) = 11.45, p < .05$). The impact force curve was divided into two time periods for analysis. Initial impact began at touchdown and proceeded until reaching peak force. Force absorption started at peak impact and ended when stability was regained. No significant differences were found between total time, initial impact time, and force absorption time. Because peak vertical forces decreased but temporal values remained relatively unchanged between pre- and posttests, the force curves between conditions were examined to determine the significance of impulse. Pretest force curves displayed one or two definite peaks, while posttest force curves revealed an absence of peaks with equal distribution. This difference between force curves could indicate that impulse is more evenly distributed during a softer landing. During kinematic analysis two distinct patterns were revealed within each subject. Simultaneous use of the segments (foot, shank, thigh, and trunk) in a short period of time for the pre- and posttests were exhibited during the initial impact, or compression stage. During force absorption, resistive flexion stage, sequential use of the segments over a longer period of time was displayed.

A two-way ANOVA with repeated measures revealed significant differences between total time of the initial impact and force absorption stages ($E(1,116) = 19.67, p < .05$) and within the pre- and posttests of each stage ($E(1,116) = 63.57, p < .05$). Additional analyses revealed a significant difference of trunk time ($E(1,116) = 31.71, p < .05$) between initial impact and force absorption stages and within the pre- and posttests of each stage ($E(1,116) = 13.08, p < .05$). Increased trunk flexion during the posttest of the initial impact stage resulted in longer time periods and possibly aided in the reduction of peak vertical forces.

Table 1. Peak Vertical Forces (N) and Ratio to Body Mass

	<u>Mean</u>	<u>SD</u>	<u>Mean</u>	<u>SD</u>
Pre 25 cm	1868.9	603.6	12.75	3.55
Post 25 cm	908.9	208.9	6.43	1.96
Pre 30 cm	1550.3	525.8	10.60	2.92
Post 30 cm	963.7	214.3	6.69	1.60

Discussion and Conclusions

It was concluded that the landing technique of young female gymnasts could be altered to reduce peak impact forces encountered. The subjects in this study significantly reduced their peak vertical forces after several soft landing trials. Two distinct stages, initial impact and force absorption, were apparent within each landing trial. When initial impact time was short and a large peak or peaks appeared in the force curve, large peak forces were experienced by the subject. These large peak forces occurred when the subject landed in a knee-locked position with a hyperextended back. When subjects were instructed to land softer, they flexed forward at the trunk with a flat back, increased flexion at the knee and hip, and spent a longer time in the force absorption stage.

During the initial impact stage subjects exhibited simultaneous use of the lower body segments. A sequential pattern of segment use was evident in the force absorption stage. Mizrahi and Susak (1982) suggested that pre-programming muscle action and joint motion is important when reflex activity is absent. Softer landings, shown by reduced peak forces, were achieved when subjects pre-programmed their muscle action and joint motion prior to landing. In training young gymnasts, due to their stage of growth, it is important to emphasize flexion of the joints and bending forward of the trunk. Landing in a knee-locked position with a hyperextended back could possibly lead to a variety of short and long term injuries.

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ELASTIC ENERGY IN ATHLETIC SHOE CUSHIONING SYSTEMS

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Introduction

Recent experiments on the viscoelastic properties of human feet suggest that the arch of the foot acts as an elastic energy conserving system¹. These findings have been interpreted as meaning that athletic shoes should be similarly designed² and have excited popular interest in the concept of elastic energy storage in athletic footwear³. While it has been predicted that elastic energy storage and recovery in a shoe may enhance performance, there is no experimental evidence to support this claim and the actual levels of energy exchange in athletic shoes have not been previously determined. The purpose of this study was to determine the response of viscoelastic athletic shoe cushioning systems to the loads imposed during running and vertical jumping and hence determine the amount of strain energy stored in the shoe during these activities.

Methods

The shoe cushioning system was modelled using a multiple element approach, with each element having an area of 0.5 cm² and a thickness determined from measurements made on real running and basketball shoes. The nonlinear load-displacement behaviour of the cushioning material was modelled by the differential equation:

$$F = \lambda(x) \cdot x + \eta(x) \cdot \frac{dx}{dt} \quad (1)$$

where F is the normal stress applied to an element, x is the strain and the stiffness, $\lambda(x)$, and damping, $\eta(x)$, are given by:

$$\lambda(x) = \lambda_0 + \lambda_1 x \quad (2) \quad \text{and} \quad \eta(x) = \eta_0 + \eta_1 x \quad (3)$$

The coefficients λ_0 , λ_1 , η_0 and η_1 were determined experimentally by mechanical testing of sole samples taken from running and basketball shoes of the kind used by competitive athletes. Elemental force-time histories were determined by cubic spline interpolation of plantar pressure distributions sampled at 70 Hz while a 76.0 kg male subject ran at 3.8 ms⁻¹ and performed maximum-effort vertical jumps. For each element, equation 1 was solved numerically for $x(t)$ using a 4th order Runge-Kutta iterative method with a time step of 1.43 ms (1/700th s). At each step, the vertical compression of each element, work done and strain energy stored in the elastic component were calculated.

Results

The peak deflections of the model cushioning systems under the applied loads of running and jumping are shown in Figure 1 together with the distribution of peak forces, work done on the midsole during compression, work done by the midsole during recovery and peak strain energy. Work and energy totals are shown in Table 1. Energy exchange in the midsole model was greater during a running stride simulation than during a vertical jump. The applied forces in the vertical jump were lower than during running and concentrated in the forefoot of the shoe where the midsole is thinner and less compliant. This resulted in less work being done to deflect the stiffer midsole and reduced strain energy storage. The midsoles dissipated 4.8 J during the running step and 3.7 J during the jump.

MODEL	Running	Basketball
No SHOES	1	2
LOAD REGIME	3.8 ms ⁻¹ run	Vertical jump
Total strain energy	9.6 J	8.6 J
Peak strain energy	5.0 J	3.8 J
Work done on the midsole	12.3 J	9.7 J
Work done by the midsole	7.8 J	6.0 J
Energy dissipated	4.5 J	3.7 J
% Energy loss	37%	38%

Table 1: Work and Energy Totals.

Although viscoelastic cushioning systems are net energy dissipaters, they do store a small quantity of strain energy when compressed. This means that the midsole is capable of doing some work during elastic recovery from peak deflection. In the simulations described here, a total of 9.6 J were stored in the running shoe system and 8.6 J in the pair of basketball shoes, of which 7.8 J and 6.0 J, respectively, were recovered. These figures represent totals for all elements over the whole simulation, however. The highest levels of strain energy recorded at any one time were 5.0 J and 3.8 J respectively.

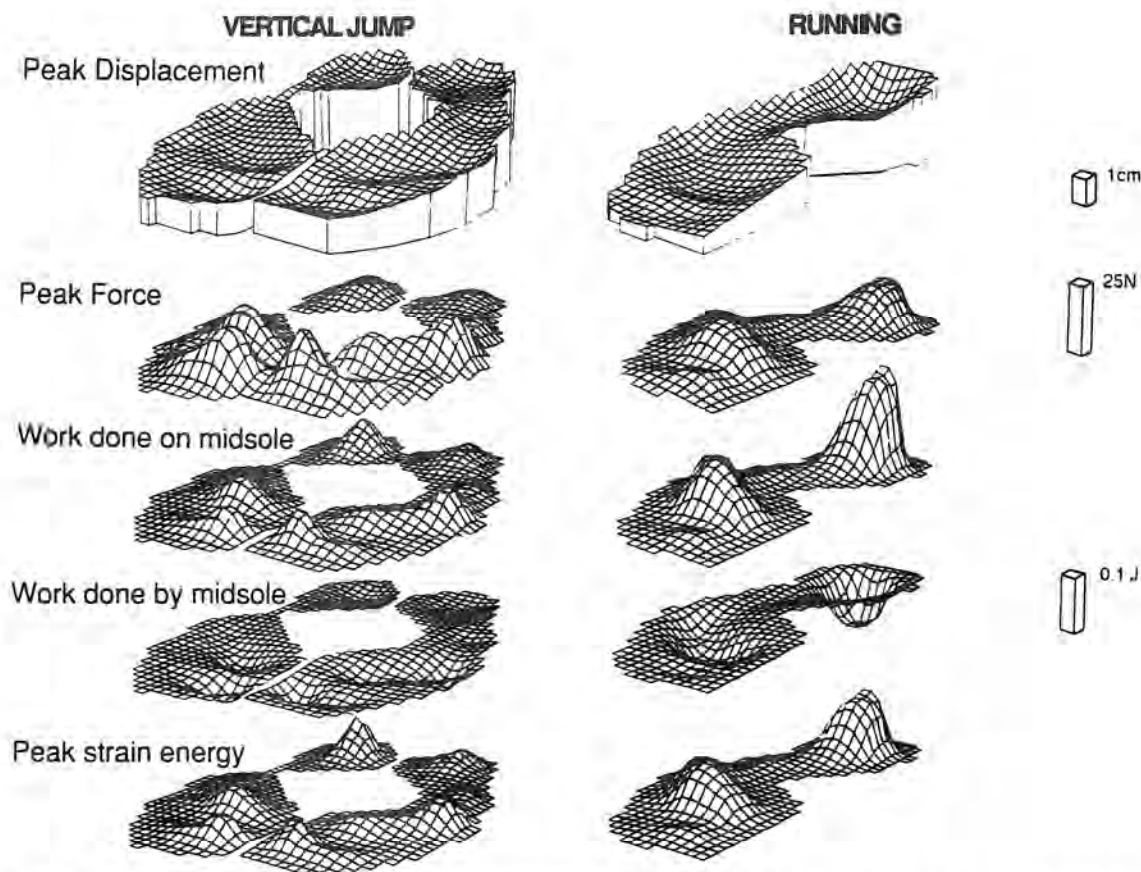


Figure 1: Peak displacement, force, work and strain energy during a running step (one shoe) and a vertical jump (two shoes). Graphs show peak values of individual elements from the whole movement. Element area is 0.5 cm^2 . Vertical scale of displacement graphs is exaggerated.

Discussion

Energy storage and recovery in the model cushioning systems was low compared with other passive energy transfer mechanisms that occur in human motion. Ker et al (1987)¹ report estimates of 35 J stored in the Achilles tendon and 17 J in the arch of the foot during a running step. Passive energy exchange between and within segments is approximately 430 J per step when running at 3.8 ms^{-1} and changes in the net energy of the body are approximately 160 J per step⁴. The net energy losses of approximately 37% reported here are representative of the most resilient athletic shoes currently available. During mechanical impact testing of real shoes⁵, energy losses typically range between 35% and 60%. Optimistically, if all the energy recovered from the basketball shoe model was converted to potential energy, the difference in jump height between the the best and worst shoes would amount to approximately 3 mm. Similarly, the energy recovered from the most elastic running shoe midsole would equate to a reduction in oxygen uptake of approximately $0.3 \text{ ml O}_2 \text{ kg}^{-1} \text{ min}^{-1}$.

The results of this study suggest that the lack of experimental evidence to support the theory that elastic energy storage and recovery in athletic shoes can enhance athletic performance may be due to the relatively small quantities of energy recovered from typical cushioning systems and the minimal effects of such energy exchange on body dynamics.

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PLANTAR PRESSURES DURING BAREFOOT WALKING

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INTRODUCTION

The distribution of load on the plantar surface of the foot has implications for the diagnosis of some foot pathologies¹, for footwear design, and for the study of lower extremity function. Despite pioneering studies of plantar pressure distributions during walking², running³ and falling⁴, only those during barefoot standing have been extensively documented¹. More studies of large groups of healthy subjects are required to establish "normal" pressure distribution patterns during gait. The purpose of this study was to establish baselines for the effects of walking speed on barefoot plantar pressure distributions, and to examine possible differences between male and female subjects.

METHODS

Sixty five male subjects ($78.87 \text{ kg} \pm 9.02 \text{ sd}$, $1.791 \text{ m} \pm 0.070 \text{ sd}$) and 66 females ($63.20 \text{ kg} \pm 10.80 \text{ sd}$, $1.654 \text{ m} \pm 0.063 \text{ sd}$) were studied while walking barefoot at 1.0, 1.5 and 2.0 ms^{-1} . Following subject familiarisation, plantar pressure distributions under the right foot were recorded using a 23 x 44 cm capacitance mat transducer system (EMED-SF, Novel GmbH) mounted in a laboratory walkway. The sensor mat incorporated 2016 force transducers in a 32 x 63 matrix with a density of 2 sensors cm^{-2} and was sampled at 70 Hz. For each subject, a typical trial at each speed was selected for analysis of the peak pressures recorded by each transducer.

Foot size, shape and placement on the transducer mat vary considerably between subjects. A shape transformation algorithm was developed which enabled pressure distributions from different subjects to be normalised to a standard foot shape, size and orientation. For normalisation, bicubic spline interpolation was used to estimate pressures at equidistant points on a polar reference frame covering the plantar surface of the standard foot shape. The polar reference frame was also used to determine the locations of discrete regional pressure peaks from raw pressure data.

RESULTS

Figure 1a shows the distribution of peak pressures on the plantar surface of the foot, averaged across all subjects at each speed. Mean values of regional peak pressures at the three walking speeds studied are shown in Table 1. Table 2 records the relative locations of heel and forefoot pressure peaks. Between-subject variability itself varied, the greatest coefficients of variation between subjects occurring nearer the borders of the plantar contact surface (Figure 1b). In general, the magnitude of regional peak pressures increased significantly with increasing walking speed. However, while peak pressure under the first metatarsalphalangeal joint (MTPJ) increased with speed, peak pressures under the 2nd and 3rd MTPJs did not increase between 1.5 and 2.0 ms^{-1} and peaks under the 4th and 5th MTPJs decreased across the three speeds. The shift in weight towards the medial side of the foot is confirmed by measurements of the location of the forefoot pressure peak (Table 2) which moved medially by a small but statistically significant ($P < 0.05$) amount as speed increased. No correlation was found between body mass and peak heel pressure at any walking speed. Significant

Table 1:
Peak plantar pressure (kPa) by region (mean $\pm$ sd)

Speed ms^{-1}	1.01 $\pm$.03	1.51 $\pm$.04	1.99 $\pm$.05
n	129	131	130
Heel	367 $\pm$ 115	517 $\pm$ 158	734 $\pm$ 184
Medial heel	361 $\pm$ 115	505 $\pm$ 155	719 $\pm$ 185
Lateral heel	341 $\pm$ 96	476 $\pm$ 134	683 $\pm$ 171
Midfoot	136 $\pm$ 94	110 $\pm$ 58	103 $\pm$ 46
Forefoot	635 $\pm$ 211	737 $\pm$ 226	794 $\pm$ 217
MTPJ 1	362 $\pm$ 224	458 $\pm$ 231	536 $\pm$ 246
MTPJ 2 & 3	462 $\pm$ 193	554 $\pm$ 240	555 $\pm$ 248
MTPJ 4 & 5	288 $\pm$ 155	247 $\pm$ 112	216 $\pm$ 110
Toe 1	450 $\pm$ 239	542 $\pm$ 247	616 $\pm$ 241
Toes 2 & 3	202 $\pm$ 105	252 $\pm$ 112	319 $\pm$ 117
Toes 4 & 5	105 $\pm$ 71	134 $\pm$ 67	183 $\pm$ 74

Table 2:
Location of forefoot and heel pressure peaks (mean $\pm$ sd). R is expressed as % foot length from the heel. Θ is expressed as % foot width from a medial tangent.

Speed ms^{-1}	1.01 $\pm$.03	1.51 $\pm$.04	1.99 $\pm$.05
n	129	131	130
Heel	R % 11.9 $\pm$ 3.2	10.7 $\pm$ 4.1	10.3 $\pm$ 4.1
	Θ % 44.4 $\pm$ 10.4	47.8 $\pm$ 11.6	47.1 $\pm$ 10.4
F'foot	R % 78.8 $\pm$ 8.4	80.0 $\pm$ 7.8	80.7 $\pm$ 7.9
	Θ % 33.2 $\pm$ 20.7	29.4 $\pm$ 15.8	27.3 $\pm$ 13.8

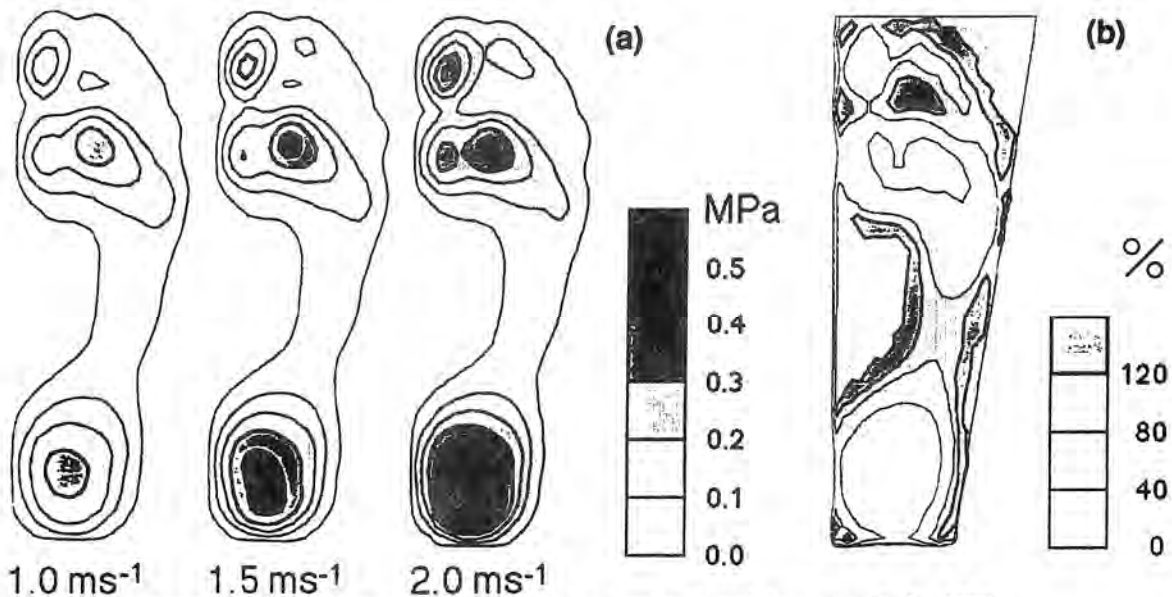


Figure 1: (a) Average peak plantar pressure distributions at 3 walking speeds ($n=130$)
(b) Between subject coefficient of variation at 1.5 ms^{-1} .

correlations ($p < 0.005$) between peak forefoot pressure and mass were found at 1.5 and 2.0 ms^{-1} but shared variance was low ($< 8\%$). Similarly, low correlations were found between peak pressures and plantar surface area. At all three speeds, male subjects had significantly greater peak pressures than females under the mid foot (40% greater on average, $p < 0.01$), 2nd and 3rd MTPJs (19%, $p < 0.05$) and 4th and 5th MTPJs (37%, $p < 0.01$).

DISCUSSION

Clarke (1980) studied 8 female and 19 male subjects walking at 1.33 and 1.79 ms^{-1} . Peak pressures were found to increase with speed, except in the lateral midfoot and lateral MTPJ regions where no significant differences were found. A summed peak pressure index had only a low correlation with body weight and did not differ significantly between sexes.

The results of this study are generally in agreement with Clarke's. However, rather than showing no change, the lateral forefoot demonstrated a significant unweighting at higher walking speeds. Increasing pressure beneath the medial forefoot and decreasing pressure in the lateral forefoot were accompanied by a significant medial shift in the location of the forefoot pressure peak. These changes are consistent with the "two-speed" model of foot function proposed by Bojsen-Møller^{5,6} who demonstrated that the push off in walking may be accomplished about a transverse axis joining the heads of the first two MTPJs or an oblique axis formed by the heads of the 2nd-5th MTPJs. The medial weight shift reported here would represent the trend away from the "low gear" oblique axis and towards the "high gear" transverse axis with increasing speed that Bojsen-Møller suggested.

Women were found to have significantly lower peak pressures in the lateral forefoot and midfoot region. Given the observed trend for lateral unweighting at higher speeds, these gender differences may be due to differences in stature between the two groups which result in a given absolute speed representing a higher relative speed for females.

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A METHOD OF MEASURING TRANSLATIONAL AND ROTATIONAL TRACTION CHARACTERISTICS OF FOOTWEAR

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Since the friction of athletic shoe outsoles is dependent on normal load and velocity (Valiant, 1987), a device was developed to measure the friction possessed by different shoe outsoles on different surfaces under realistic loading conditions. This device differs from others that measure translational and rotational (Bonstingl et. al, 1975; van Gheluwe et. al, 1983) or translational (Schlaepfer et. al, 1983) traction of footwear.

DESCRIPTION OF DEVICE

A vertical shaft is pinned to a modified shoemaker's last (footform) in a hole normal to its bottom surface at the medial metatarsal head location. A shoe pulled over the footform can be tested at any normal load by stacking steel weights on the shaft. Surfaces are interchangeable and are firmly bolted to a six degree-of-freedom force measuring platform (Figure 1).

In the translation test, a shoe is pulled 0.35 m by an air cylinder piston, which is linked to the vertical shaft. The sliding velocity of the shoe on the surface can be adjusted to speeds approaching footstrike velocities by controlling air pressure and the flow of exhaust. The footform can be pinned in many different positions to simulate different foot orientations. Friction evaluations are limited to the anterior outsole. In the rotation test, a shoe is rotated 2.15 radians about a vertical axis through the medial metatarsal heads by a set of gears connected to a pressurized rotary actuator.

Piston velocity is measured with a linear velocity transducer (LVT). Finite difference differentiation of a potentiometer signal gives angular velocity of the shoe. The frictional and normal forces, the free moment resisting rotation, velocity, and angular displacement can be sampled at 500 Hz or greater.

EXAMPLE RESULTS

Taking up the small amount of slack in the system at the onset of translation jerks the shoe, causing a temporary unloading of normal force (FZ) (Figure 2). However, shear force changes in phase with FZ fluctuations lead to a smaller change in their ratio (MU). After 165 ms of acceleration, there are small fluctuations in velocity. At slower velocities, fluctuations are less, not only in velocity, but also in FZ and in the coefficient of friction (MU). Kinetic coefficients for each outsole during relatively constant velocity motion are specified by a range of values (1.37 - 1.81 solid, 0.75 - 1.10 broken). The initial fluctuation in normal and frictional forces, and the inability to measure MU at the instant of impending motion result in a less reliable measure of static coefficients.

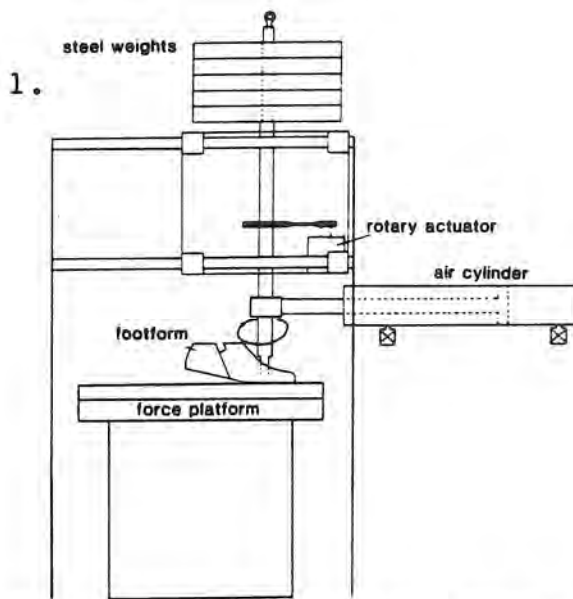


Fig 1 Traction testing device.

Fig 2 Mean translational traction characteristics ($n = 5$) of two basketball shoe outsoles tested anteriorly on a urethane finished surface.

Fig 3 Mean moments resisting rotation ($n = 5$) of two different cleated shoe designs on artificial turf.

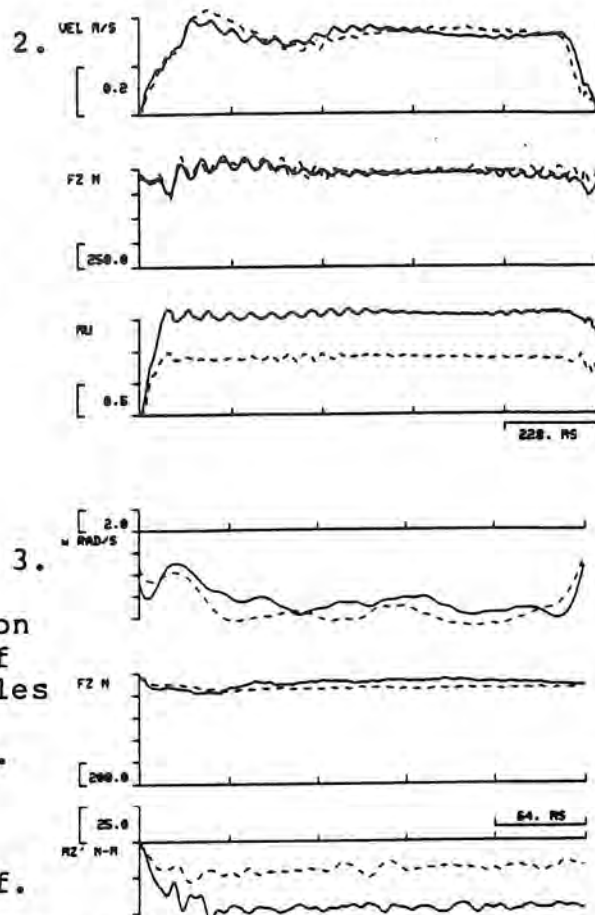


Figure 3 compares the rotational traction on artificial turf of two cleated outsoles. The magnitude of the mean peak moment resisting rotation (MZ') is analogous to a peak friction force developed during translation. The greater peak MZ' denoted by the solid line indicates increased rotational friction.

CONCLUSIONS

Translational friction is quantified by measuring static and kinetic coefficients for different outsole/surface interactions. Moments resisting shoe rotation quantify rotational traction. Measurements are made at loads and velocities equivalent to those occurring in sports for which each outsole was designed.

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DETERMINATION OF HIP, KNEE, AND ANKLE RANGE OF MOTION DURING CYCLING USING AN ADJUSTABLE CRANK ASSEMBLY

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INTRODUCTION

Endurance training can be gained by patients who ride a stationary bicycle. Some patients, however, have limited range of motion (ROM) in the hip, knee, or ankle which limits their ability to ride a stationary bicycle. They often cannot place their disabled lower limb on the crossbar or handlebar to keep that limb from being involved with the moving pedal or crank shaft. Bell and associates have shown that riding one-legged on a stationary bicycle can maintain and may even improve one's cardiorespiratory fitness. The awkward positions used to exercise one-legged, however, is a major problem. A crank/pedal system which could limit lower limb motion on the involved side would eliminate or reduce this position problem. The purpose of this study was to investigate the effects of an adjustable crank/pedal assembly (ACPA) developed by V. Fredriksson on the ROM of the hip, knee and ankle.

METHODS

Nine volunteers (four males and five females) ranging in age from 21 to 31 years participated in the study. The subjects reported no musculoskeletal disorders or limited ROM of the lower limbs. All subjects had recreational cycling experience. Informed consent was obtained from all subjects.

Each subject rode a stationary bicycle (Monark 865) during five conditions. The adjustable crank/pedal assembly (ACPA) was attached to the bicycle in place of the right pedal. The five different conditions were: (1) left side with normal pedal; (2) right side with the ACPA set to approximate a normal pedal; (3) right side with the ACPA set to its maximum motion; (4) right side with the ACPA set to its minimum motion; and (5) the right side with the ACPA set at its mid-setting.

Each subject was marked with white tape on the left and right sides at the highest point of the iliac crest, the greater trochanter, the center of the knee joint, the lateral malleolus, and along the fifth metatarsal. The bicycle seat height to the top of the pedal in its lowest position was set equal to the subject's ischial tuberosity height. The handlebar height was set so that the subject's elbows were flexed to 90 degrees. A video camera at knee height was set twenty feet from the side of the stationary bicycle. The lower limb motion was recorded at 60 frames per second. To eliminate possible blur, the subject pedaled at a rate of 30 revolutions per minute. Before each condition was recorded the subject pedaled for 60 seconds to become comfortable with the pedal setting. A ten-second segment was then saved for analysis.

A video analyzer was used to digitize the data. Each joint angle was calculated by the video system computer every 45 degrees from 0° (pedal at highest position) to 360° (return to highest position) giving eight joint positions

for each joint studied. Three trials for each subject were evaluated. The average ROM of all trials from all subjects was calculated for the hip, knee and ankle (table).

An analysis of variance ($p < .001$) determined that significant differences were present. A Duncan's Multiple Range comparison revealed the specific differences.

RESULTS

The setting of the ACPA at maximum does not change the hip ROM, but increases the ROM for the knee and ankle. The setting of the ACPA at minimum greatly decreased the ROM for all three joints. The setting of the ACPA at the mid position decreases the hip and knee ROM, but does not significantly decrease the ankle ROM.

TABLE

Comparison of the hip, knee and ankle range of motion for five different crank/pedal settings.

<u>Pedal type</u>	<u>Hip</u>	<u>Knee</u>	<u>Ankle</u>
Normal	32.7±6.1	67.7±6.1	24.3±10.5
ACPA set near normal	28.1±9.0	70.2±22.7	28.0±9.7
ACPA set to maximum	30.7±5.1	86.9±6.1*	41.1±11.9*
ACPA set to minimum	11.4±2.7*	12.5±3.0*	12.9±8.4*
ACPA at mid-setting	20.7±4.1*	48.8±7.4*	16.1±6.4

*significantly different from normal $p < .001$.

DISCUSSIONS AND CONCLUSIONS

The results of the normal and approximated ACPA setting pedal conditions were similar to the ROM results found by Ericson et al. The minimal adjusted setting showed that the ROM of the three joints can be greatly reduced compared to the normal pedal. Several settings are available between minimal and normal so that the motion of the joints can be adapted to the patient's needs.

The ACPA can be adjusted to meet the needs of the patient and goals of the treatment.

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THE BIOMECHANICS OF TRAINING SPECIFICITY OF STANDING THROWS IN ATHLETICS

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GENERAL

Many shot putters and discus throwers use standing throw drills in training and as parts of their warm-up routines before competition, quite often performing many such throws. Some coaches feel that this type of throwing should be carefully restricted. For example, Jones (1987) says "The shot put is one dynamic movement and, although the technique can be broken down into part techniques, these must always be brought back into the whole. The closer the season comes, the more the whole technique must be the prime consideration". He suggests that a training session in the competitive season should contain only 10% of standing throws. And, as far as discus is concerned, the same author, Jones (1985), does not even mention standing throws, only recommending half-turn throws and longer movements in this sense. The same problem does not affect javelin and hammer throwers who rarely perform standing throws.

METHOD

In order to quantify the differences in the above statement elite throwers in shot ($n=4$) and discus ($n=6$ including two Olympic finalists) were filmed perpendicular to the line of throw at 45 frames per second (exposure 0.002s each frame) while performing standing and full throws, in which the actual delivery phase was such that the forces at the feet were capable of being measured separately by two force platforms. (Payne, 1973). Two-dimensional film analysis followed, with joint centres and other landmarks being digitized for subsequent kinematic study. Force time records were processed to obtain driving (in direction of throw) and braking (opposite to direction of throw) impulses. Moments of force from each foot about the centre of mass were integrated with respect to time to determine the contribution of each foot to the forward rotation of the thrower.

RESULTS AND OBSERVATIONS

Some of the more important kinematic measurements are shown in Table 1. Although the range of the lean angle (angle between the horizontal and the line joining CM to the point of application of the force from the front foot) showed little variation between types of throws, the combined effects of both feet produced the differing centre of mass paths as shown in Figure 1. It was difficult to pinpoint the start of delivery in standing throws but contact times of the front foot as well as the rear, showed differences from full throws. Centre of mass and C7 horizontal velocities revealed large differences for the two types of throw, whereas CM vertical velocities were very similar.

Table 1. SELECTED KINEMATICS (SDs in brackets)

	LEAN ANGLE RANGE (DEGS)	TRUNK ANGLE RANGE (DEGS)	CONTACT TIME FRONT FOOT(S)	CM HORIZ VELOCITY (m/s)	CM VERT VELOCITY (m/s)	C7 HORIZ VELOCITY (m/s)	TRUNK ANGULAR VELOCITY (Degs/s)
SHOT GLIDE	45-78(3)	40-92(6)	0.37(0.04)	1.74(0.14)	0.84(0.09)	2.39(0.32)	138(40)
SHOT STANDING	46-80(5)	41-89(8)	0.42(0.08)	1.17(0.03)	0.90(0.09)	1.66(0.07)	115(19)
DISCUS TURN	51-76(3)	57-83(7)	0.25(0.03)	1.80(0.20)	0.80(0.14)	2.40(0.30)	117(18)
DISCUS STANDING	52-69(5)	56-75(5)	0.30(0.03)	0.90(0.22)	0.80(0.06)	1.26(0.37)	61(24)

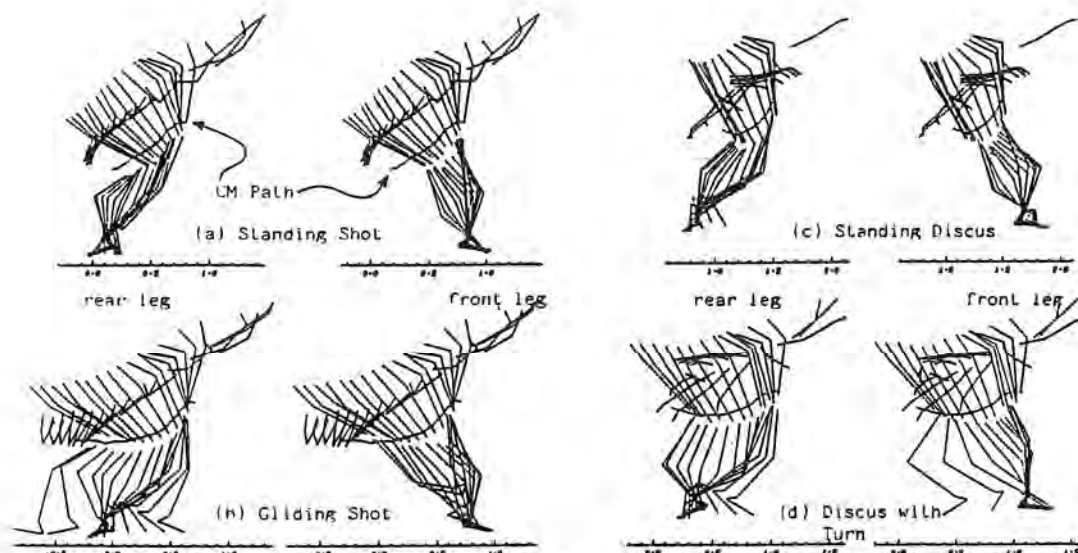


FIGURE 1 SHOT PUTTING AND DISCUS THROWING BY THE SAME ATHLETE. (Segment "sticks" for left leg and left arm omitted in rear leg sequences and right leg and left arm omitted in front leg sequences. Scales in metres.)

Some of the more important kinetic measurements are shown in Table 2. All the force time integrals revealed differences between standing and full throws, but the largest were those between rear foot driving impulses and, to a slightly lesser degree, between front foot braking impulses. The values obtained for the shot glide compared well with those of Dessureault (1978). Discus front foot braking impulse differences appeared to be slightly less than those in the shot. Integrals of moments of force about the centre of mass were unreliable because of measurement error compounding but they suggested that forward angular momentum in standing throws was less than full throws, especially in the shot, the lower values for discus being due, possibly, to contributions to the throw of movements in planes other than the vertical.

Table 2 SELECTED KINETICS (SDs in brackets)

	REAR FOOT INTEGRALS				FRONT FOOT INTEGRALS			
	VERTICAL IMPULSE (N.s)	HORIZONTAL IMPULSE (N.s)		INTEGRAL MOMENT ABOUT CM (Nms)	VERTICAL IMPULSE (N.s)	HORIZONTAL IMPULSE (N.s)		INTEGRAL MOMENT ABOUT CM (N.m.s.)
		BRAKING	DRIVING			BRAKING	DRIVING	
SHOT GLIDE	365(37)	35(6)	39(13)	53(11)	263(31)	105(25)	4(2)	-14(4)
SHOT STANDING	760(215)	10(5)	169(21)	37(13)	233(47)	51(16)	7(2)	-25(11)
DISCUS TURN	362(73)	38(17)	22(12)	36(14)	258(59)	100(23)	4(3)	-28(22)
DISCUS STANDING	960(427)	8(5)	160(20)	46(15)	416(63)	85(21)	15(5)	-39(16)

CONCLUSION

Although apparently similar skills, analysis reveals that standing throws are only superficially related to full throws in shot and discus. Instantaneous body positions may seem similar but horizontal velocities are very different. Rear foot driving impulses dominate in the standing throws, whereas the braking impulses of both feet are predominant in the full throws. These differences suggest that standing throws are so weakly related to full throws that although they may be useful training and warm-up routines, their use should be carefully limited in order not to compromise the skill of full throws.

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NUMERICAL DIFFERENTIATION OF NOISY DATA

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In order to conduct mechanical analyses of human movement it is necessary to determine velocities and accelerations from noisy displacement data. In general this is accomplished by fitting a mathematical function to the data. For each type of fitting function there are constraints imposed on the derivatives at the endpoints of the interval. In the case of a natural quintic spline the endpoint constraints are that the third derivatives be zero. Such constraints are arbitrary and may be inappropriate for a given data set. This paper presents a method of curve fitting in which the endpoint constraints are determined by the quality of the data. In this way endpoint derivatives are only set to zero when their estimates are not significantly different from zero.

Background

In this section a method for increasing the rate of convergence of an interpolating cosine series is given (adapted from Lanczos, 1966). The endpoint constraints for a cosine series are that the odd derivatives be set to zero. If the fitting function is the sum of a cosine series and a quartic polynomial whose first and third endpoint derivatives are close to those of the function which underlies the data, the endpoint derivatives below the fifth are unconstrained for the cosine series. In this situation the cosine series terms decrease rapidly so that fewer terms are needed for an accurate representation of the data. Before this formulation can be applied, methods are needed for estimating the endpoint derivatives and the number of effective terms in the cosine series.

Procedure

This section describes the procedure for determining the endpoint derivatives of the underlying function and the number of effective terms of the cosine series. Let a_k be the coefficients of the cosine series which interpolates the data set. If α_k are the cosine coefficients when the quartic p is subtracted from the data and p has a cosine representation with coefficients c_k then a_k is the sum of c_k and α_k . If k is sufficiently

large, α_k will consist almost entirely of noise and a_k will be close to c_k . This yields the approximation:

$$\bar{a}_k = [p^1(0)+p^1(1)]\phi_1(k)+[p^3(0)+p^3(1)]\phi_2(k)\dots\dots\dots(1)$$

where the data is defined on the interval [01] and ϕ_1, ϕ_2 are functions of k . Values for the four required endpoint derivatives may be calculated using equation (1) for all k greater than a certain m . The closeness of fit of the overdetermined system gives a measure of the error in the a_k . This error estimate may be used to obtain error estimates of the values of the endpoint derivatives at the endpoints. If this error estimate is small compared with the third derivatives, m is used as the number of effective terms of the cosine series. If the error estimate is large, $p^3(0)$ and $p^3(1)$ are set to zero and the whole procedure is repeated using a quadratic rather than a quartic and using first derivatives rather than third derivatives.

Results

Second derivatives were determined using this method and using the quintic spline procedure of Woltring (1986). Four mathematical functions with three levels of added noise were used to produce the data sets. The two methods were compared by evaluating the errors at the interval endpoints and the maximum errors within the interval. The quartic-cosine representation produced lower errors in 23 out of 24 comparisons. In general these errors were an order of magnitude smaller than those produced by the quintic spline.

Acknowledgement

The development of this procedure was supported by the Natural Sciences and Engineering Research Council of Canada.

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THE RELATIONSHIP BETWEEN DENSITY AND STIFFNESS OF LONG BONE

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Introduction

The knowledge and the ability to measure the mechanical properties of long bone is one of the objectives in tissue biomechanics. Fields of application include problems like osteoporosis, fatigue fractures in endurance sports, and so on. At the present time in vivo measurements of the density of bone, which is related to the E-modulus, are based on invasive or radiological methods. These methods, however, are not suitable for periodic routine checks. The goal of the present study was to develop a noninvasive, nonradiologic system to estimate the bone mineral content by measuring the phase velocities of the first mode of flexural waves in human tibiae.

Methods

For the measurements the diaphysis of the tibia was chosen. Using an electromagnetic hammer, a mechanical impulse was applied to the tibia. The measuring system consisted of two accelerometers mounted on an adapter. The adapter was placed on the fascies medialis tibia (see figure 1, left). The line of application of the impact was parallel to the axes of the accelerometers and thus perpendicular to the fascies medialis. The preamplified signals were stored in a digital oscilloscope and then transferred to a minicomputer for further analysis. Figure 1, right, shows an example of the accelerometer signals. A dispersion analysis was performed following the theory of Pochhammer (1876) who proposed a solution of the threedimensional wave equation based on the relationship between the cross-sectional diameter of the body (a) and the wave length (λ). As a reference, the bone mineral content was estimated using the Cameron-Sorensen technique (Cameron and Sorensen, 1963). For this purpose the bone mineral analyser (BMA) from Northland Instruments Corporation was used. The

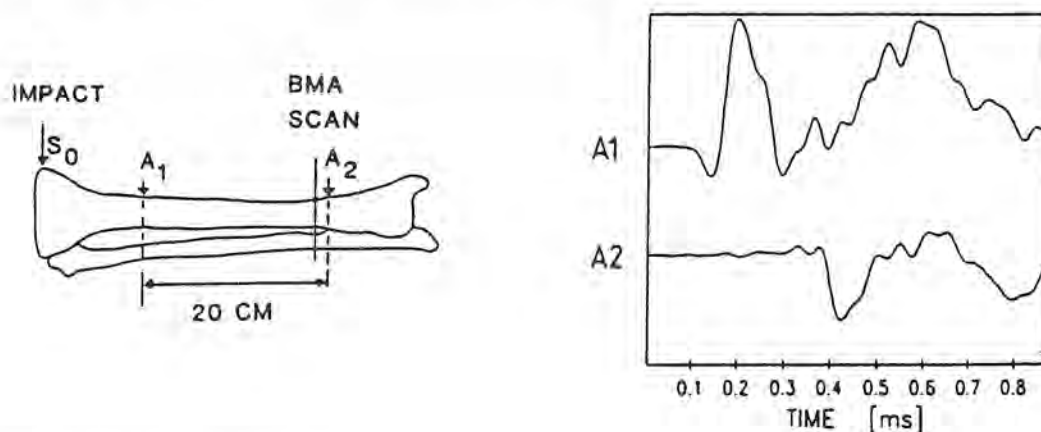


Figure 1: (left) Point of application of the impact (S_0), position of accelerometers (A_1 and A_2). (right) Example of accelerometer signals.

results from the dispersion analysis were then correlated to the measured BMA values. The measurements were performed on a sample of 43 subjects (19 females, 24 males). 11 healthy subjects and 32 patients suffering from renal diseases were included. The latter group was chosen in order to obtain a broad range of bone mineral content.

Results

Figure 2 shows the phase velocities ($a/\lambda = 0.185$) for the 43 subjects calculated from the in vivo measurements and plotted against the BMA values from the bone mineral analyser. For the sample tested the correlation coefficient $r = 0.83$ with an error probability of $p = 0.01$ is statistically significant.

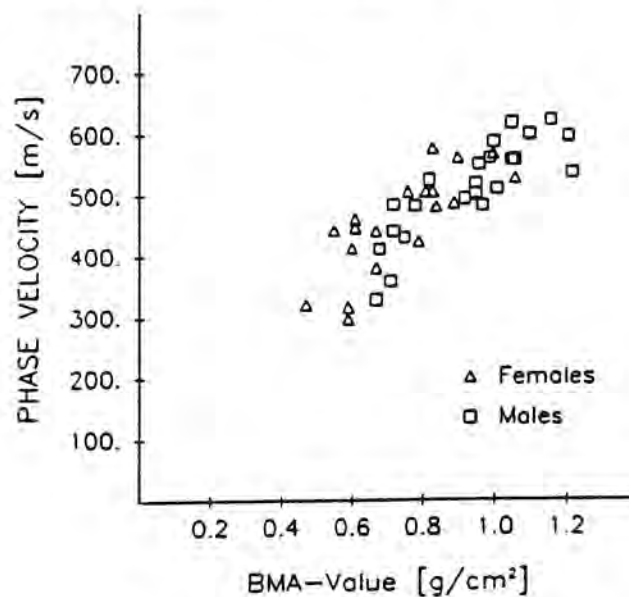


Figure 2: BMA values and phase velocities.

Discussion

The dispersion analysis seems to be an appropriate approach to determine bone mineral content in human tibia. The system is non-invasive and non-radiological. The results show that the present method may be further developed in order to obtain a valid tool for routine diagnostic purposes. Furthermore, this method would allow an estimation of a mean E-modulus over the measured section of the tibia provided that more geometric data are available. This information would improve the knowledge of ultimate load of this biological structure in vivo.

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SCALING OF FOOT DIMENSIONS

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Introduction

Various scaling models have been developed to predict the changes in structure necessary to allow proper function when comparing large animals and small animals. A power curve regression equation, $Y = A \cdot X^b$, is often used to describe the various types of scaling¹. Y and X are morphological variables, b (the scaling factor) is an exponent which describes the changes in their relationship and A is a constant. Isometric scaling suggests proportionately similar measures while allometric scaling suggests proportionately different measures with a change in size. Allometric scaling is the general rule for comparisons of animals or plants greatly differing in size while isometric scaling is the general rule for limited size range differences or within species comparisons^{3,4}. This suggests that except for special situations Humans scale isometrically.

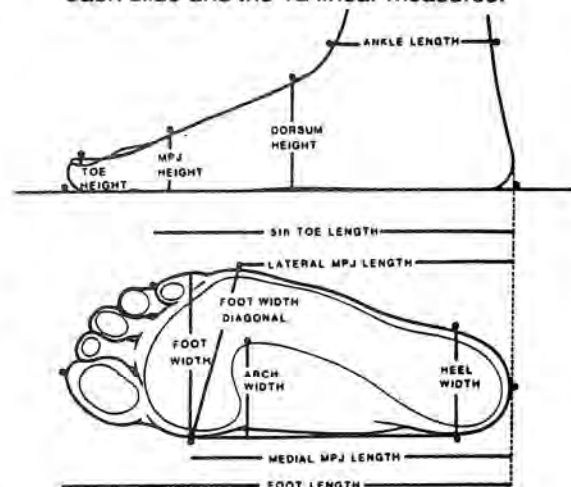
If the foot scales isometrically then the following relationships would be expected: plantar area $\propto$ foot length², body mass $\propto$ foot length³, static plantar pressure $\propto$ foot length. These relationships are particularly important in understanding foot function. The purpose of this study was to determine the type of scaling exhibited by foot structures.

MATERIALS AND METHODS

A total of 897 active, predominantly Caucasian volunteers from various regions of the United States participated in this study. The sample consisted of 225 females and 672 males ranging in age from 18 to 69 years (average 32 years).

A purpose built platform with three attached mirrors which allowed anterior, posterior, medial and plantar views of the foot in one picture was used. Each mirror was positioned to allow only one side of a calibration cube, aligned with a reference mark which was parallel with the platform's front edge, to be viewed in each mirror through the view finder of a correctly positioned camera. Subjects were asked to step on the platform, place full body weight on the right leg and use the left foot (placed on a foot rest) only to maintain balance. The photographer then checked to make certain the leg was approximately perpendicular to the plate and that the plantar surface contact pressure pattern of the foot was evenly distributed. The foot was palpated and marked with black ink at the appropriate landmarks. The photograph was then taken.

Figure 1. A schematic foot diagram showing the 15 points digitized from each slide and the 12 linear measures.



Photographic slides were digitized using a SAC GP-3 Graph/pen Sonic Digitizer with cursor (Science Accessories Corporation). The digitizer was mounted on a back projection system which displayed approximately life size images for analysis. An appropriate scaling factor to convert the measurements from film to real dimensions was obtained by digitizing the calibration cube. The problem of aligning the image to the digitizer's axes was eliminated by digitizing two reference points from each view and referring all measurements to the reference line formed by these points. Foot dimensions were calculated from the coordinates of 15 points on each slide. The initial analysis included 12 linear measurements (figure 1). The outline and contact areas of the plantar surface were also digitized. Two estimates of static plantar pressure were calculated:

reported body mass divided by contact area (contact pressure) and also mass divided by outline area (outline pressure).

The mean measurements for each centimeter of foot length were determined and used in a power curve regression analysis where Y is each measurement of interest, X is foot length, A is a constant and b is the scaling factor.

Results

The scaling factor and the correlation coefficient for the power curve regression analyses are shown in the table.

The isometric scaling model was the best predictor of length and area measures when considering changes in foot length. None of the traditional scaling models could be used to predict body mass, stature, width measures or height measures with changes in foot length.

A power curve regression analysis was completed using mass as the dependent variable and stature as the independent variable. An isometric relationship was indicated.

Y	b scaling factor	R correlation coefficient
mass	1.94	.99
stature	.65	.99
5th toe length	.98	.99
medial MPJ length	.99	.99
lateral MPJ length	.99	.99
foot width	.74	.99
heel width	.76	.98
dorsum height	.50	.90
MPJ height	.64	.99
toe height	.75	.95
plantar contact area	1.98	.99
plantar outline area	1.80	.99
contact area pressure	.01	.01
outline area pressure	.18	.84

Discussion

None of the traditional scaling models were sufficient to predict all the differences in dimensions between large and small feet. This suggests that the foot scales differently from the rest of the body in order to accommodate different selective pressures. The relationship of mass and stature to foot length is one indication of this difference. Mass and stature scale isometrically in comparison to each other, but, as foot length increases, mass and stature become proportionately smaller. Because both total body mass and plantar area increase in proportion to foot length squared, the calculated static plantar pressures remain constant as foot length changes. This indicates that plantar pressure is one possible selective pressure influencing foot scaling. Isometric scaling of length oriented foot measures suggests that when the foot is modelled as a type 1 lever⁵ the mechanical advantage remains constant at different foot lengths. As foot length increases foot height oriented measures are proportionately lower which suggests a mechanism to compensate for the instability associated with increased height.

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A NOVEL APPROACH FOR MODELING TRANSIENT LENGTHENING WITH A HILL-BASED MUSCLE MODEL

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Hill-based muscle models, characterized by a "contractile element" (CE) in series and in parallel with viscoelastic elements [1,2], have often been considered to be a good "compromise" between overly simple lower-order "systems" models and extremely complex biophysically-based "cross-bridge - sliding-filament" models [3,4]. This type of lumped-parameter model is especially useful if care is taken to include each of the following nonlinear properties: CE force-velocity, CE length-tension, SE force-extension and PE force-length [2]. Parameter estimation for such a formulation is straightforward and values shouldn't need to change as a function of task [2]. Such models have helped illuminate the sources of classical phenomena such as impedance modulation via co-contraction of antagonistic muscles [2,5].

Identification of Problem. There is, however, one type of behavior that Huxley-based biophysical models have been able to adequately simulate for which Hill-based models have traditionally had difficulty [3-5]: yielding during transient lengthening. Although the amount of actual yielding is controversial, especially under in situ conditions in which muscles are attached to inertial skeletal systems, it is nonetheless an interesting phenomena. In this paper we develop a novel framework that allows us to simulate experimental "yielding" phenomenon while maintaining the Hill-based model. These results appear to eliminate a fundamental criticism of Hill-based models (recent personal communication with authors' of [4,5]).

Method. The conceptual foundation behind our approach rests on the observation (assumption) that, with yielding, the number of attached bonds has decreased but that those instantaneously attached may still be described by Hill's equation, as formulated for lengthening [6]. From this context, the "problem" lies not with the mechanical model but with assumptions for activation-attachment dynamics. The latter is thus a function of not only of neural activity but also of recent muscle mechanical history. Thus, one might establish a relationship between the "activation from above" (F_{ac}) and the actual instantaneous attachment of bonds (F_{at}):

$$F_{at}(N_e, v_{ce}) = F_{ac}(N_e) - F_y(N_e, v_{ce}) \quad [1]$$

Determination of F_y , the instantaneous "yielding", however, is non-trivial. Review of the (sparse) literature reveals the following: i) yielding increases as lengthening velocity increases, with there being a "threshold" velocity below which yielding does not occur [5,7]; ii) the amount of yielding is inversely proportional to the current activation level [7]; and iii) the amount of yielding is greatest with steady or decreasing activation and nearly non-existent when the muscle is increasing activation [8]. Based on these three observations, the following heuristic approach was formulated:

$$T_y \dot{F}_y + F_y = \begin{cases} \left(\frac{v_{ce}}{v_{ce_{max}}} \right)^{0.5} \left(\frac{k F_a + F_{max}}{k + 1} \right) \left(1 - \frac{F_a}{F_{max}} \right) & v_{ce} > v_{cr} \\ 0 & v_{ce} \leq v_{cr} \end{cases} \quad [2]$$

$$\text{where: } T_y(F_a, v_{ce}) = \frac{T_{yo}}{1 + (v_{ce}/v_{ce_{max}})} \quad [3]$$

This formulation was chosen over one presented previously [7] due to the fact that there are only (up to) three new parameters: T_{yo} (by default simply 0.1 sec), the "critical velocity" V_{cr} (small, e.g. 0 in Fig. 1); and the weight k (simply 1 in Fig. 1). The end result is that, relative to activation, the current force depends not only on the present activation and velocity (as in past formulations) but also on the present rate of change of activation and on the previous history. Figure 1a represents a convenient conceptual illusion of the range of yielding behavior via a force-velocity relation; in fact the top hyperbolic curves are always satisfied, but the current instantaneous curve (defined by y-intercept) scales downward appropriately.

Results and Conclusion. The simulations presented in Fig. 1, for the cat soleus system, demonstrate the basic effectiveness of the new approach. Similar results for elbow and knee systems are compatible with our experimental data. We have also explored other types of movements, including a variety of voluntary movements (e.g. fast movement turnarounds), where we find a few interesting results related to drifting (to be presented at the meeting). In summary, the approach presented here represents a substantial improvement over past Hill-based models and appears to effectively eliminate the primary previous criticism of such models.

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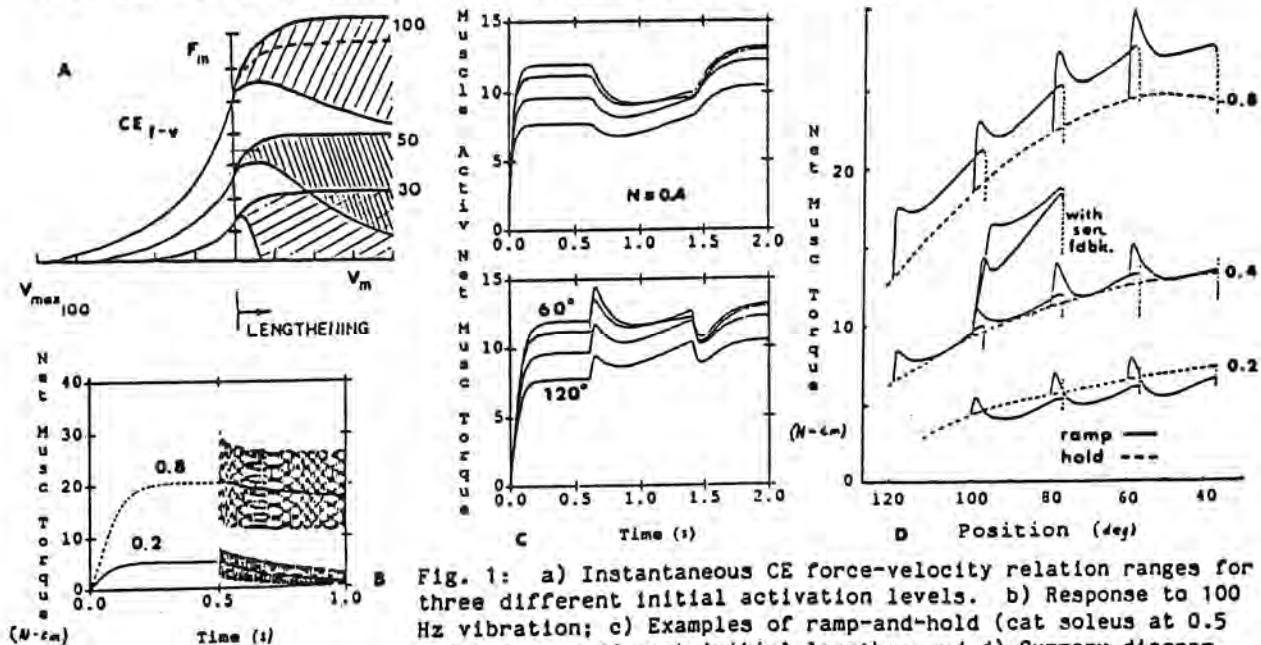


Fig. 1: a) Instantaneous CE force-velocity relation ranges for three different initial activation levels. b) Response to 100 Hz vibration; c) Examples of ramp-and-hold (cat soleus at 0.5 Lo/s) from different initial lengths; and d) Summary diagram, force-length behavior for cat soleus (similar to Fig. 3 of [7]).

CHOICES FOR OPTIMIZATION CRITERIA FOR GOAL-DIRECTED SINGLE-JOINT MOVEMENTS

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Humans perform a variety of tasks without much conscious planning. The nature of the underlying control strategies used during goal-directed tasks has been under investigation by several groups. Consistent patterns, each as a function of the experimental paradigm and the command to the subject, have been identified experimentally (e.g. [1,2]) and in some cases hypotheses have been proposed that appear adequate for a limited set of tasks [1-4]. Attempts to identify possible "optimal" performance criteria driving various human movements have met with only moderate success and often controversy. In our quest toward deeper understanding of underlying neuromotor control strategies for various moderate- and high-speed goal-directed movements, we have re-examined the concept of "optimized" (nearly optimal) movements. The "system" being controlled as an eighth-order nonlinear antagonistic muscle joint model, described elsewhere [5]. We have concentrated on elbow, wrist and head movements. An optimization algorithm, described elsewhere [6,7], is utilized to determine the widths and heights of neuro-controller input pulses which minimize a generalized performance (cost) criterion of the form:

$$J_c = k_t MT^n + \int [k_p PE^n + k_j JK^n + \sum_m K (k_n NE^n + k_m MS^n + k_d MD^n)] dt \quad [1]$$

where MT is movement time relative to some desired time, PE is position error relative to a reference trajectory, JK is jerk (derivative of acceleration), NE is neural activity, MS is muscle stress (muscle force normalized to maximum), MD is muscle dissipation (product of CE force loss and internal CE velocity normalized to their maximum), and typically $n = 2$. Notice that these criteria fall into two categories: 1) kinematic measures based on the joint position trajectory (MT, PE, JK); and 2) neuromuscular "activity" measures (NE, MS and MD). Effects of relative weights between pairs of subcriteria on optimized control strategies of the movement and on inactive other subcriteria have been explored [8]. The goal of this paper is to use our assessment of the merits of each subcriteria, based on their ability to correctly predict experimental data, to suggest "default" relative penalties between certain subcriteria which we have found to be effective. In particular, we propose that all three neuromuscular measures be considered as a unit, with relative weights between them kept invariant regardless of task.

ASSESSMENT OF INDIVIDUAL OPTIMIZATION SUBCRITERIA

Opposing performance criteria can be weighed against each other in the cost function for various tasks [7,8]. Based on hundreds of optimized solutions, we offer the following brief insights on each subcriteria:

Movement Time (MT) and Position Error (PE). Duration of movement as well as magnitude and speed of desired trajectory may be specified to obtain different movements. These criteria, by themselves, cause aggressive multi-pulse control strategies with high co-contraction that do not seem to correspond to experimental data. Except for cases of extreme accuracy or speed, fast goal-directed movements seem to be bound by other considerations.

Joint Jerk (JK): The smoothness of the joint trajectory comes at a cost of co-activation of antagonistic muscles for smooth transitions between dynamic phases. Despite the publicity that this criterion has enjoyed recently for uni- and multiarticular movements [e.g. [4]], simulation results for single

joint movements indicate that the smoothness of trajectories is just a by-product of the nonlinear filtering properties of muscle [8].

Neural Effort (NE). This penalty for high neural activity represents the fact that it is "costly" for motoneurons, the "final common pathway" of massive excitatory and inhibitory information (including negative feedback via Renshaw cells), to reach and maintain high excitation levels. Since activation during high shortening velocity does not result in much muscle force, with penalty for NE included the first pulse width is reduced and the antagonist pulse is adjusted in height. This gives nearly reciprocal innervation, which is reported by many [1]. With increased penalty, this subcriterion increases movement time, switches strategies from pulse width to pulse height modulation, and loses the antagonist burst.

Muscle Stress (MS). This criterion has been used with limited success to determine load sharing of synergistic muscles during tasks such as gait. Justification of MS is the fact that sustained muscle forces causes fatigue and relates to the energy expenditure of the biomechanical system. For this reason, MS penalizes inefficient muscle forces (e.g. co-activation of antagonistic muscles) [8]. Pulse width modulation is prevalent since higher excitation levels result in only moderate muscle forces during movement initiation; thus "speed insensitive" [1] strategies are enforced.

Muscle Dynamic Dissipation (MD). Inclusion of MD in the performance index has effects similar to MS during movement initiation. However, since it allows high neural inputs as long as the internal velocity is low, it promotes the tri-phasic patterns sometimes seen experimentally. Unlike MS, it allows co-contraction during quasi-static conditions.

RECOMMENDATIONS AND CONCLUSION

Each subcriterion serves a specific, separate purpose. Through systematic variation of relative weights of the subcriteria, many continuums of moderate to high speed movement were obtained. A comparison with existing experimental data shows that physiologically-consistent, smooth trajectories occur as a byproduct of the natural "tension" between neuromuscular "activity" measures and "kinematic" subcriteria, the latter serving primarily to help define the task (e.g. movement time, distance). Our collective knowledge of all studied subcriteria suggests no one neuromuscular "activity" subcriterion is adequate. Rather, we strongly suggest that all of NE, MS and MD be utilized, with the relative weights between them being 1, 5 and 5, respectively. The single parameter K crudely represents the "energy" cost of neuromuscular activity. Values of about 50 for kp/K result in fast optimized movements, while efficient moderate speed movements, with neuro controller signals compatible with experimental results, are obtained with $kp/K = 5$. Although we suggest that movement strategies are greatly affected by the instructions as well as the test environment, with there being no magic "global" criterion, we also suggest a lumped neuromuscular subcriterion.

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SIMULATION OF NEUROMUSCULOSKELETAL SYSTEMS: A NOVEL COMPUTER PACKAGE

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This paper presents the framework behind a three-dimensional (3-D) simulation package for neuromusculoskeletal systems. This menu-driven program, an extension of JAMM [1], is written in Ratfor (preprocessor for Fortran) on the UNIX operating system. It is currently utilized in three courses taught by the author (biomechanics, biocontrol systems, neuromuscular control systems) as well as for ongoing research projects involving 3-D eye-head-neck movements, 3-D knee biomechanics, planar upper-limb movements and human and cat ankle movements. As with most past approaches, it assumes that bones are rigid links and that muscles are point-to-point elements. However, it includes a number of novel advanced features that will now be described.

A. Menu-Driven, Flexible. This package currently supports three types of human movement systems: i) a multi-link head-eye-neck model; ii) an upper-limb shoulder-elbow-wrist model; and iii) a hip-knee-ankle model. Each utilize the same program but different parameter data bases. Creation of new models and parameter scaling for body size can thus be accomplished independently of the program (e.g. we also have a model for the cat ankle system). All states are initialized automatically once the kinematic configuration, any external loading, and co-contraction levels are specified. Neural inputs and external loading are driven either by a user-specified combination of pulses, ramps and sinusoids or via data files (e.g. with EMG's). Separate 10-column output files are created for each muscle and joint. Automated graphing of any column versus any other, as specified within a spreadsheet table, is possible, courtesy of programs written within the RS1 program by Amir H. Seif-Naraghi.

B. Realistic Nonlinear Muscle Models. A Hill-based muscle model structure, improved to include several new features, is utilized. Each of four classical nonlinear elements is by default included [1]: i) contractile element (CE) force-velocity; ii) CE length-tension; iii) series viscoelastic element; and iv) parallel viscoelastic element. In addition, the muscle excitation-contraction coupling process is represented by [2]: i) first-order dynamics between an idealized neural excitation and an estimate of the EMG (if model not driven by EMG); then ii) a nonlinear first-order model between EMG and activation, and then, iii) for the case of lengthening muscle, a nonlinear "yielding" function that is a function of both activation and CE velocity. Subsets of the above are selectively available. Model parameters are estimated by systematic techniques described elsewhere [1,3].

C. Curved Muscle Paths. Experience suggests that the assumption of a straight-line muscle path between origin and insertion is often totally inappropriate. Muscle paths are curved via the use of a "minimum radius" constraint surrounding joints - if the straight-line path between origin and insertion sites causes a moment arm less than this minimum, then the path instead arcs around the sphere within the same instantaneous planar slice of 3-space [3]. The muscle moment arms, muscle length and the normal vector defining the planar slice are calculated. It turns out that if one pre-calculates the "critical" angles and segment lengths where arcing first occurs, the calculations for the arced path actually require less computational cost than the straight-line approximation.

D. Closed-Form, Efficient, 3-D Dynamics. A computationally-efficient closed-form technique is employed for 3-D link dynamics. This includes the solution of the inverse dynamics problem by recursive Newton-Euler equations [4], followed, when required, by an "composite center of mass and inertia" formulation for solution of the direct (integral) dynamics problem [5]. Our method includes the use of generalized spherical joints and an approach that minimizes the calculated terms. For instance, all internal positions (e.g. muscle origin-insertion) are only known relative to neighboring links. Numerical integration is performed by a "predictor-corrector" method.

E. Neural Circuitry. The model includes multiple feedback loops, including loops for both kinematic (position, velocity, acceleration) and muscle force information. Basic neural circuitry (interneurons, Renshaw cells) and "noise" sources are available. By default all of these are turned off.

F. Task-Specific Linearization. A option is available that allows each muscle constitutive relation to be linearized for a given run. The linearized model is then re-run with the same neural and loading input.

G. Advanced Tools: Sensitivity Analysis, Optimization. The model is embedded within a general framework that allows advanced techniques such as sensitivity analysis and global optimization to be utilized. Sensitivity analysis, including sensitivity coefficient calculation, can be performed on both internal parameters or on "control" parameters (e.g. feedback gains, neural pulses, and external loads). For global optimization, the following generalized performance criterion is available:

$$J_c = \int \{ |PE^n| + K_v(|VE^n| + K_j|JK^n| + K_l|JL^n|) + \sum (K_n|N^n| + K_m|MS^n| + K_d|MD^n|) \} dt \\ + K_{pf}|PE_f|^n + K_{vf}|VE_f|^n + K_{tf}|MT_r - MT^n|$$

where PE is position error relative to a reference, VE is velocity error, MT is movement time, JK is jerk (derivative of path acceleration), JL represents joint loadings, N is neural activity, MS is muscle stress, and MD is muscle CE "dissipation"; typically all n's are set to 2. The control parameters to be modified can include neural pulses (or ramps or sinusoidal amplitudes), feedback gains, or parameters defining skewing of the EMG. A modified version of the Bremermann Global Optimizer is utilized to find a solution. We utilize these techniques to gain insight into possible neuromotor control strategies.

SUMMARY. The neuromusculoskeletal simulation package described here, while not as graphically "pretty" as some packages, appears to be the most thorough and accurate general-purpose package currently available for simulating human and animal movement. The program, including options for sensitivity analysis and optimization, but not graphics, requires 320 KB of RAM. We will be attempting to download from the Harris HCX-UNIX mainframe system to a UNIX-based PC-386 system in the near future. A detailed manual for the package is being written and our current goal is for the package, complete with manual and examples, to be available for distribution at the June meeting.

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A Model Relating EMG Signals to Elbow Torque during Rapid Loaded Movement.

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INTRODUCTION

A quantification of the relationship between EMG and muscle force during highly dynamic movement is relevant to issues in both motor control and biomechanics. The following paper presents the results of preliminary work towards the development and validation of a general biomechanical model of the elbow joint designed to predict elbow torque from surface EMG measurements of the biceps and triceps brachii muscles and kinematic variables. Model components have been based upon the known characteristics of skeletal muscle as reported in the literature, with individual parameters for each component empirically derived using a sequential calibration procedure on human subjects. Model validation findings presented at this time will consider only rapid elbow flexion and extensions movements with the wrist fully supinated under a high inertial loading condition. Results are reported in terms of goodness-of-fit values for data employed in the calibration process and data independent of the calibration procedure.

MODEL SPECIFICATION

The biceps brachii and triceps brachii muscles were assumed to be representative muscles of two *equivalent* muscle groups independently associated with flexion and extension of the elbow (Bouisset et al., 1977; Bozec et al., 1980). Multiple surface EMG signals from each muscle were collected using separate arrays consisting of three sets of multiple surface electrodes based upon the methods of Redfern (1988). Raw EMG signals were processed using a modified RMS integrator (time constant 50 msec) and stored digitally with the force signal at a rate of 100 Hz. Post-processing of the EMG signals included averaging the three channels for each muscle to achieve a spatially consistent signal and low-pass filtering using a Walsh-Hadamard filter (cut-off frequency approximately 12.5 Hz.).

Following convention, the processed EMG signals were assumed to represent the Active State (AS) of each of the equivalent muscle groups. Electromechanical characteristics of the muscle tissue were modelled using a simple exponential filter and an optimally derived pure delay. A linear function was employed to represent the dynamic relationship between the smoothed and delayed AS of each muscle and internal muscle tension (Lippold, 1952). Quadratic functions (Woods et al.,

1983) were also explored but proved to be of only minimal benefit. Additional model components consisted of: 1) a geometric model of the musculo-skeletal system surrounding the elbow joint; 2) a quadratic scaling factor (LC_B and LC_T) compensating for the length-tension relationships of each muscle considered (Gordon et al., 1966) as a function of included elbow angle; and 3) an exponential scaling factor (VC_B and VC_T) compensating for speed-of-shortening relationship of each muscle considered (Hill, 1938) as a function of included elbow angle and angular velocity. Mathematically, the model is realized as:

$$T(t) = (c_B * E_B(t-\tau))(MA_B)(LC_B)(VC_B) + (c_T * E_T(t-\tau))(MA_T)(LC_T)(VC_T),$$

where T is the estimated elbow torque, E_B and E_T are the exponentially smoothed processed EMG signals from the biceps and triceps, τ is the optimally derived time constant, MA_B and MA_T correspond to estimates of the moment arm function for each equivalent group, c_B and c_T are estimated constants and the other variables are as previously defined.

PARAMETER ESTIMATION

Parameter estimates for individual components of the model presented above were obtained using a sequential piecewise-calibration procedure similar to that employed by Redfern (1988). Three male right-handed subjects completed a limited number of sinusoidal isometric and slow constant velocity calibration movements prior to completing a much larger number of very fast movements to a specified target. Isometric calibration trials were used to verify electrode placement and to establish the proper time delay value τ using cross-correlation techniques. Slow constant velocity elbow flexion and extension movements over the subjects entire range of motion were then employed to estimate the muscle length compensation functions for each subject using standard regression methods. Much faster Fitts type movements (Fitts, 1954) with an index of difficulty equal to 4.42 bits were used to establish both the relationship between EMG and dynamic muscle force and the velocity compensation function.

Trials were conducted with each subject seated in a large fixture that rigidly supported the torso and fixed the shoulder posture at approximately 90°

abduction and 0° flexion. A leather cuff was tightly fixed about the wrist of each subject and was in turn connected to a load cell for force measurement. The entire wrist cuff assembly was mounted on top of a rotating lever arm which in turn was connected to a potentiometer to monitor limb position. Position signals during the movement were digitally filtered and differentiated for use in the muscle length and velocity calibration procedure. In addition to the mass of the rotating lever arm, 9.07 kg of mass was added to the assembly 0.18 m distal to the wrist cuff attachment to increase the inertial loading during the rapid movements.

VALIDATION

The sequential, nonlinear nature of the model proposed above makes it difficult to evaluate goodness-of-fit using traditional measures. For the present work, the following statistic will be used as a general measure of the distance between the measured and predicted data values over any n discrete samples equally spaced in time:

$$R^2_{\text{Equiv}} = 1 - \frac{\sum_{i=1}^n (\hat{y}_i - y_i)^2}{\sum_{i=1}^n (\bar{y} - y_i)^2}$$

The value produced by this statistic can be thought of as representing a distance between the measured and predicted value equivalent to the distance between a measured and predicted values of a model derived by more conventional means with an equal R^2 value.

Goodness-of-fit values for each subject are presented below in Figure 1. Values shown represent the R^2_{Equiv} distance values averaged over the two flexion and two extension trials used in the calibration procedure, the thirteen remaining extension trials, and thirteen remaining flexion trials not used in the parameter estimation process. As indicated by the values in the figure, the model accurately predicts the elbow torque values found during the rapid movements. Further, parameters estimated using only a small portion of the data are shown to be very robust in nature. Systematic deviations between the measured and predicted values can not be easily identified using a simple goodness-of-fit measure and can be best examined by graphically comparing the measured to predicted values. Figure 2 presents a single example of the such a graphical comparison. As shown in Figure 2, the estimates deviate from the measured values most consistently during the secondary corrective phase of each movement. Underestimation of the peak torque values for the flexion trial is seen in

this single example, but was not consistent over all trials as seen in Figure 1.

CONCLUSIONS

The validation results indicate that the elbow torque values accompanying rapid flexion and extension of the elbow can be accurately modelled as a function of measured surface EMG's. Inaccuracy of the model at relatively low torque levels is most likely due to visco-elastic properties of the system not captured in the present model. Future work will center on quantification of these passive components and extension of this work to unloaded rapid movement.

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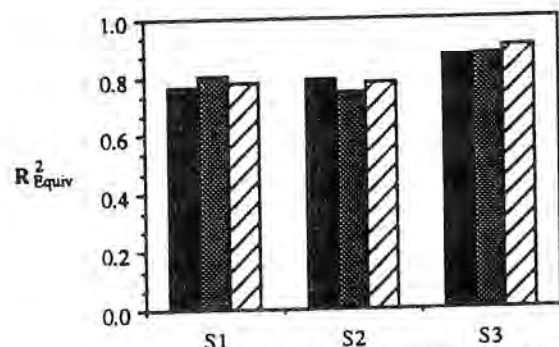


Figure 1. R^2_{Equiv} values for combined data used in parameter estimation (■), and for extension (■), and flexion trials (▨) not used in parameter estimation process.

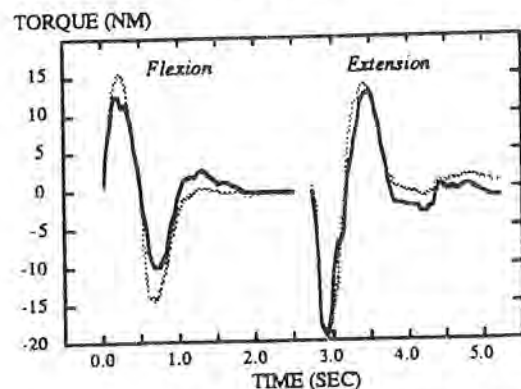


Figure 2. Estimated (black) and measured (gray) elbow torque for a typical rapid flexion and extension

TOWARDS A 3D BIOMECHANICAL TORSO MODEL FOR ASYMMETRIC LOAD HANDLING

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Introduction: It is clear from both a biomechanical and epidemiological perspective that lateral loading of the torso is more stressful to the low back than symmetric loading. The mechanism of injury in most cases is not obvious. This paper reports on the development of a biomechanical model of the torso musculature and passive supporting tissues that comprise a spinal motion segment. It is proposed to simulate various asymmetric loading conditions to determine possible tissue failure conditions, and the manual parameters that cause such failures in industry, sport and leisure activities.

Technical Background: The general logic being used to develop such a torso model is diagrammed in Figure 1.

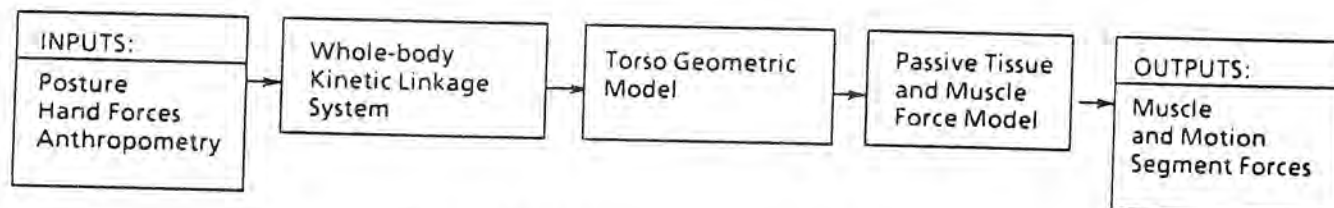


Figure 1. General Torso Model Logic

The Whole-body Kinetic Linkage Model currently being used is depicted in Figure 2. It is a 12 link model which allows the user to easily manipulate the posture, hand force and anthropometry to predict the reactive moments and forces at each of the articulations. The general model is described elsewhere by Garg and Chaffin (1975).

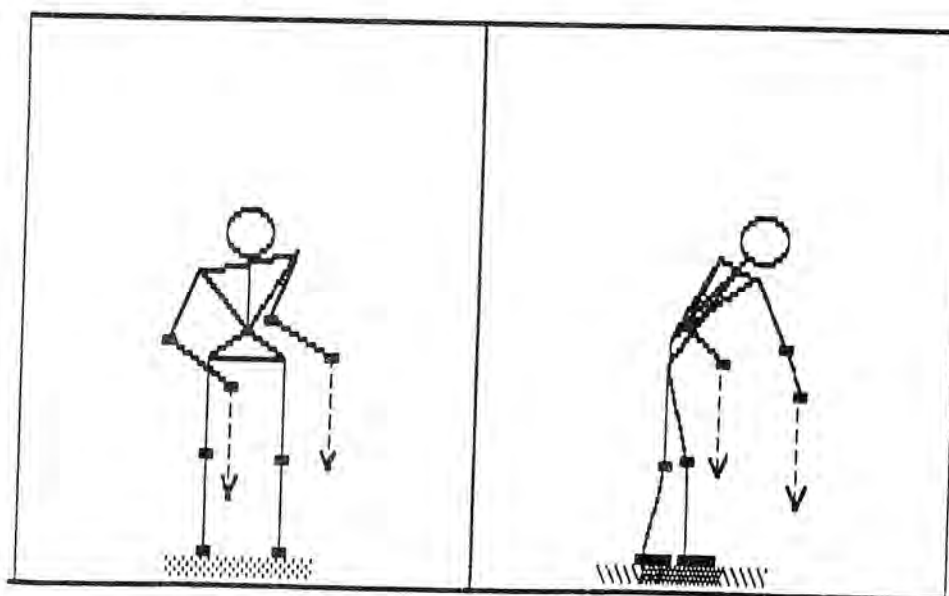


Figure 2. Input Screen from University of Michigan 3D Static Strength Model™

The Torso Geometric Model takes the general torso posture and loading predictions and adds the size, location and shape data necessary to represent the torso muscles and motion segments. Anatomic and recent CT Scan data are used to reconfigure the basic spinal geometric model of Andriacchi et al. (1974) for this purpose.

The Passive Tissue and Muscle Force Model uses optimization methods to predict the necessary internal forces needed to stabilize the column against the reactive moments predicted by the Whole-body Kinetic Linkage Model. Methods described by Bean et al. (1988) and Hughes et al. (1987) are being validated by comparison with torso muscle EMG measurements.

Example Output: One graphic output from the use of the preceding system is depicted in Figure 3 for the posture depicted in Figure 1. The load on each hand is 111N. The erector spinae force vectors represent 915 N on the left and 892 N on the right acting at the L3/L4 level. Other muscle forces are proportionally scaled in the figure.

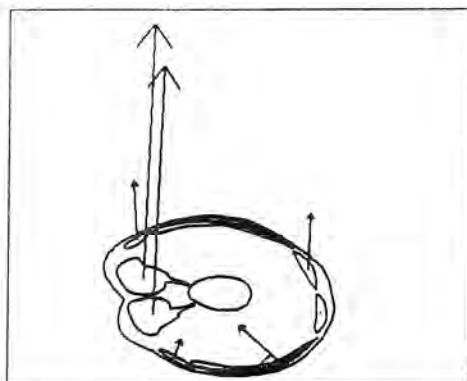


Figure 3. University of Michigan Torso Graphic Model™

Discussion: The models described allow researchers to systematically study the effects of many different biomechanical parameters that are often hypothesized as being important in asymmetric load handling. It is proposed that only with such a comprehensive system modelling approach can the cause and prevention of occupational low back pain be realized.

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Acknowledgment: This work is partially supported by NIOSH Grant OH07207, the Ford Body and Assembly Division, and the Association of American Railroads.

USING KARUSH-KUHN-TUCKER MULTIPLIERS TO EVALUATE THE EFFECT OF ANTAGONIST MUSCLE ACTIVITY

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INTRODUCTION. Human joints are in general statically indeterminate since there are more muscles than mechanical equilibrium conditions. Optimization methods have been formulated to rationally select one muscle force combination. While there is experimental evidence that agonist and antagonist muscles cocontract under many circumstances, many optimization models do not predict such activity. Optimization theory can be used to evaluate the effect of antagonistic muscle activity on the optimal objective function value.

REVIEW AND THEORY. Suppose the moment equilibrium conditions and net force conditions can be written in matrix form as $Ax=b$, where x represents the muscle forces, the i th column of A represents the contribution to joint moment of 1 newton of force in muscle i , and b is the vector of moments the muscles are required to produce to maintain equilibrium. If the objective function to be minimized is $f(x)$, the optimization problem to be solved is $\text{minimize}_x \{f(x) \mid Ax=b, l \leq x \leq u\}$ where l and u are vectors of minimum and maximum allowable muscle forces, respectively. The optimal solution x^* is the vector of predicted muscle forces. Introducing multiplier row vectors λ , π , and μ gives the Lagrangian:

$$L(x, \lambda, \pi, \mu) = f(x) + \lambda(Ax - b) + \pi(x - u) + \mu(l - x)$$

The Karush-Kuhn-Tucker Theorem (Peressini *et al.*, 1988) states that if x^* is an optimal solution, there exists a Lagrange multiplier λ^* and Karush-Kuhn-Tucker (K-K-T) multipliers μ^* and π^* satisfying $\nabla L(x^*, \lambda^*, \pi^*, \mu^*) = 0$, $Ax^* = b$, $x^* \geq l^*$, $\mu^* \geq 0$, $\mu^*(l - x^*) = 0$, $x^* \leq u^*$, $\pi^* \geq 0$, $\pi^*(x^* - u^*) = 0$. The Karush-Kuhn-Tucker multiplier μ^* represents the sensitivity of the optimal objective function value to changes in the lower bound on x , i.e. $\mu^* = \partial f(x)/\partial l_i$. Most optimization models set the lower bound to zero (muscles can only act in tension), so $\mu^* = \partial f(x)/\partial x_i$ for muscles having zero predicted tension. Since many optimization models predict zero muscle force in antagonist muscles (Hughes and Chaffin, 1988), μ^*_i represents the marginal cost of tension in antagonist muscle i .

METHODS. The use of K-K-T multipliers to analyze antagonist activity is applied to a popular model for predicting lumbar region muscle forces during asymmetric loading of the torso (Schultz *et al.*, 1983, and reformulated in Bean *et al.*, 1988). The model chooses ten muscle forces to minimize the compression force on the spine such that three moment equilibrium conditions are satisfied. The objective function is therefore $f(x) = c'x$, where c_i is the component of force normal to the intervertebral disc for a unit of tension in muscle i . All computations are for the L3 level of the spine using the anthropometry of Schultz *et al.* (1983). The muscles included in the model are the left and right erector spinae, rectus abdominus, internal oblique, external oblique, and latissimus dorsi muscles. The model was solved using the simplex method of linear programming. Table 1 summarizes the objective function coefficients used in the model.

Table 1. Objective function coefficients c_j									
LES	RES	LRA	RRA	LIO	RIO	LEO	REO	LLD	RLD
1.0	1.0	1.0	1.0	.71	.71	.71	.71	1.0	1.0

RESULTS. The torso model was solved for three tasks: attempted flexion; attempted right lateral bending; and attempted right axial twisting of the trunk (10 N-m each). Table 2 gives the muscle forces predicted by the model; Table 3 gives the K-K-T multiplier vector for each task.

Table 2. Predicted muscle forces										
TASK	LES	RES	LRA	RRA	LIO	RIO	LEO	REO	LLD	RLD
Flexion	0.0	0.0	22.7	22.7	63.6	63.6	40.9	40.9	0.0	0.0
R. Bending	0.0	76.6	15.2	15.2	15.2	42.5	0.0	27.3	0.0	0.0
R. Axial Rot.	41.6	0.0	6.5	0.0	0.0	56.3	36.2	0.0	0.0	12.4

Table 3. Karush-Kuhn-Tucker multipliers μ^* (marginal costs)										
TASK	LES	RES	LRA	RRA	LIO	RIO	LEO	REO	LLD	RLD
Flexion	2.6	2.6	0.0	0.0	0.0	0.0	0.0	0.0	1.8	1.8
R. Bending	7.6	0.0	0.0	0.0	0.0	0.0	12.3	0.0	3.0	2.1
R. Axial Rot.	0.0	2.3	0.0	1.6	2.7	0.0	0.0	6.9	1.6	0.0

DISCUSSION. The marginal costs are positive for the antagonists and zero for agonists. During attempted flexion the agonist muscles are the rectus abdominus and abdominal oblique muscles; the antagonists are the erector spinae and latissimus dorsi muscles. The marginal cost of 2.6 for the erector spinae muscle means that the spinal compression force increases 2.6 newtons for an additional newton of muscle force in the erector spinae. The marginal cost interpretation of the K-K-T multipliers is also valid for muscles that are not clearly defined as antagonists. In attempted axial rotation, for example, the right rectus abdominus muscle is not a clear agonist or antagonist. Since its predicted muscle force is zero, the K-K-T multiplier of 1.6 means spinal compression force increases 1.6 newtons for an additional newton of right rectus abdominus force. Attempted right lateral bending shows how high the effect of cocontraction on the optimal objective function value can be. The spinal compression force increases 12.6 newtons for an additional newton of force in the left external oblique. The marginal costs are much higher than the parameters used in the objective function (12.6 versus .71 for the external oblique) because the K-K-T multipliers incorporate the moment equilibrium conditions while the objective function c does not. The K-K-T multipliers account for an increase in agonist muscle force to counter the increased antagonist force.

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AUTO-REGRESSION ANALYSIS UPON EMGs POWER SPECTRA DURING DYNAMIC EXERCISE

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INTRODUCTION

Surface electromyogram were recorded as interference signals, which contained many sorts of motor units during dynamic exercise. Those complex signals have been transformed to be a simple power spectra, that have used as the index of muscle fatigue or participation of the motor unit. This study was programed to apply Auto-Regression Model (ARM) for those EMGs Power Spectra, and had a proposal of the new classification in the motor unit.

This study was purposed to deceede the new classification of motor units from EMGs Power Spectra, because surface EMG signals have been much difficult to analyzed about the motor units or the recruitment of these units. EMGs signals have been thought to be complex waves that are contained many components of various motor units. Those interference waves have not effective to classify about various motor units during dynamic exercise.

METHOD

Cranking of the upper limb and pedalling of the lower limb during the dynamic exercises were applied for twelve healthy men to collect data of EMGs. Surface bi-polar electrodes were used to the Deltoides, the Triceps Brachii, the Biceps Brachii, and the Rectus Femoris. EMGs signals of those muscle groups were processed to get Power Spectra in the process of the Fast Fourier Transform, and then a Auto-Regression Model(ARM) were calculated about these Spectra waves (referred to Fig.1) by the measure of PC9801NF(NEC).

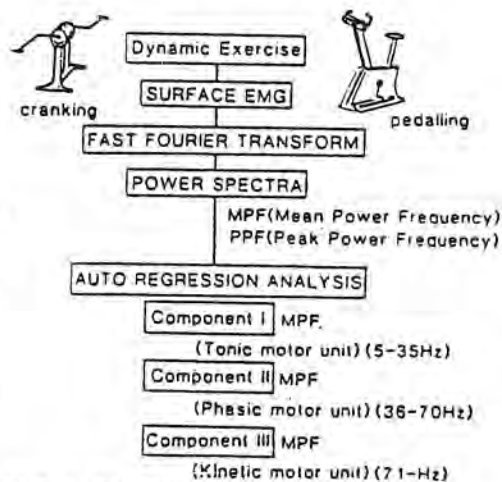


Fig.1 Block diagram of the Auto-Regression Analysis

RESULTS

As shown at Fig.2 and Fig.3, the new divided-components were appeared the motor unit recruitment or synchronization during intensities of 65 watt (cranking) and 175 watt (pedalling) in the Power Spectra. During low intensities (5 watt of cranking and 25 watt of pedalling) of those exercise, only single component of the motor unit were showed in upper charts of those figures. Surface EMGs Power Spectra have changed in linear relationships in accordance with the ramp load increasingly. In lower parts of those figures, light load (5-15 watt of cranking and 25-50 watt of pedalling) added to the limb during the exercise, only one motor unit were proposed in EMGs Power Spectra, while, heavy load (over 65 watt of cranking) and over 175 watt of pedalling added to these leg or hand at the lower side of these chart, two or three motor units were appeared in these Power Spectra.

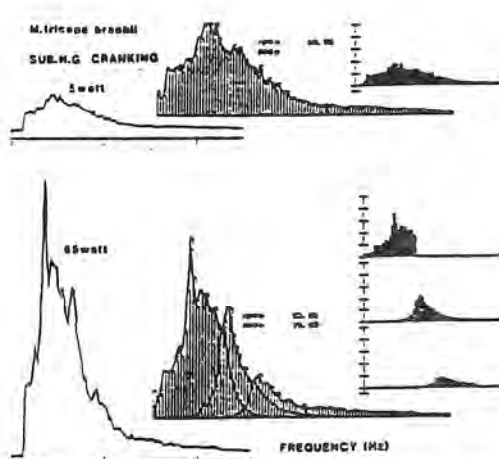


Fig.2 Auto-Regression Analysis on EMG power spectra during the cranking

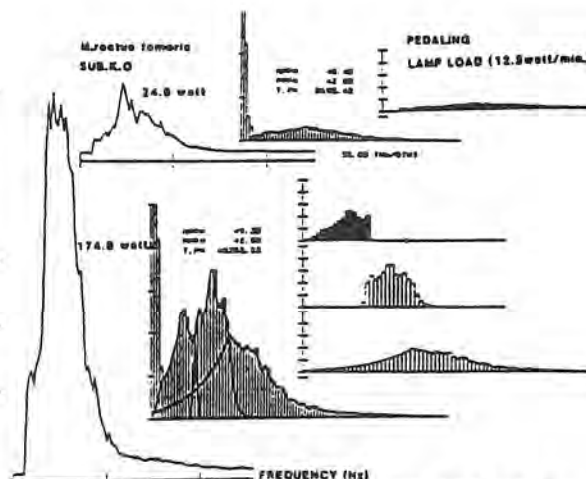


Fig.2 Auto-Regression Analysis during the pedalling movement

DISCUSSION

New analysis were practiced with application of the Auto-Regression Model upon the surface EMGs Power Spectra. Those complex Spectra were divided to get a simple component, and then these newclassified elements should be identified to the new motor units, according to increasing intensities or loads, during the ramp exercise.

Those units were dignified with the frequency band of Power Spectram to be three components of similar motor unit : the low (5-35Hz), the middle (36-70Hz), the high (71Hz-) frequency bands. The low frequency band might be corresponded to the tonic motor unit (slow-twitch muscles), and the middle and the high frequency band might be corresponding to phasic or kinetic units (fast-twitch muscles). The low frequency element should be depended upon contractive function of anti-gravitied or posture muscles of the body. And the middle and high frequency components should be identified to contractive function of the fast-twitch muscle (FOG fiber-type) and more faster and more heavy load-twitch muscles (FG fiber-type).

It is much difficult for surface EMGs to analyze at recruitment of the motor unit similar to needle EMGs signals, while to surface the EMGs Power Spectra, those simulation should be possible to divide the new motor units with ARM, mathematically. In future, comparing experiments would be practised between this new motor unit and the classic motor unit.

CONCLUSION

This ARM analysis were appeared to have possibility to adapt a new-type classification upon surface EMGs Power Spectra, which were collected according to ramp loads and intensities from the external stimulus. Then, in the this experiment three sorts of frequency bands were divided to get the tonic (low freq.), the phasic (middle freq.) and the kinetic (high freq.) motor units, that should be depended upon two classic motor units of slow and fast twitch components during muscle contractive patterns.

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EFFECTS OF LOAD CARRIAGE ON THE LOWER LIMBS

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INTRODUCTION

Manual Material Handling encompasses a long list of activities which Man performs to move objects from one locality to another. In spite of widespread automation in the industries, people still perform many tasks either out of necessity or by choice, involving lifting at the stationary position or carrying loads over varied distances. Such activities can result in high forces and moments being developed in the lower limbs, and these have been thought to be responsible for many accidents and injuries.

The objective of this study (Ang, 1987) was to investigate the effects of carrying a frontal load equivalent to 20% body-weight on the magnitude of the ground reaction forces and moments, intersegmental forces and moments of the knee and hip joints and the range of motion of these joints in the coronal and sagittal planes during gait initiation and rhythmic stages of level walking.

METHODOLOGY

Ten subjects were recruited for this study. Of these, five were males. All volunteers were healthy individuals with no history or evidence of neurological or orthopaedic dysfunctions in the lower limbs.

The instrumentation used in this study is the Vicon television system and a Kistler force measurement platform. The configuration of this television system comprises of 3 cameras arranged orthogonally, with one front camera and 2 lateral cameras. In order to provide maximum camera field from foot to pelvis, the cameras and lenses are positioned at knee level.

The television system and the force platform are interfaced with a PDP 11/34 mini-computer. The digital data collected from both the Vicon system and force platform are transferred to a Vax-mainframe computer for analysis. The resident programmes in the mainframe calculates the ground reaction forces, joint intersegmental forces and the joint angles during the entire cycle of a walking routine.

RESULTS AND DISCUSSION

The parameters where significant increases were detected during frontal load carriage are presented in Table 1 and Table 2.

At the rhythmic stages of walking, the right leg alternates between a stance and a swing phase. During the stance phase, the foot first struck the force platform ahead of the body creating a backward thrust which retarded the forward momentum of the body. This resistive force lasted from the heel strike until the centre of gravity of the body moved in front of the supporting foot. At heel strike, with the 20%BW load, the knee and hip joints had to flex more to maintain a higher backward thrust to prevent the higher inertia comprising of both the body mass and a added 20%BW frontal load from causing the trunk to move to far forward beyond the base of the foot, thereby off-balancing the subject making him fall-over.

During gait initiation, the knee and hip moved inward in similar abduction movements. The shifting of the c.g. towards the swing limb allow the subject to make effective use of the gravity's pull to enable him to take the first step. With added frontal load, there was more abduction developed in the knee necessary to shift the body c.g. towards the swing limb in the early stance phase. At the end of the stance phase, a higher hip adduction was developed to drag the body c.g. back towards the stance limb.

It is obvious from this study that modification in the walking gait both in the initiation and rhythmic stages has to be effected to accommodate to the extra demands on the musculo-skeletal system introduced by the 20% body weight frontal load.

REFERENCE

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Table 1

PARAMETERS WHERE SIGNIFICANT INCREASES WERE DETECTED DURING GAIT INITIATION

PARAMETERS	MALE		FEMALE	
	NORMAL INITIATION	WITH LOAD CARRIAGE	NORMAL INITIATION	WITH LOAD CARRIAGE
MAX. FORWARD GRF FX (+ve)N/H	0.44	0.78	0.23	0.30
MAX. UPWARD GRF FY (+ve)N/H	1.05	1.24	1.06	1.22
MAX. KNEE FLEXION MOMENT (Nm/kg)	0.74	1.06	0.74	1.03
MAX. HIP EXTENSION MOMENT (Nm/kg)	1.57	1.89	1.47	1.74
MAX. HIP ADDUCTION MOMENT	0.87	1.20	0.77	0.94

NOTE : ALL UNITS EXPRESSED IN TERMS OF BODY WEIGHT

PARAMETERS WHERE SIGNIFICANT INCREASES WERE DETECTED DURING RHYTHMIC STAGES OF WALKING

Table 2

PARAMETERS	MALE		FEMALE	
	NORMAL WALKING	WITH LOAD CARRIAGE	NORMAL WALKING	WITH LOAD CARRIAGE
MAX. FORWARD GRF FX (+ve)N/H	0.23	0.27	0.22	0.26
MAX. BRACING GRF FX (+ve)N/H	0.20	0.26	0.19	0.27
MAX. UPWARD GRF FY (+ve) N/H	1.17	1.42	1.15	1.42
MAX. KNEE FLEXION MOMENT (Nm/kg)	NS	NS	0.79	1.28

NOTE : ALL UNITS EXPRESSED IN TERMS OF BODY WEIGHT

Flexibility of the Human Knee under Pure Varus/Valgus and Axial Moments in vivo

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Knee injuries remain a persistent problem in alpine skiing. To appreciate why this is so, we must first recognize that ligamentous injuries occur when an excessive tibia deformation with respect to the femur causes an excessive ligament strain. Clearly, if the strains are large, an injury results. Knee strength therefore depends on joint stiffness which may vary markedly depending on such variables as muscle contraction and weight bearing³. But unfortunately, present ski bindings only release when the loads developed between the boot and the ski reach a preset level. These loads, however, often exceed the load capacity of the knee joint in its weakest state (ie. muscles relaxed and no weight bearing)³. A new approach to binding design must be followed if knee injuries are to be prevented. This approach calls for basing release decisions on knee deformation in conjunction with externally applied loads.

To provide the information base necessary for such designs, relationships between external loads and joint deformation are required. Since in injury situations loading is the independent variable and deformation across the knee is the dependent variable, this study concentrates on flexibility relations rather than stiffness relations. In considering which loads to study, axial and varus/valgus moments are thought to account for the majority of injuries². Hence, the objectives of the present work are to first construct the equipment necessary to measure flexibility and second, to obtain knee flexibility of six test subjects in pure varus/valgus and axial moments.

METHODS

To obtain joint deformation under pure varus/valgus and axial moments, we developed a relative motion three degree-of-freedom goniometer and an instrumented load stand (see Figure 1). The three degree-of-freedom goniometer measures flexion/extension, varus/valgus and external/internal angulation using three miniature potentiometers which are oriented according to the coordinate system that Grood and Suntay (1983) present. The benefit behind selecting this system is that joint motions are obtained sequentially independent of the order in which they occur. To decouple the inherent joint translations while transmitting the rotations between the tibial and femoral attachment clamps, a dual-parallel linkage was designed. This linkage simultaneously decouples medial/lateral and anterior/posterior translations while sliding hardened steel rods decouple distal/proximal translations. Specially designed fixation clamps attach the goniometer to the leg at the tibial crest, the medial and lateral epicondyles, and the thigh.

The instrumented load stand supports the test subject and has the capability of either varying and/or isolating the five variables which affect knee strength (muscle contraction, weight bearing, knee flexion angle, hip flexion angle, and rate of loading) while simultaneously applying either pure varus/valgus or axial moments to the right knee. By definition, pure moments are those which minimally create unwanted shear forces at the knee and are desirable in isolating which loads contribute to total joint angulation. With the load stand capable of also applying combination varus/valgus and axial moments, isolating the contribution of a load to joint deformation is vital if comparisons between individual and combination moments are to be done. Shear forces were effectively minimized by allowing the foot to translate medial/laterally and anterior/posteriorly via two sliding loading carriages. The application of moments also follows the joint coordinate system of Grood and Suntay (1983); axial moments are applied along the long axis of the tibia and varus/valgus moments are applied perpendicular to both the flexion/extension and external/internal rotational axes. Gearmotors provide the necessary torques while a bicycle seat and backrest prevent movement of the upper body during testing. A removable fiber glass cast fitted to the foot of each test subject transmits the moments to the knee. Knee flexion angle is then varied by lowering the torso such that the leg and loading carriage slide forward. A data acquisition computer and custom software convert the raw data into moments and displacements for later plotting.

We cycled pure varus/valgus and axial moments to the right knees of six male test subjects with the knee at 0° and 45° knee flexion angles. Average and standard deviations for the age, height, and weight were 28.5±6.4 years, 1.75±0.04 meters and 712±45 newtons respectively. Throughout all tests, the test subjects minimized muscle contraction and weight bearing. Varus/valgus tests consisted of cycling between 60 Nm of varus and 60 Nm of valgus over a time span of 60 seconds. During testing, the foot was unconstrained in external/internal rotation to insure that only a pure moment was developed at the knee. Before testing, the data acquisition program obtained the goniometer offsets with the test subject's leg muscles relaxed and his knee joint fully extended. Similarly, axial moments were cycled between 20 Nm of external and 20 Nm of internal axial moments except that the foot was unconstrained in varus/valgus.

RESULTS and DISCUSSION

All flexibility plots showed significant nonlinearities. Figure 2 is a typical example of a varus/valgus flexibility plot for 0° and 45° flexion angles. Note the increase in angulation at 45° flexion over 0° and the

significant amount of hysteresis for a complete load cycle. Figure 3 is a typical example of a cycled external/internal flexibility plot for 0° and 45° flexion angles. Note that the cycled curves closely trace over each other and that hysteresis is minimal. Also note that in external rotation, total deformation decreases whereas in internal rotation, total deformation markedly increases. Seering et al (1980), in an in-vitro study of knee stiffness, applied similar axial moment levels and observed similar external displacement trends to the ones we obtained. Tables 1 and 2 tabulate the average varus/valgus and external/internal displacements respectively of six test subjects at the load reversal points. Both tables also tabulate laxity and compute the percent differences in displacement relative to 0°. We defined laxity as the range of motion up to a predefined slope beyond which the joint becomes significantly stiffer. We determined these slopes to be 0.17°/Nm and 2.0°/Nm for varus/valgus and axial moments respectively.

CONCLUSION

Relieving knee injuries in alpine skiing calls for a new approach to binding design, one where release is based on knee deformation. Required by this new approach is information on knee flexibility. With the present three degree-of-freedom goniometer and instrumented load stand, we were able to obtain the flexibility of the knee for both pure varus/valgus and axial moments. Since the results show that flexibility is flexion angle dependent, knowledge of this relationship may be important in the design of ski bindings which protect against ligamentous injury. To determine this conclusively, flexibility data collected in vivo must be supported by ligament strain data collected under externally applied loads.

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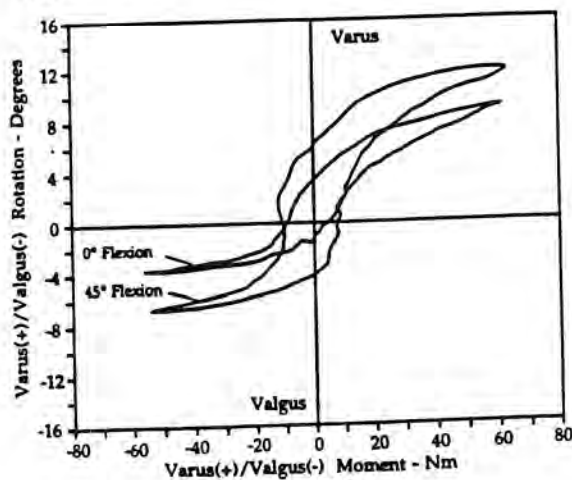


Figure 2.

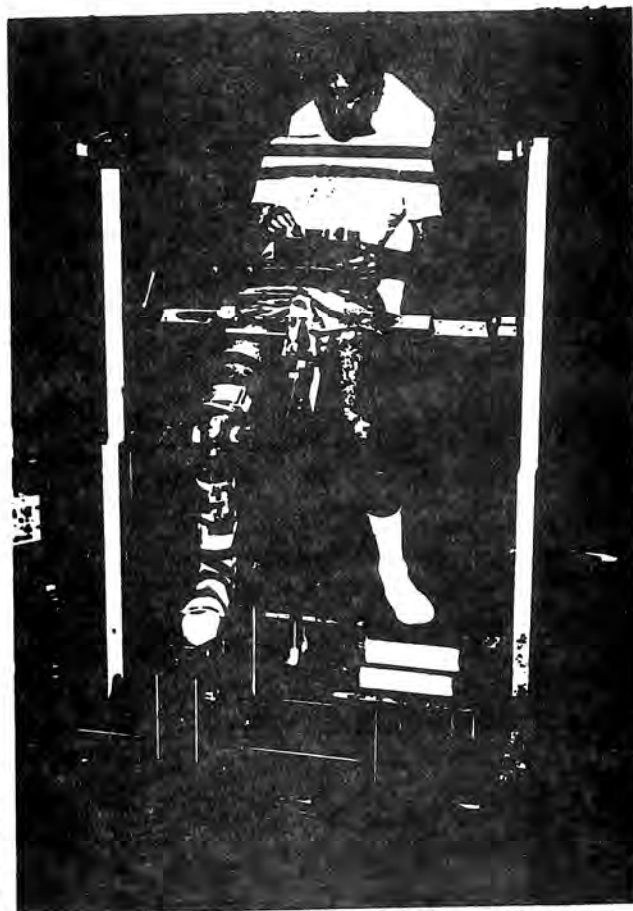


Figure 1. Experimental Apparatus

Table 1.

Flexion	Varus*		Valgus*	
	Laxity	@ 60 Nm	Laxity	@ 60 Nm
0°	3.4°±1.8°	6.8°±1.7°	-3.5°±0.9°	-5.2°±1.4°
45°	8.1°±4.2°	11.2°±2.8°	-5.2°±2.1°	-7.8°±1.9°
% Difference	142.3	64.9	49.7	48.7

* Varus/valgus data are for five subjects.

Table 2.

Flexion	External Rotation		Internal Rotation	
	Laxity	@ 20 Nm	Laxity	@ 20 Nm
0°	12.8°±6.0°	25.6°±2.8°	-4.1°±3.3°	-17.6°±3.8°
45°	8.5°±4.5°	24.3°±2.4°	-7.4°±3.9°	-22.4°±3.2°
% Difference	-33.6	-5.1	81.6	27.4

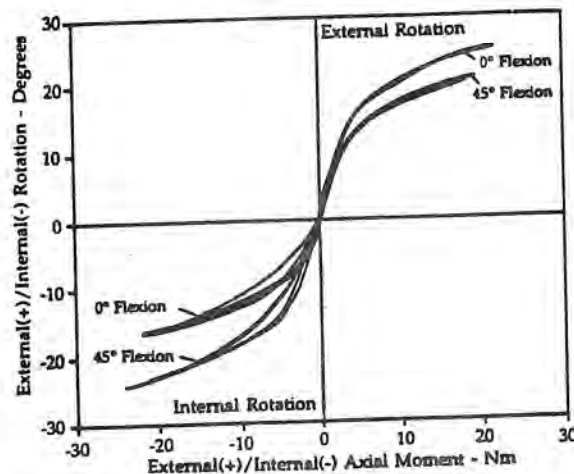


Figure 3.

EFFECT OF CADENCE ON THE EMG OF VASTUS LATERALIS AND OXYGEN UPTAKE DURING STEADY-STATE SUBMAXIMAL CYCLING

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INTRODUCTION

Few studies have reported how cadence and load changes at constant power output affect the integrated electromyographic (IEMG) activity of the leg muscles, particularly the vastus lateralis. Goto et al. (1975) reported that at equivalent power outputs, the IEMG activity of the vastus lateralis increased in a stepwise fashion. Bigland-Ritchie and Woods (1974) found that a linear relationship existed between $\dot{V}O_2$ and IEMG activity during contractions of human quadriceps.

A number of studies have determined that during cycling there exists an optimal pedal rate somewhere in the range of 60 to 100 rpm (eg. Coast & Welch, 1985). This relationship has been derived from data collected during a number of incremental exercise tests at different cadences and then plotting $\dot{V}O_2$ and cadence for equivalent power outputs. It would seem reasonable to conclude that there would be a similar parabolic relationship between the IEMG from the leg muscles and cadence. However, this has not been shown. Thus, the purpose of this experiment was to determine whether activation of the vastus lateralis muscle indicated a minimum level associated with a particular cadence during cycling at a constant power output.

METHODS

Subjects (n=6) completed four successive rides, in random order, on a Monark bicycle ergometer at each of four cadences, 50, 60, 80, and 100 rpm. The resistance was adjusted to maintain a power output of 200 Watts for each ride. Subjects used the output of a handlebar-mounted tachometer to maintain the cadence within 1 rpm of the required cadence. The rider remained seated on the bike and rested between rides until his heart rate was below 80 beats/minute (at least 10 minutes).

$\dot{V}O_2$ was collected at 15-second intervals using the standard open-circuit method with a Beckman gas analyzer. The $\dot{V}O_2$ for each ride was computed as an average of these values over the final five minutes of the ride. The EMG from the right vastus lateralis was recorded via bipolar surface electrodes placed 2 cm apart over the lateral aspect of the thigh, approximately 10 cm proximal to the patella. The amplified EMG signals were sampled through a 12-bit analog-to-digital converter at 500 Hz for a minimum of eight consecutive pedaling cycles. The signal was rectified, integrated and then normalized for each cycle. An average per cycle EMG (CIEMG) was then computed from these normalized data. All values were averaged across subjects and analysis of variance for repeated measures over one factor (either IEMG or $\dot{V}O_2$) was used to determine the effect of cadence. The chosen level of significance was $p < 0.05$.

RESULTS AND DISCUSSION

The results of this study are summarized in Table 1. There was little change in either CIEMG or $\dot{V}O_2$ in spite of the wide variation in cadence. Analysis of variance showed no significant difference at each condition for either the $\dot{V}O_2$ ($F_{3,15}=2.083$, $P = 0.144$) or the CIEMG ($F_{3,15}=2.0$, $P = 0.156$).

The VO_2 and CIEMG data paralleled each other confirming that VO_2 and muscle activation are linked. However, neither variable showed the parabolic trend found by Coast and Welch (1985) and Boning et al. (1984). One explanation may lie in the narrower range of pedal rates used for this experiment. For example, Coast and Welch (1985) used pedal rates that ranged from 40 to 120 rpm. There is a tendency of the VO_2 at 40 rpm and 120 rpm to exaggerate the parabolic effect by shifting the ends of the curve upwards. Thus, the present data may be consistent with a flatter portion in the range of 50 to 100 rpm.

Another explanation may be that there is a fundamental difference in cardiovascular system response to cycling at a constant load versus an incremental load. If this were the case a cyclist performing a series of increments and then maintaining at a submaximal level, say 200 watts, would have a different steady-state VO_2 than if he had only ridden at that 200 watt level. This implies that the metabolic cost of cycling at 200 watts is very much affected by the manner in which the rider achieved that power output. The magnitude of difference in response of the cardiovascular system would likely be exacerbated at loads greater than the anaerobic threshold. Thus, the concept of minimal cadence based on VO_2 may well be protocol specific. Clearly further examinations of the differences in the response to incremental loads versus steady-state loads is warranted.

Table 1. The mean and standard deviations for the per-cycle integrated EMG (CIEMG) and the VO_2 . The CIEMG is in arbitrary units because of the need to normalize to each pedal duration.

Cadence (rpm)		50	60	80	100
VO_2 (l/min)	mean	2.46	2.48	2.36	2.52
	sd	0.23	0.35	0.27	0.27
CIEMG (arb)	mean	0.60	0.57	0.52	0.54
	sd	0.13	0.13	0.17	0.17

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A COMPARISON OF DISTANCE RUNNING KINEMATICS BETWEEN TREADMILL AND COMPETITIVE RUNNING AND BETWEEN GENDERS

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The analysis of distance running mechanics is often hindered by a lack of an adequate sample size. The time involvement and expense linked with kinematic data acquisition can impose limits the number of subjects that can be involved. This paper presents some preliminary information from a database of kinematics for elite distance runners developed over the last few years. Comparisons are made between kinematic measures obtained during laboratory treadmill running and similar data collected during elite level competition, as well as between genders using the laboratory data.

Methods: Results for this study come from a kinematic database consisting of 32 displacement, velocity, and temporal measures for elite male and female distance runners. Films were collected using a LOCAM camera operating at 100 fps and giving a side view of the runner, and the digitized coordinate data were digitally filtered at 6 hz. Data for each kinematic measure were obtained from two cycles of running and averaged. Kinematic measures included maximum and minimum angles, and angles at foot strike and toe off, for the thigh with the vertical, the knee, and the ankle; maximum and minimum angular flexion and extension velocities for the thigh, knee, and ankle; linear velocities for the knee and heel during the running cycle; vertical oscillation; stride length; and cycle, support, and non-support times.

Laboratory data were obtained during five different sessions over a two and one-half year period, with the same analysis procedures used in each session. Subjects ran on the treadmill at speeds varying from 4.13 to 6.31 m/s, with some subjects running only at a criterion test speed (5.96 m/s for men and 5.36 m/s for women) and others running at several speeds within this range. A total of 55 sets of data were available for 33 female runners and 96 sets of data were available for 34 male runners. Films during competitive runs were obtained at the 1987 TAC championships and from the 1988 U.S. Olympic Trials. From these two meets 37 sets of data were available for 28 different female runners and 42 sets of data were available for 34 different male runners.

Comparisons were made between men and women using the laboratory treadmill data over a range of speeds. Differences between the competitive and treadmill data were analyzed for both males and females when comparing across speeds, and only for the males when comparing at the 5.96 m/s speed. Only a few of the runners are included in both treadmill and competitive data sets, and thus the majority of the comparisons made were between sets of different runners. Treadmill data was available at several speeds for the majority of the runners, but at only one speed for some subjects. Competitive data was available for one run for most subjects and for two runs for some subjects. These limitations may affect the generalizability of the data.

Results: Figure 1 shows an example of data comparing men and women across a range of speeds on the treadmill. This figure indicates that the women had less vertical oscillation (relative to leg length) than did the male runners. A similar trend of consistent differences across speeds was found for several variables, including data showing females to have a greater maximal angle of the thigh during hip flexion, shorter support and cycle times, and shorter absolute stride length. Stride length relative to leg length was similar for both genders, as was maximal thigh flexion angle, maximal plantar flexion angle, and maximal linear knee velocity.

Three methods were used to evaluate the data between the treadmill and competitive runs. In one method data from both males and females were compared across a range of speeds, as illustrated in figure 2 for maximal flexion of the knee during the swing phase. A consistently greater knee flexion angle was found during running on the treadmill compared to the competitive situation. Similarly, vertical oscillation was lower on the treadmill throughout the speed range. Stride length

and the maximal thigh extension were similar across the speed range, though there were slight differences at slower speeds. The second method of comparison involved data from a subset of male runners ($N=12$ in competition and $N=34$ on the treadmill) running at a 5.96 m/s speed. Figure 3 provides an example of such comparisons for the angle of the thigh with the vertical at toe off. Among the significant differences ($p<.05$) for data at this speed were results showing runners on the treadmill to have shorter (29.8 ms) support and longer (25.6 ms) non-support times, a smaller angle of the thigh with the vertical at foot strike (2.7 deg) and toe off (5.8 deg), greater knee flexion during swing (6.4 deg) and support (4.8 deg), a shorter stride length (8.7 cm), and a lower vertical oscillation (1.2 cm). The third method involved a direct comparison of data for eight male runners who were involved in both experimental conditions. Figure 4 shows results for maximal knee flexion in swing. Data for different variables for these eight subjects showed results very similar to those just described for the other two methods, adding further support that there are differences between the two experimental conditions.

Discussion: The data across speeds comparing men and women suggests that there may be some gender differences in kinematics. Some of the differences can be explained by differences in size, such as those for absolute stride length, support and cycle times. Differences for vertical oscillation relative to leg length and some of the lower extremity angles cannot be similarly explained. Because data were not available for all subjects across the speed range further study is needed to confirm the differences identified.

There appear to be some consistent differences between the kinematic data obtained under laboratory conditions on the treadmill and data from competition. These data generally agree with previous studies which have found differences between overground and treadmill running, though the element of competition introduces the possibility that some of the differences are due to factors involving the competitive situation.

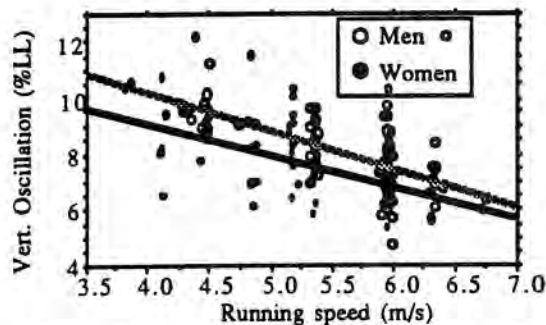


Figure 1. Vertical oscillation (%LL) for male and female distance runners on a treadmill.

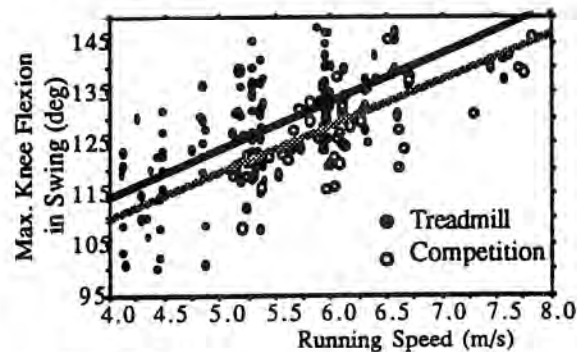


Figure 2. Differences between treadmill and competitive data for maximal knee flexion in swing.

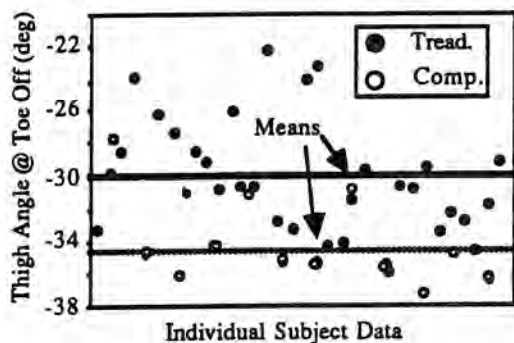


Figure 3. Thigh angle at toe off for groups of different subjects at 5.96 m/s.

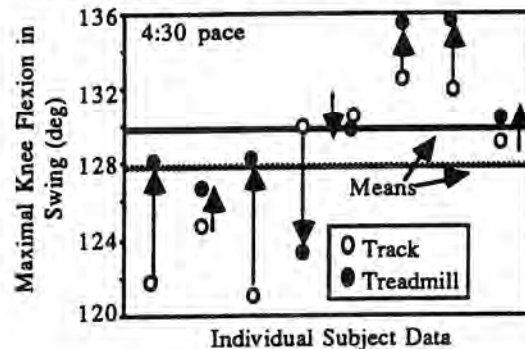


Figure 4. Maximal knee flexion during swing for eight subjects on the treadmill and in competition.

ANALYSIS OF U.S. AND SOVIET EFFORTS TO MAINTAIN MUSCLE AND BONE MASS WITH EXERCISE DURING PROLONGED BEDREST AND SPACEFLIGHT

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Measures taken to prevent disuse bone loss have included ergometer exercise, heel impact loading, static and cyclic longitudinal compression, standing, drug therapy, and lower body negative pressure. None have eliminated the negative calcium balance induced by bed rest (Schneider and McDonald, 1984). The inability of any of these methods to prevent disuse atrophy has led researchers to propose solutions in space which mimic ground-based normal activity. Treadmills currently are the principal high load devices used by both the Soviet cosmonauts and US astronauts during spaceflight (Thornton, 1981; Thornton, 1986). Large and small radius centrifuges are also being considered as a means of producing equivalent gravitational forces (Burton, 1988). The objective of this paper is to use a recently published mathematical model which expresses bone density in terms of the daily stress history 1) to analyze several countermeasures used in bed rest studies and 2) to estimate the effectiveness of the current treadmill devices to apply sufficient loading in space.

MODEL DEVELOPMENT

A mathematical model of bone density has been developed which predicts bone apparent density in bones subjected to patterns and levels of loading which do not vary significantly over long periods of time (Carter *et al.*, 1987). The principal assumptions incorporated into the model are that bone mass is determined by a scalar quantity of the stress state and that all daily loading cycles contribute to bone maintenance. Whalen *et al.* (1988) showed that for bones in which the pattern or orientation of loading does not vary significantly from one activity to the next, the relative difference in bone density between two activity levels is, to a first approximation, a function of only the peak cyclic joint reaction forces. Historically, space life science researchers have considered the calcaneus to be a very important bone to analyze. The model, applied to the calcaneus, takes a particularly simple form. Calcaneal loading is expressed in terms of the daily history of ground reaction forces, G_z , measured at late stance phase when the moment about the ankle is the largest, or

$$\frac{\rho_2}{\rho_1} = \left[\frac{\sum_{j=1}^{k_2} n_{2j} (G_{z2j})^m}{\sum_{i=1}^{k_1} n_{1i} (G_{z1i})^m} \right]^{\frac{1}{2m}} \quad (1)$$

where the numerator and denominator represent the daily loading histories for the two activity states. Based on animal experiments and exercise studies, the stress exponent, m , was estimated to be approximately 4.0 (Whalen *et al.*, 1988). In this study normal daily activity was assumed to consist of 4 hours of normal daily walking which produced cyclic peak ground reaction forces of 1.2 body weight (BW) and 12,240 load cycles to each leg per day. Relative bone apparent density of the calcaneus, compared to a normal baseline of 4 hours daily walking, can be estimated from equation (2)

$$p/p_n = (n/12,240)^{\frac{1}{8}} (G_z/1.2)^{\frac{1}{2}} \quad (2)$$

where n is the number of daily exercise loading cycles and G_z is the peak cyclic ground reaction force produced by the exercise. To maintain bone mass with only treadmill exercise, the

following relationship must hold

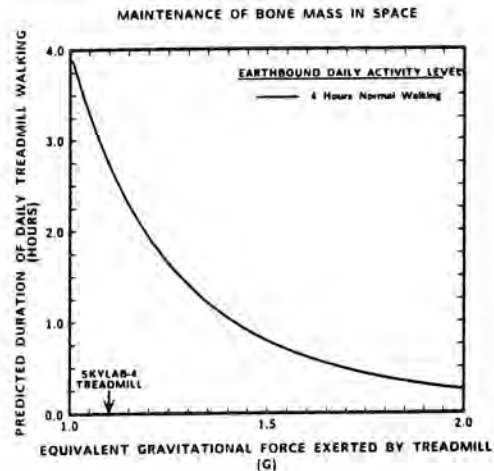
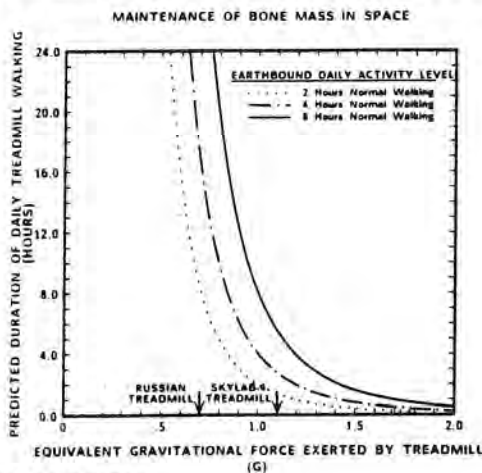
$$t = t_n (W_n/W)^4 \quad (3)$$

where t is the duration of treadmill walking, t_n is the duration of walking on earth, and W/W_n is the equivalent gravitational loading in space compared to earth. This relationship assumes that the loading frequency is the same for treadmill walking and earth-bound walking.

RESULTS

Bed rest activities analyzed were 4 hours of ergometer exercise, and 3 hours of passive standing (Issekutz *et al.*, 1966). Calcaneal bone apparent density estimates, compared to normal, were 39 % for cycling, and 60% for standing. These significant differences concur with the finding that urinary calcium losses were somewhat reduced during standing, but unchanged during ergometer exercise.

The application of equation (3) to treadmill walking in space is shown in Figures 1a and 1b.



DISCUSSION

The low level of forces generated in the lower limbs by the above exercises have been reported elsewhere (Thornton, 1981; Thornton, 1986), and have been suggested as a probable cause for failure of the regimens to maintain bone mass. The importance of this work is that an estimate for duration of equivalent loading, which can be tested in bed rest studies, has been made based on a mathematical model derived from a tissue level stimulus. The non-linear nature of the prediction is evident from Figures 1a and 1b which show that 15 minutes of walking at $2g$ provides the same level of stimulus as 22 hours of walking at $.7g$. The impact of this non-linear relationship on dose-response estimates between load magnitude and bone density has not been appreciated by the space community.

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THE CONCORDANCE BETWEEN MILD LEG LENGTH INEQUALITY AND GROUND REACTION FORCE DATA ASYMMETRY DURING RUNNING

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The effect of leg length inequality (LLI) on the mechanics of posture and gait is not clearly understood. A link has been suggested between mild LLI (absolute limb length difference < 3 cm) and the etiology of unilateral overuse injuries such as stress fractures and low back pain. The rapid loading that occurs during the initial period of foot contact when running has also been linked with these injuries. An asymmetry in parameters describing the loading has been reported. The purpose of this study was to assess the concordance between LLI and bilateral asymmetry for the vertical ground reaction impact force during running.

METHODS

Three male runners (training more than 20 km/week for more than 2 months) volunteered as subjects. All were injury free during the two months prior to testing, and none of the subjects was aware of a LLI. Ground reaction force data were obtained using a Kistler force platform (sampling frequency = 500 Hz) mounted flush with the floor in the middle of a 30m runway. Each subject completed two sessions (one week apart) of force data collection. 25 trials were collected for each foot/session. Foot order was counter-balanced among subjects and across days. Each subject ran at a speed slightly slower ($+30$ s/km) than his usual training pace. Speed was monitored over two 2m intervals using a photoelectric timing system. In addition to speed, 4 other criteria were used to ensure that each trial was representative of "normal" performance. Four parameters defining the loading during the initial period of contact (impact force magnitude, absolute time to the impact force peak, impulse from foot contact to impact force, and loading rate) were evaluated for all vertical ground reaction force curves.

LLI was assessed using femoral head radiography. Each subject was barefoot and wore a protective gonadal shield. Subjects were positioned in a "parallelogram posture", with the medial malleoli separated by 186 mm, equal calcaneal stance with both feet, knees and hips in extension, and no lateral sway. This technique provides a measure of LLI as the absolute difference in the height of the superior surface of the femoral heads. Two radiographs were taken of each subject, and LLI was measured twice from each radiograph by each of two testers using two different rulers. The mean measurement error between and within-testers was found to be less than 1 mm (McCaw et al, unpublished data). For this study, a LLI was operationally defined as a measured difference greater than the 1 mm mean measurement error.

RESULTS AND DISCUSSION

The magnitude and direction of the LLI for each subject is presented in Table 1. The direction (limb with greater value) and the inter-day mean bilateral differences for the 4 parameters defining initial impact loading are given in Table 2.

Table 1. Magnitude and Direction of LLI

Sub #	mean delta	Longer Limb
1	9	right
2	1	right
3	1	left

Units: delta in mm

Table 2. Direction of GRF Asymmetry and Mean Difference.

Sub	Day	IFM	AT	Imp	LR
1	1	R	L	R	R
1	2	R	R	R	R
	M	2.18	--	0.05	32.49
2*	1	-	-	-	R
2	2	R	L	R	L
	M	--	--	--	--
3	1	R	L	R	L
3	2	R	L	R	L
	M	1.02	0.001	0.01	14.97

IFM : Impact Force Magnitude (N/kg)

AT : Absolute Time to Impact Force Peak (s)

Imp : Impulse to Impact Force (N/kg * s)

LR : Loading Rate (N/kg/s)

M : Mean difference

* : No impact force evident on Day 1

Table 1 data indicates that only S1 had a LLI greater than the 1 mm measurement error. The range of LLI measured for this subject was 8 to 10 mm. For S2 and S3, the ranges were 0 to 1 mm and 0 to 2 mm, respectively. For both these subjects, the mean delta did not exceed the 1 mm criterion established for LLI. Table 2 data indicates that the two-session mean parameter values of the longer right leg of S1 were greater than his left leg values. For S2 and S3, the observed inter-day bilateral asymmetries were inconsistent in direction and of lesser magnitude than those of S1. These results suggest a concordance between LLI and impact loading asymmetry in runners. Future research should include the evaluation of joint dynamics and the effect of foot orthotic devices for reducing the magnitude of bilateral asymmetry among runners with LLI.

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WALKING VS RACEWALKING: 3D KINEMATIC PATTERNS FOR THE TRUNK

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NOMENCLATURE

AX, XT	rotation about, translation parallel to the anterior-posterior axis (X)
AY, YT	rotation about, translation parallel to the bilateral axis (Y)
AZ, ZT	rotation about, translation parallel to the vertical axis (Z)
RFC, LFC	right (start-stop of stride cycle), left foot contact
RFO, LFO	right, left foot off
CW, CCW	clockwise (positive), counterclockwise (negative) rotation
DS, SS	double, single support

INTRODUCTION

Racewalking (competition walking) is a very special mode of locomotion—the gait being a function of the rules which govern the sport. A legal racewalk gait requires constant contact with the supporting surface at all times, and a DS period must be established during each stride cycle. The rules also state (International AAF Handbook, 1984) that the supporting leg be "straight in the vertically upright position". It is obvious, even to the untrained eye, that these gait constraints cause changes to the kinematic patterns of the trunk and limbs. Various aspects of racewalking biomechanics have been investigated by: Cavagna and Franzetti (1981), Cairns *et al.* (1986), etc. However, to our knowledge, no studies have looked at the 3D kinematics of the pelvis-thorax during racewalking. In this study we have focused on the 3D spatio-temporal patterns obtained from the rotation/translation variables (AX, AY, AZ, XT, YT, ZT) of the pelvis-thorax for a competitive racewalker. More specifically, the purpose of this study was to identify variable pairs which changed and which *did not* change from walking to racewalking at the same speed.

METHODS AND PROCEDURES

LEDs (light emitting diodes) were attached to triangular plates fitted to the posterior of the pelvis and thorax via a waist belt and shoulder harness. The position of each LED was tracked by a SELSPOT system using two cameras with time-division multiplexed analog inputs of foot off/foot

contact while the subject was walking on a precision speed control treadmill. The equipment, calibration procedure, and data acquisition-analysis were described Stokes, *et al.* (1988).

A world class female racewalker, well trained on the treadmill, was the single subject over a period of three consecutive days. A total of 153 trials (approx. 10s each) were stored on magnetic tapes for later off-line analysis. After several practice treadmill sessions three sets of speeds were chosen: (1) preferred walking, 1.50ms^{-1} ; (2) walking/racewalking, 2.00, 2.10, 2.20, 2.30, 2.40, 2.50ms^{-1} ; (3) preferred racewalking, 3.00ms^{-1} . Repeated trials on the same day and different days were used to assess subject/system variability.

Seven contiguous stride cycles were chosen from each trial for analysis. These data segments were detrended recursively on consecutive

cycles for subject drift and normalized to 100%. The rotations and translations with respect to the anatomical positions of the segment plates (Fig. 1) were calculated from the normalized and ensemble averaged stride cycles. These data were used to generate bivariate parametric curves (66 at each speed). In addition, the SNR (signal-to-

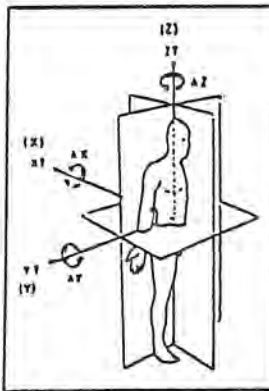


Fig. 1. Reference position.

noise ratio), first and second derivatives were computed for each variable.

RESULTS AND DISCUSSION

SNR estimates with displays of variables and their derivatives indicated low subject/system variance and high homogeneity of patterns over repeated trials.

The bivariate parametric plot in Fig. 2 showed differences between walking and racewalking at the same speed, especially during DS. A CW AZ

(thorax) rotation with respect to reference position was measured during walking at all speeds for the first DS phase and CCW during the second DS phase. Racwalking showed both CW and CCW AZ (thorax) rotations with respect to the reference position during each DS phase. Also there was a tendency for AX (pelvis) and AZ (thorax) to approach zero simultaneously (i.e. return to their reference position) at mid-DS during racewalking. Note that as preferred racewalking speed was approached "kinks" began to form in the racewalking patterns. Of

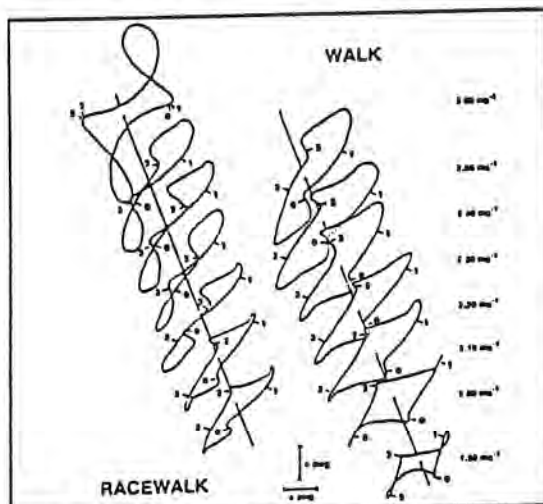


Fig. 2. AX(pelvis) vs AZ(thorax) for different speeds. 0—RFC, 1—LFO, 2—LFC, 3—RFO. DS: 0-1, 2-3; SS: 1-2, 3-0.

course, we expected symmetry of trunk patterns since rotations and translations exhibit glide-reflection symmetry (Stokes, *et al.* 1988); but the symmetry of the preferred walking and racewalking bivariate plots was striking (glide-reflection-reflection with respect to any line through the centroid of the pattern).

Fig. 3 shows that AZ (pelvis) rotation was closely synchronized with YT (pelvis). Their minimum/maximum occurred simultaneously during DS. At preferred walking and racewalking speeds the bilateral translation and vertical rotation of the pelvis were never zero simultaneously; i.e. anatomical position was not realized for preferred speeds.

The racewalking kinks in Fig. 2 could have clinical importance; i.e. hip musculature problems for racewalkers may be related to these intersegmental twists imposed on the trunk. The synchronization of AZ (pelvis) and YT (pelvis) suggests dependency between these variables. Understanding this dependency and their invariance across locomotion modes/speeds could have important neural control consequences. Also, I conjecture that variable pairs which

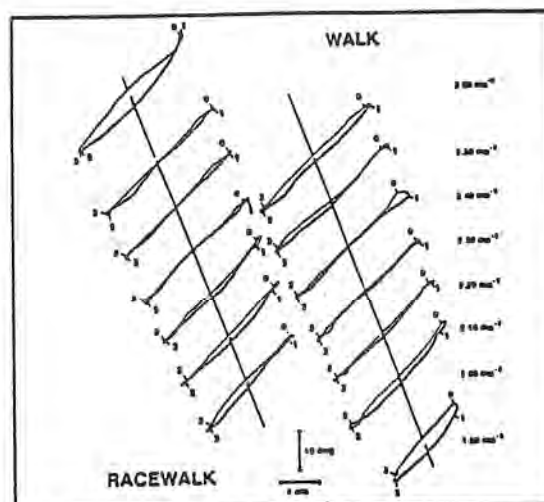


Fig. 3. AZ(pelvis) vs YT(pelvis) for different speeds.

show glide-reflection symmetry across a stride cycle will exhibit glide-reflection-reflection symmetry in their bivariate parametric plots. There was a tendency for asymmetry at nonpreferred walking and racewalking. It has been hypothesized that preferred walking speed is the minimum energy cost speed (walking—Cavanagh and Williams, 1982; racewalking—Morgan and Martin, 1986). Perhaps preferred locomotion speed is equivalent to the minimum energy cost speed. Furthermore, I suggest that the minimum energy costs are directly related to symmetry of trunk movement, since symmetry admits reduction in control costs. This could be relevant to the design and control of bipedal robots.

Bivariate parametric plots of 3D kinematics can be useful in obtaining phase and symmetry relationships between variables of interest. Future extensions to this project will include more subjects and additional locomotion modes.

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TRANSCRANIAL MAGNETIC STIMULATION OF THE HUMAN BRAIN: EFFECTS OF STIMULUS LOCATION

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INTRODUCTION

The amplitude and spatial distribution of electromyographic responses evoked by transcranial magnetic stimulation of the motor cortex are critically dependent upon the site and intensity of stimulation. In the present investigation we sought to establish a standardized procedure for locating the stimulating coil electrode in such a way as to optimize the appearance and amplitude of the motor evoked potentials (MEPs) in target muscles viz the muscles of the thenar eminence and the tibialis anterior. The location of the stimulating electrode on the scalp was referenced to a grid, taking its origin from the vertex. By varying the location of the stimulating coil we were able to establish the anatomical distribution of MEPs.

METHODS

Ten informed healthy adult subjects participated in the study. Magnetic stimuli (70% intensity) were generated from a Cadwell MES-10 Stimulator and delivered to the scalp through a 9 cm diameter coil electrode. In each case the current passed in a clockwise direction around the coil. The electrode was positioned with its center located at select points on a grid. The grid represented 9 points, positioned 4 cm apart on an 8cm x 8cm rectangular matrix centered over the vertex (Cz in the conventional International 10-20 system for electroencephalography). Electromyographic recordings were made bilaterally from the thenar eminence (Abductor Pollicis Brevis, Flexor Pollicis Brevis) and from the Tibialis Anterior muscles. Signals were led off from surface electrodes in a bipolar configuration, amplified (3dB down 10-1kHz) in Disa Type 15 C amplifiers, A/D converted at 2000 Hz and then stored and processed for latency and amplitude of the MEP in a Tektronix 6130 microcomputer.

RESULTS

MEPs were evoked bilaterally in thenar muscles when stimulation was applied at midline locations on, or anterior to, the vertex. The distribution and reproducibility of these responses in consecutive stimulations are shown in Figure 1. Quite frequently there were asymmetries in the amplitude of the evoked response; this may reflect hemispheric variation, differences due to direction of coil current relative to the underlying tissue, or differences in the peripheral neuromuscular apparatus. There were also marginal differences in the latency of responses evoked bilaterally; for example, the latency of MEPs recorded from the anterior midline location illustrated in Figure 1 differed: left $\bar{x}$ =22.2 ms; right $\bar{x}$ =23.5 ms. The waveform configuration also varied, with the occasional appearance of long duration polyphasic potentials, reflecting temporal dispersion in the activation of motor units.

MOTOR EVOKED RESPONSES IN THENAR MUSCLES: EFFECTS OF STIMULUS LOCATION

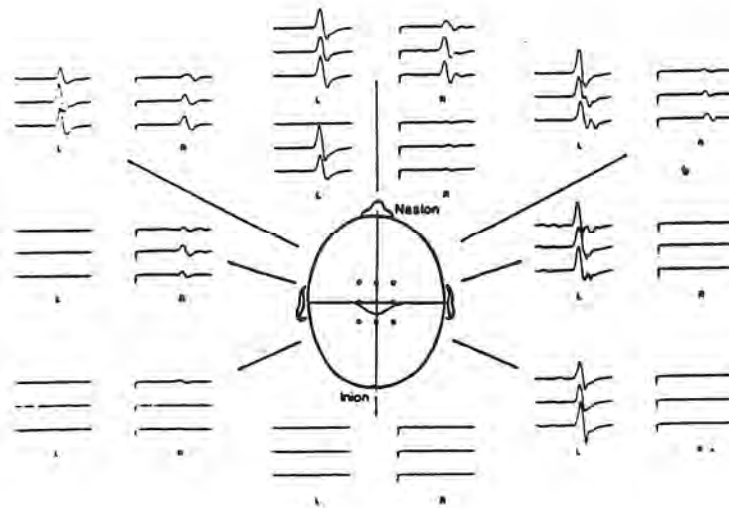
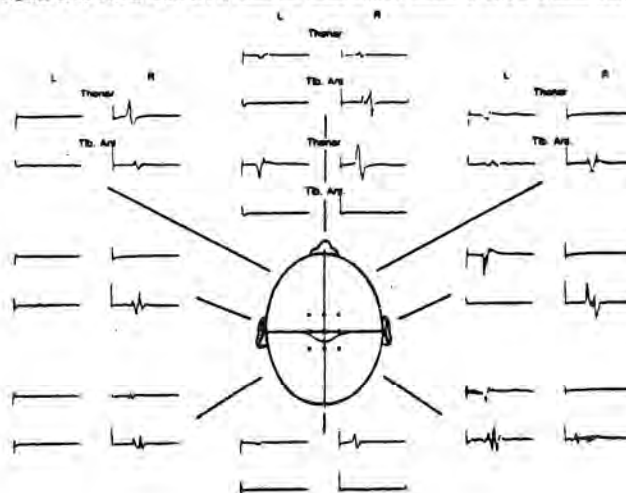


Figure 2 shows averaged MEPs recorded bilaterally from upper (thenar) and lower limb muscles (tibialis anterior). The latency of thenar MEPs in this male subject (aged 28 years) was $\bar{x}=23$ ms compared to $\bar{x}=32$ ms for the tibialis anterior. With the stimulus intensity at 70%, and with relaxed muscle, it was only possible to evoke low amplitude responses $\bar{x}=0.5$ mV in tibialis anterior. Involuntary muscle activity potentiated the MEPs.

AVERAGE MOTOR EVOKED RESPONSES IN THENAR AND TIBIALIS ANTERIOR MUSCLES: EFFECTS OF STIMULUS LOCATION



CONCLUSIONS

1. MEPs may be evoked in select combinations of muscles depending on stimulus location.
2. Stimulus location at, or anterior to, the vertex generally yield higher amplitude MEPs in tibialis anterior.

(Funded by The Spinal Cord Society of Canada)

ELECTROMYOGRAPHIC RESPONSES IN UPPER EXTREMITIES INDUCED BY MAGNETIC BRAIN STIMULATION

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INTRODUCTION

Motor evoked potentials (MEPs) elicited by transcranial magnetic stimulation of the motor cortex are facilitated by contraction of the target muscle, contraction of other muscles of the ipsilateral limb, and contraction of muscles in the contralateral limb (Hess et al, 1986). This facilitation of MEPs is an important consideration when employing brain stimulation to identify the integrity of corticospinal pathways in patients with CNS disorders (cf Hayes et al, 1989) and for elucidating mechanisms of motor control in normal subjects.

The present investigation was undertaken to determine the degree of facilitation of MEPs in forearm musculature associated with ipsilateral and contralateral isometric wrist flexion contractions of varying percentage of maximum voluntary contraction (MVC). Of interest were the questions of whether the antagonist muscle (extensor carpi ulnaris) MEP is facilitated by contraction of an ipsilateral agonist muscle (flexor carpi radialis), and whether contraction of the contralateral limb wrist flexor muscles induces facilitation of both the flexor and extensor MEPs of the ipsilateral limb.

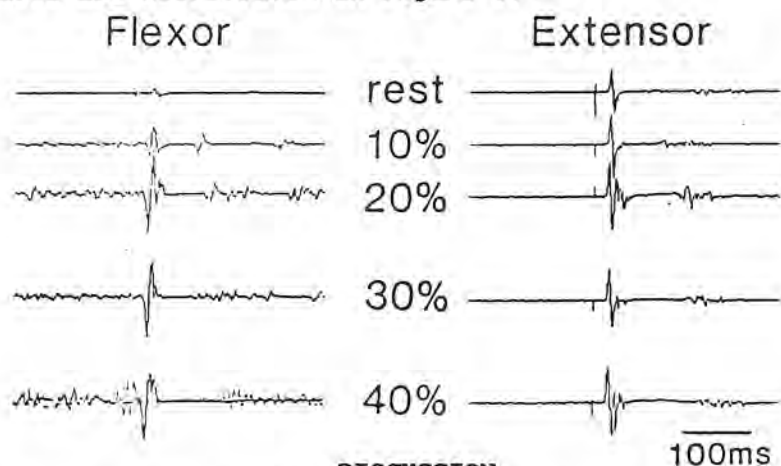
METHODS

Six informed healthy adult subjects (aged 24-42 years) participated in the study. They performed isometric wrist flexion contractions of 10%, 20%, 30%, 40% and 100% MVC, unilaterally and bilaterally. Force was measured using a Shaevitz Load Cell and displayed to the subject on an oscilloscope. Magnetic stimuli (~70% intensity) were generated from a Cadwell MES-10 Stimulator and delivered to the scalp through a 9 cm diameter stimulating coil located over the vertex. Stimuli were delivered while the subject sustained the isometric contractions. Electromyographic (EMG) recordings were made from the flexor carpi ulnaris and the extensor carpi ulnaris. The myoelectric signals were recorded from surface electrodes in a bipolar configuration, amplified (3dB down 10-1kHz) in Disa Type 15 C amplifiers, A/D converted at 2000Hz and then stored and processed for latency and amplitude of the MEP in a Tektronix 6130 microcomputer. The EMG signals were also feedback (visually) to the subject through the oscilloscope.

RESULTS

During the unilateral wrist flexion contractions MEPs were evoked bilaterally in wrist flexor and extensor muscles. Increasing intensity of the force of contraction resulted in a linear increase in amplitude of the MEP in the ipsilateral flexor muscle, as has been shown previously (Hess et al, 1986). The mean latency of response also decreased from 16.9ms at 0% MVC to 15.4ms at 40% MVC. There was a modest but consistent increase in amplitude of the MEP of the

extensor muscle with increasing flexion force. Close inspection of the background EMG of the extensor muscle revealed low level but monotonically increasing amplitude ($1/t \int |EMG| dt$) of the interference pattern associated with increased flexion force levels. This was despite the attempts by the subjects to use biofeedback to maintain silence in the extensor muscles. The increased amplitude of extensor MEPs was therefore associated with the cocontraction of flexor and extensor muscles per se rather than through independent facilitation of the alpha motoneuron pool to the extensor muscle. A silent period followed each MEP and occurred principally during the phase of increasing tension development of the evoked muscle contraction. The silent period terminated with a brief duration asynchronous burst of muscle action potentials. The duration of the silent period was shorter in the contracted flexor muscle than in the minimally active extensor. These results are illustrated in Figure 1.



DISCUSSION

These results establish that MEPs may be simultaneously evoked in both flexor and extensor muscles of the forearm. The presence of increasing levels of background contraction in the flexor muscle does not lead to a progressive reciprocal inhibition of the extensor MEP. Instead, the amplitude of the extensor muscle MEP is unaltered; or it is increased if there exists a cocontraction of the extensor muscle.

The present results also show the duration of the silent period that follows the MEP to be limited, at least in part, by the presence of the background contraction. This is consistent with the view that the duration reflects the temporal properties of the competing inhibitory inputs (Renshaw cell and GTO), motor unit refractoriness, cessation of Ia afferent discharge, and the excitatory synaptic input of the voluntary drive to the motoneuron pool.

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REINFORCEMENT OF MOTOR EVOKED POTENTIALS IN SPINAL CORD INJURED PATIENTS
FOLLOWING TRANSCRANIAL MAGNETIC STIMULATION OF THE MOTOR CORTEX

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INTRODUCTION

Transcranial magnetic stimulation of the motor cortex provides a new and painless procedure for investigating properties of the human motor control system (Barker et al, 1985). In particular the procedure allows determination of the integrity and conduction properties of the cortico-spinal pathways. Recent investigations pertaining to the methodology and clinical application of magnetic stimulation of the brain have illustrated the nature and spatial distribution of the electromyographic responses (Rothwell et al, 1987; Miyagawa et al, 1989) and the diagnostic utility of these "motor evoked potentials" or MEPS (Mills et al, 1987).

In the present study we examined the influence of various methods of reinforcement of MEPS in subjects with spinal cord trauma. Our early observations that MEPS could not be readily elicited in paralysed muscles or even in muscles with some preserved voluntary motor function, prompted us to try to find methods of neurological reinforcement likely to maximize the probability of detecting MEPS.

METHODS

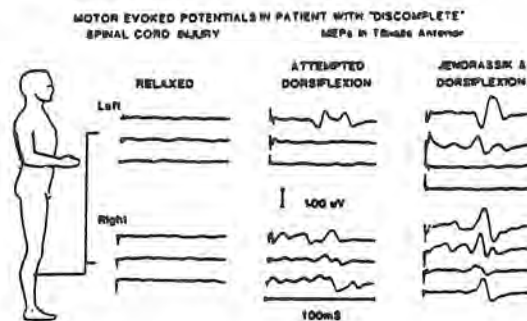
Eighteen informed volunteer subjects (3 female; 15 male) ranging in age between 8-58 years participated in the study. Twelve subjects had sustained lesions to their cervical cord and six had thoracic cord injuries. All subjects underwent electrodiagnostic evaluations of their peripheral nerves to rule out root avulsion, denervation, atrophy or neuropathy.

Magnetic stimuli (~70% intensity) were generated from a Cadwell MES-10 Stimulator and delivered to the scalp through a 9 cm diameter coil positioned at various points on a vertex referenced grid system. The grid, which takes its origin from the International 10-20 system convention for electroencephalography, enables location of the stimulating coil with some confidence over the motor cortex. Electromyographic (EMG) recordings were made from the thenar eminence (Abductor Pollicis Brevis, Flexor Pollicis Brevis) in quadriplegic subjects and from the Tibialis Anterior muscle in both quadriplegic and paraplegic subjects. EMG signals were recorded from surface electrodes in a bipolar configuration, amplified (3dB down 10-1kHz) in Disa Type 15 C amplifiers, A/D converted at 2000 Hz and then stored and processed for latency and amplitude of the MEP in a Tektronix 6130 microcomputer.

The methods of reinforcement included passive joint displacement to induce muscle stretch, 100Hz vibration of muscle (Sammons BK 5206 vibrator), handgrip, shoulder shrug, jaw clench (ie remote muscle contractions - Jendrassik maneuver), and cutaneous stimulation of the plantar surface of the foot. The effects of eye closure and head ventrifleflexion were also examined.

RESULTS

Figure 1 shows the effects of Jendrassik maneuver reinforcement on the amplitude and reproducibility of MEPs evoked in a quadriplegic subject with paralysis of Tibialis Anterior. MEPs could not be elicited in his relaxed muscles. Attempted dorsiflexion coupled with brain stimulation yielded inconsistent appearance of MEPs. Remote muscle contractions together with attempted dorsiflexion clearly enhanced the amplitude of MEPs and increased the frequency with which they were observed.



In all eighteen spinal cord injured (SCI) subjects, we were unable to elicit MEPs in muscles innervated below their level of lesion with the stimulus intensity at 70% and with the subjects' muscles relaxed. This was in contrast to what we have observed in control subjects. Reinforcement procedures enabled identification of MEPs in the paralysed muscles of 3 SCI subjects thereby confirming preservation of spared corticospinal pathways. Jendrassik maneuver, muscle stretch, eye closure, vibration and cutaneous stimulation all provided facilitation of the MEPs, although the effectiveness of each form of reinforcement varied according to the subjects' clinical status.

CONCLUSIONS

MEPs are not readily evoked in paralysed SCI subjects when employing intensities of stimulation (~70%) that typically evoke responses in lower limbs of control subjects. Reinforcement procedures are effective for increasing the net excitatory synaptic input to the target motoneuron pool. When coupled with attempted voluntary muscle contraction, reinforcement allows MEPs to be recorded in some SCI subjects, thereby confirming preserved continuity in corticospinal pathways.

(Funded by The Spinal Cord Society of Canada)

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GROUND REACTION FORCES AND DOMINANT LEG
KINEMATICS DURING A RAPID STOP AND JUMP

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Recent investigations have discussed the impact characteristics associated with landing from a vertical jump and have related these findings to possible injuries in sports such as basketball. Valiant and Cavanagh (3) reported average peak vertical ground reaction forces (VGRF) in excess of four times body weight (BW) when landing from a vertical jump. Gross and Nelson (2) using accelerometers on the lower leg as well as forceplate input also observed VGRF of approximately four times BW when landing from a vertical jump. In both of these investigations, subjects performed a vertical jump from a standing position. Horizontal motion was not allowed. As Gross and Nelson indicate, the athlete who becomes injured is typically moving and then performs a jump and landing. The athlete will usually run to a site on the court, make an extremely rapid stop followed by an immediate jump. The purpose of this investigation was to expand the previously cited investigations by examining the ground reaction forces during a rapid stop and vertical jump which was initiated by a run.

Methods and Procedures

Eight elite male collegiate basketball players were subjects for this investigation. Each subject gave written informed consent prior to participating. Subjects ranged in age from 18 to 23 years. Their weights ranged from 756 N to 1024 N. After warm-up and acclimation to the forceplate and runway each subject was asked to perform three trials in which they were to run to the forceplate from a distance of five to eight meters, stop and jump. The jump was to simulate a jump shot. The subjects contacted the plate with their dominant leg (determined prior to data collection). Ground reaction data were obtained from a Kistler Type 9803 piezoelectric forceplatform. Analog voltages from the force platform were sampled and converted to digital values at 2,000 Hz per channel with a Data Translation A/D converter and stored in digital format on a microcomputer. Simultaneously, the motion of the subjects were recorded at 200 Hz utilizing three NAC 200/60 high speed video cameras. The images of selected anatomical markers were analyzed with the Motion Analysis System. Analysis of these markers provided a description of ankle, knee and hip angle kinematics during the movement. Digitized image data were stored and all subsequent data analyzed on a SUN 3/260C.

Results

Horizontal velocities during the run-up ranged from 3.3 to 6.0 m/s. Subjects were in contact with the forceplate from 0.25 s to .475 s. Several different prejump landing styles were observed. Five subjects exhibited a bimodal VGRF. The other three exhibited a single impact spike. Maximum VGRF ranged from 1.6 to 4.25 times BW. In one subject the impact peak was followed by a increase in propulsive force. In the other seven VGRF

decreased rapidly with typical VGRF during propulsion of approximately twice BW. Anterior posterior ground reaction forces mimicked the wave form of the VGRF. Peak values ranged from 0.5 to 2.5 times body weight. These forces served to stop the subjects' forward horizontal motion. This force reduced to zero or changed direction (less than 0.1 times BW) just before ground contact ended. Medial-lateral forces were typically below 0.25 times BW for all subjects. Jump heights ranged from 0.62 m to 0.89 m. All subjects demonstrated a forefoot prejump landing pattern. During the prejump phase subjects dorsiflexed through a range of approximately 60 degrees. Subjects landed with the dominant leg nearly fully extended at the knee. During the initial landing phase flexion occurs at the knee, followed by rapid knee extension. Action at the hip was less consistent. Three athletes upon landing maintained their hip angle until they entered the propulsion phase of the movement. The other jumpers demonstrated hip flexion during this phase, followed by hip extension.

Discussion

Ankle, knee and hip angle kinematics were similar to those previously reported. It did not appear that knee flexion was preceded by ankle dorsiflexion as indicated by Gross and Nelson (2). The immediate flexion at the knee and the action at the hip may be in response to the need to immediately begin a vertical jump upon completion of the prejump phase. Vertical ground reaction forces were within reported ranges for the landing. Anterior-posterior forces were high with peaks of 0.5 to 2.5 times body weight observed. This A-P force corresponds to the force magnitude measured beneath the lead foot of baseball pitchers (Blow (1)). The A-P force is necessary to decelerate the subject prior to the jump.

Implications

The rapid stop required by an athlete when performing a jump from a running start as observed in basketball generates large anterior-posterior ground reaction forces in addition to the typical vertical ground reaction forces. Since the knee is nearly extended at touchdown the potential for knee injury exists. Joint forces resulting from this activity require further examination.

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Analysis of Loads on the Canine Lumbar Spine: An Experimental and Theoretical Study.

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INTRODUCTION

The objective of this paper is to present a biomechanical analysis of the loads acting on the canine lumbar spine in different quasistatic postural activities. The model is adapted from that developed and validated by Schultz et al. (1983) to estimate loads on the human lumbar spine in lifting activities. The analysis is validated by comparing predicted muscle forces with measured trunk myoelectric activity in awake behaving animals.

METHODS

The internal forces and moments acting on and about L4 in order to equilibrate the external forces acting on the body above an imaginary transverse L4-level cut were calculated knowing the mass distribution of the animal, the locations of the body segment mass centers and the location, magnitude, and direction of reaction forces acting under the animals forelegs. Because the canine spine motion segment supplies negligible resistance to bending moments ($<0.2 \text{ Nm}^\circ$) the moments must be equilibrated by active lumbar muscle forces. For a given posture, the L4 unknown lumbar muscle and three unknown joint reaction forces were calculated using a linear optimization scheme (Bean et al. 1987) that incorporates the dual assumptions of minimum muscle contraction stress and spine compression. Experimental Analyses. Mass distribution and geometry: A 32 Kg healthy, intact male German shepherd cadaver was frozen at 20°C , X-rayed and photographed in sagittal and lateral views, then sectioned every 5 cm using bandsaw. Sections were weighed and photographed after their C.G. had been found using a knife edge. Using the photographs, geometric measurements and lateral X-ray films, the mass distribution was then reconstructed. Force plate and myoelectric study: A 36 Kg Doberman served as experimental subject. Ten bipolar surface myoelectric (MES) electrodes were taped to the shaved skin over the L1, L3-4 and L6 paravertebral muscles, the L4-level external abdominal obliques at their most lateral point and over the belly of the L4 rectus abdominis muscles. Two strain-gaged force plates (AMTI) were used to record the reaction forces acting under the fore and rear limbs. MES signals were digitized at 900 Hz, rectified and RMS levels calculated over at least two second intervals for each task studied. Tasks studied: We studied how trunk muscle activity and spine loading varied in ten different postural activities. These included: a) quadrupedal standing, b) tripedal standing, c) standing with the head raised or lowered, d) relaxed lying, e) sitting with trunk at 45° inclination, f) bipedal standing with trunk as in e). Similarly, MES data were also recorded from seven additional dogs performing tasks a, d, e and f without the force plates. Model

validation: Predicted L4-level erector spinae, external oblique and rectus abdominis muscle forces were linearly correlated with measured MES amplitude (in uV). Linear correlation coefficients were then reported.

RESULTS

The magnitude of spine compression ranged from 0.07 to 3.68 times body weight (BW). Progressively larger compression loads were predicted from lying, through standing to sitting and bipedal stance activities. The P/A shear force ranged from 2 to 31% BW, and was greatest in sitting. When standing, spine loading amounted to 7% BW in compression and 2% BW in P/A shear. A twenty cm increase in front-rear paw spacing resulted in a 5-fold increase in spine loading. Increasing the trunk inclination angle to 45° increased spine compression force from 24 N in quadrupedal standing to 918 N. Correlation coefficients were greatest for the erector spinae (0.99) and rectus abdominis muscles (0.81). Their values are such that the model appears to capture the essential behavior of the quadruped trunk muscles and consequent spine loading in these tasks.

DISCUSSION

Despite its orientation in the horizontal plane, the lumbar spine of the dog can be loaded under surprisingly large compression forces, even in quasistatic postural tasks. These loads arise because trunk muscle activity, quantified in the MES study, is necessary to equilibrate the forebody relative to the hindbody. The results suggest that the modelling techniques first developed and validated in human studies can successfully be adapted for use in the quadruped. We believe these are the first estimates of quadruped spine loading in quasistatic activities. Knowledge of spine loads is important for the interpretation of experimental results involving the quadruped spine.

CONCLUSIONS

1. Despite its horizontal orientation, significant compression loading of the canine lumbar spine can occur due to trunk muscle activity engendered in postural activities.
2. Spine compression and shear loads are normally less than 10% BW in quadrupedal standing. Activities that involved a 45° trunk inclination resulted in the largest trunk myoelectric activity and spine loads.

ACKNOWLEDGEMENTS

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A Study of Cervical Muscle Response to Experimentally-Induced Muscle Pain

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INTRODUCTION

Low back and neck pain, mostly diagnosed as idiopathic, affects 80% of the population. The role of muscle, a richly innervated and highly sentient structure, is unclear. Muscles must be used to execute activities of daily living, even when pain is present, raising the possibility for a pain-spasm-pain cycle. Because little is known about how pain affects muscle synergies, or vice versa, we have developed an experimental model to study the effect of pain on muscle recruitment, reporting our initial results in this paper.

METHODS

Five healthy adult males (aged <40 years) volunteered as subjects. Eight bipolar surface myoelectric signal (MES) electrodes were placed bilaterally at the C4 level over the infrahyoid, sternocleidomastoid (SCM), splenius capitis and erector spinae muscles. Subjects lay supine on a table with head supported by a load cell. To test repeatability subjects were asked to practice performing 10% isometric flexion and 4% isometric extension efforts against the load cell. As a non-painful control 5 ml isotonic saline solution was infused into the right-side SCM with the subject relaxed. Every 30 seconds subjects were asked to rate any pain they experienced using the visual-analog scale (VAS) as a percentage of the worst pain they have ever experienced. Every 60 seconds, for eight minutes, they were asked to exert a 10% flexion effort and hold it for 3 seconds. MES, VAS and load cell data were digitized at 900 Hz over eight seconds (three sec. being relaxed data and three being the 10% effort). MES data were then rectified and RMS levels calculated over each 3 second interval. The left-side SCM was then injected with 5 ml 5% hypertonic saline to cause pain rated by subjects as 39% VAS. The response of the cervical muscles in the relaxed and active state were then analyzed. The experiment (control, followed by painful injection) was then repeated an identical fashion in the C4-level right and left erector spinae.

RESULTS

The experimental procedures resulted in good repeatability: coefficients of variation (CV) were 31 and 13% respectively for neck flexor and extensor muscle MES on mean relaxed signals of 3.3 and 2.4 uV respectively. The CV for the 10% effort was 49% on a mean SCM signal level of 8.8 uV, where maximum (sd) MES was measured 208(84) uV. Figure 1 shows the time-course of the mean (se) VAS pain level reported by the subjects for the SCM and multifidus painful and sham-control injections. Figure 2 shows the time course of the mean MES (bars denote se) for the right and left SCM for each experimental condition (normal saline: injected muscle (denoted by NI), hypertonic saline: uninjected muscle (HU), etc). Pre-injection (i.e pre-pain) values are given at time zero. The results show that relaxed ipsi- and contralateral muscles at first respond to the pain with a contraction,

then show a progressive decrease (33%) in MES activity. Similar results were obtained for the 10% flexion task: pain resulted in a progressive 40% ($P=0.09$) inhibition of the active ipsi - and contralateral SCM muscles.

DISCUSSION

These experiments show that the central nervous system responds to muscle pain with a progressive inhibition of the painful muscle and its synergists. The period of inhibition appears to outlast the period of significant pain, although more work is needed to examine this effect specifically. Our results obtained with *relaxed* axial musculature are consistent with those of Cobb et al. 1975, who studied the response of extremity muscle to induced pain. The results obtained for *active* muscle are novel. The results demonstrate that pain causes muscle recruitment strategies to alter. At present we are investigating these changes in more detail.

CONCLUSIONS

1. Hypertonic saline caused significant pain of eight (or more) minutes duration in neck muscle.
2. Pain causes an initial increase in relaxed muscle myoelectric activity followed by an inhibition of that muscle.
3. In active muscle, pain caused a progressive inhibition of the affected muscle as well as other synergists.

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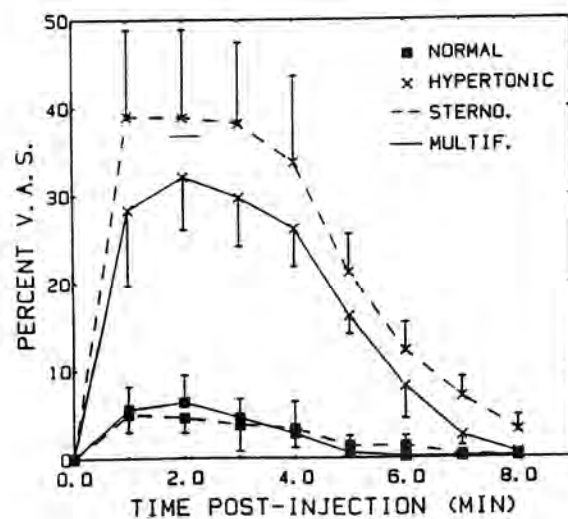


Fig. 1

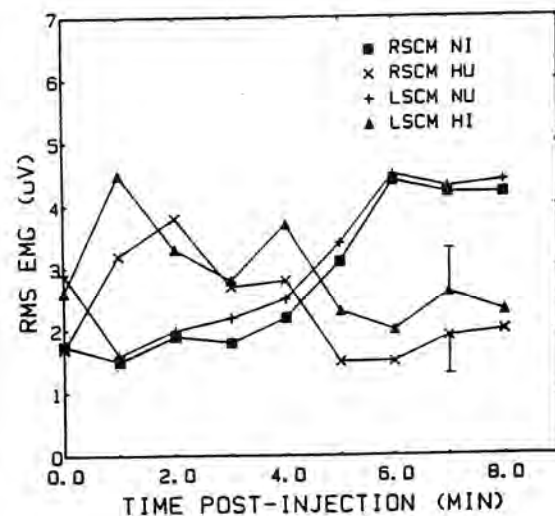


Fig. 2

THE MEASUREMENT AND ROLE OF PROSTHETIC FOOT DAMPING

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ABSTRACT

Designing prosthetic legs to provide improved mobility for amputees requires design specifications which include the dynamic properties of the prosthesis. The purpose of this work is to quantify the dynamic behavior of prosthetic feet and determine the effect of foot properties on gait biomechanics. A computer controlled fatigue testing machine was modified to determine the planar foot compliance and damping. A traditional SACH foot and a prototype carbon fiber composite foot were tested. The results show that the SACH foot is softer in the heel section but is stiffer in the mid-foot section than the carbon fiber foot. The SACH foot is shown to have viscous damping which increases with foot deflection and results in a damping ratio as high as 0.66 for the amputee/prosthesis system. This, combined with the natural frequency of the amputee/prosthesis system of 5.4 Hz, corresponds to a settling time of 0.18 seconds. This time is of the same order as the stance phase characteristic time and thus the damping appears to be important to the gait biomechanics. The testing procedure appears useful for determining the dynamic properties of prosthetic feet.

INTRODUCTION and BACKGROUND

Stein et al., (1989) demonstrated the feasibility of using explicit dynamic models of the amputee/prosthesis system to gain an increased understanding of gait biomechanics and to provide a tool for quantitatively assessing the effects of prosthesis design parameters on gait mechanics. Factors such as foot geometry, compliance and damping are discussed as important to the model's prediction of gait kinematics. This paper presents a methodology for determining the dynamic macromechanical properties of prosthetic feet including foot damping. Results obtained with a SACH foot and a carbon fiber composite foot are compared and discussed.

Previous investigations into prosthetic foot performance have included tests to determine stiffness and hysteresis (Daher et al., 1974; Skinner et al., 1985; Stein, 1983), energy storage (Contoyannis, 1987) and kinematic behavior (Chan (1973)). None of these investigators measured the damping in the prosthetic feet which may also be important to prosthetic gait performance. Some evidence to support this claim is as follows. Energy efficiency of prosthetic feet is a design objective of many prosthetic foot manufacturers. Contoyannis used hysteresis measurements obtained from quasi-static (5 cm/min) load/deflection tests on a variety of prosthetic feet to determine their energy storing potential and energy return efficiency. The mechanism responsible for the hysteresis Contoyannis measured was not reported, but the published raw load/deflection curves clearly show that the mechanism cannot be due to plastic deformation. It may be due to the viscous damping in the foot. Therefore, to assess the energy efficiency of prosthetic feet, foot damping should be determined. In addition, the role of damping may be an important design parameter because damping controls the rate at which energy can be stored and released from the foot. The rate of energy release has been discussed by Contoyannis (1987) and Michael (1987). Contoyannis assumes that the foot is approximately critically damped such that energy is returned within 0.1 seconds. No testing has been provided to verify this assumption. Michael reports the clinical findings that insufficiently damped feet tend to "hurry" the amputee along. He observes that feet with greater damping tend to solve this problem. All of these results reinforce the idea that damping is a critical parameter to the behavior of prosthetic feet.

METHODS

A fixture was designed and installed on an Instron fatigue testing machine (Mooradian, 1988). The fixture allows pure axial loads in the sagittal plane to be imposed on the foot and the corresponding deflection measured. (The fixture will not support a shear load.) The foot can be oriented to any physiologically reasonable shank angle and then be subject to static and/or dynamic loads. The static load/deflection test provides stiffness data, a dynamic (sinusoidal) load/deflection test provides damping information. The damping force is determined by subtracting out the inertial forces measured with an accelerometer, and the compliance force determined from the deflection and the previously determined stiffness coefficient, from the load cell output. The remaining force is the damping force which is 90 degrees out of phase with the compliance force and in phase with the velocity (for a linear mass spring system) (Owens, 1973). The damping is determined by dividing the damping force by the velocity. Damping tests were made using a 5 Hz sinusoid with an amplitude of 0.1 in. applied at various shank angles and foot preloads. Calibration and system performance specifications can be found in Mooradian, (1988). Note that a 30 degree phase shift compensation between the damping force and velocity information was required. Two feet were tested. A SACH foot manufactured by Otto Bock (Pedilan 28, soft heel wedge, Model 1S41) and a carbon fiber foot called Springlite (manufacturers prototype, designed for 138 lbf amputee).

RESULTS

Figure 1 shows a comparison between the load/deflection characteristics of the SACH and the carbon fiber foot as a function of shank angle. These curves show that the SACH foot is softer in the heel and stiffer in the mid-foot region than the carbon fiber foot. The carbon fiber foot has more linear load/deflection curves.

The damping of the SACH foot is given in Table 1 as a function of shank angle and preload. The damping can be seen to increase with shank angle (from heel contact to toe roll) and increase with an increase in deflection (preload).

DISCUSSION

The linearity of the load/deflection plots shown in Figure 1 is somewhat surprising. Complex composite structures with materials such as polymers are often nonlinear. Linearity in stiffness, however, while convenient from a modeling perspective may not be essential or even desirable for optimum prosthetic foot performance. Indeed from an energy storage perspective a softening spring characteristic would provide more energy storage capacity when capacity is defined as by Contoyannis, (1987).

The damping values measured for the SACH foot are based on a linear model of the damping phenomenon. The small amplitude displacement input, which corresponds to a periodic load amplitude of 20 lbf, should not violate this assumption even though, as shown in Table 1, the linear assumption does not hold for large deflections. However, the need to compensate the phase between the damping force and velocity by 30 degrees could mean that the system is not well represented by a linear spring/damper model.

The damping magnitudes by themselves are difficult to interpret. However, by viewing the amputee as a mass, and the foot as a parallel spring damper system, a natural frequency and damping ratio can be calculated. As can be seen in Table 1, the damping ratios obtained indicate that the system is nearly critically damped in some cases. For the case where the damping ratio is .66 and natural frequency is 5.4 Hz, the corresponding settling time is 0.18 seconds. This is on the same order as the characteristic stance phase time and is consistent with the assumption made by Contoyannis (1987) with regard to the time required for the foot to release its energy. Thus, damping appears to play an important role in the behavior of the prosthetic feet and their role with respect to the resulting gait biomechanics.

ACKNOWLEDGEMENT

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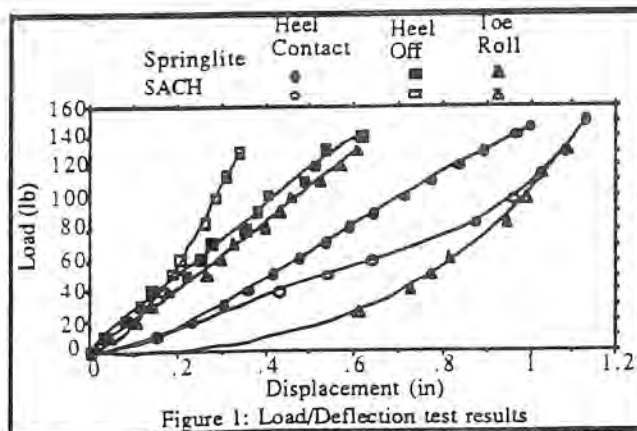


Figure 1: Load/Deflection test results

Table 1: Damping Coefficients - b [Nsec/m], Natural Frequencies - ω_n [Hz] and Damping Ratios - ρ as a function of Shank Angle - θ [deg] and Preload - F_0 [lb].

Heel Contact $\theta = +25^\circ$	b	368	788	1506
	ω_n	2.5	3.67	4.6
	ρ	0.169	0.251	0.383
	F_0	50	100	126
Heel Off $\theta = -10^\circ$	b	1051	2160	2568
	ω_n	5.6	6.3	6.8
	ρ	0.217	0.399	0.444
	F_0	63	96	191
Toe Roll $\theta = -25^\circ$	b	1576	1926	3094
	ω_n	3.5	5.1	5.4
	ρ	0.524	0.443	0.666
	F_0	63	96	191

Note: + denotes degrees before vertical

EFFECT OF ORTHOTICS IN ALTERING REARFOOT MOTION

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The motion of the subtalar joint during the contact phase of the running gait has been the concern of both researchers and clinicians. Normal motion at the subtalar joint includes pronation to a position of maximum within the first 30% of foot contact followed by supination until push-off. This range of motion allows the foot to be flat on the ground and act as a mobile adapter. Abnormal pronation is defined as compensatory in nature and a result of accommodative changes necessary for structural or functional distortions. Excessive pronation at the subtalar joint has been linked with a number of running related injuries including chondromalacia and shin splints. The identification of shoe characteristics which affect rearfoot motion including heel flare, heel height, and midsole composition has led to the production of athletic footwear for the excessive pronator.

PURPOSE: It was the purpose of this investigation to examine the effect of an orthotic in controlling rearfoot motion.

METHOD: Eight male recreational runners who had been wearing semisoft orthotics prescribed by the same podiatrist to correct rearfoot abnormalities associated with excessive pronation were selected as subjects. Each subject was familiarized with the treadmill and ran a 15 minute warm up prior to data collection. Four light reflective reference markers were used to represent the position of the calcaneus and the axis of the lower leg according to the method described by Clarke et al. Subjects were videotaped from the rear running at 10kph under 4 shoe/orthotic conditions presented in random order: neutral shoe with no orthotic (NN), neutral shoe with orthotic (NO), control shoe with no orthotic (CN), and control shoe with orthotic (CO).

Two shoes manufactured by the same company were used in combination with each subject's prescribed orthotic. The control shoe (midsole EVA 42 durometer) was a shoe designed specifically for the excessive pronator to provide maximal rearfoot control. The neutral shoe (midsole EVA 35 durometer) provided moderate rearfoot control and was designed for the normal foot.

The position of the heel and Achilles tendon line were determined throughout the stance phase of three consecutive footfalls for each shoe/orthotic condition for each subject using a VP110 video processor, Motion Analysis System, and Expertvision software. Angular and temporal parameters associated with rearfoot motion were determined and an analysis of variance with repeated measures was used to determine the significance of the difference between the condition means for each variable.

RESULTS: Table 1 displays the means and standard deviations of selected rearfoot angular and temporal variables for each shoe/orthotic condition for the right leg.

Table 1. Means and Standard Deviations of Rearfoot Parameters

Parameter	Shoe/Orthotic Condition			
	NN	NO	CN	CO
Max Rearfoot Angle* (deg)	11.10 (4.54)	10.10 (5.13)	7.15 (2.64)	6.77 (2.27)
Time to Max Rearfoot Angle (sec)	.15 (.04)	.14 (.03)	.15 (.02)	.13 (.05)
Total Pronation* (deg)	15.47 (2.96)	12.98 (4.01)	13.33 (2.64)	9.66 (4.42)
Max Heel Angle** (deg)	92.17 (4.51)	91.59 (4.65)	88.21 (1.59)	87.94 (1.57)
Time to Max Heel Angle (deg)	.14 (.03)	.13 (.03)	.15 (.02)	.14 (.04)
Total Heel Motion* (deg)	11.99 (1.86)	10.22 (2.50)	10.28 (2.17)	7.37 (3.02)

* $p < 0.05$

** $p < 0.055$

The addition of an orthotic to the neutral shoe resulted in a 1% reduction in maximal rearfoot angle, a 16% reduction in total pronation, and a 16% reduction in total heel motion. The addition of an orthotic to the control shoe resulted in a 5% reduction in rearfoot motion, a 29% reduction in total pronation, and a 28% reduction in total heel motion. The use of an orthotic with a control shoe resulted in significant ($p < 0.05$) reductions in maximal rearfoot angle and total pronation and heel motion.

The results of this study support the benefit of the use of a prescribed orthotic device in altering rearfoot motion parameters. The most beneficial combination of shoe and orthotic in controlling rearfoot motion seems to be the use of an orthotic in combination with a rearfoot control shoe.

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THE LUMBARATOR- A REAL TIME SIMULATOR OF
LUMBAR MUSCLES FORCE DISTRIBUTION

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INTRODUCTION

This paper describes a system that can display in real-time the distribution of muscle forces in the lower-back in response to a generalized gravitational loading. Using a precalculated data-base of Muscle Activity Surfaces for twenty two muscles crossing the L3 level in the lumbar region, and a user interface that determines the loading condition, the system expresses the load distribution among all the different muscles using a color (or a gray-level) mapping. The user is thus able to study immediately the effect of given loading conditions (or quasi-static exercises) on the muscle force distribution in all the muscles of the cross-section of interest.

METHODOLOGY

The large number of muscles crossing any transverse lumbar cross-section (up to twenty two for the L3 level), and the limited number of equilibrium constraint equations that can be written (only six for the most general case of external force and moment loading), create a mechanically undetermined system. Ladin and Murthy [1] described an approach that summarized the activity of a single muscle caused by an arbitrary gravitational loading in an erect posture and with no external torsion as a Muscle Activity Surface (MAS). This is a surface that describes the individual muscle force due to a given combination of two independent moments: flexion and lateral bending. The two moments describe the two axes of the 'loading plane' over which the MAS of a given muscle is span.

The 'loading plane' was discretized by creating a mesh of points at intervals of 10 Nm. As each point represents a different loading condition, the following optimization problem was solved: for the given moment loading, find the muscle force distribution such that the spinal compression force due to the muscles is minimized, without exceeding a maximum predetermined stress level in all the muscles. The formulation for a given loading condition is similar to the original one by Schultz et al. [2], it assumes that the spine balances the external forces while the muscles balance the external moments. It iteratively increases the maximum allowed stress level until a linear programming algorithm utilized to solve the above optimization problem converged to an optimal solution. The individual MAS for all the different muscles were stored in the form of a third order array and a bilinear interpolation was used to estimate the force levels of the individual muscles inside the discrete mesh of the loading plane.

RESULTS AND DISCUSSION

Figure 1 describes the screen design of a simulation program that was written on a Silicon Graphics 4D workstation, that implements the above ideas. The left side of the screen describes the loading plane - the horizontal axis is the flexion moment and the vertical axis is the lateral bending moment. By placing the mouse cursor inside the plane, the coordinates of the cursor are read, the data base of the MAS for all the twenty two muscles is accessed and the individual muscle forces that correspond to that loading condition are calculated. The forces are displayed on the right side of the screen using a color or a gray visual display of the overall muscle force distribution that corresponds to a given loading condition. The table in the center bottom of the

screen lists the different muscles, and by placing the mouse cursor inside the table, the 'switching-curve' of that particular muscle can be displayed on the 'loading plane'. The 'switching curve' separates loading combinations that will activate the muscle (on one side of the switching curve) from those that will not (on the other side of the curve).

By moving the cursor in the loading plane the program can simulate in REAL-TIME the variations in the individual muscle forces due to quasi-static changes in the loading conditions. A video-tape simulating the effects of quasi-static physical exercises on the lumbar muscle-force distribution will be presented.

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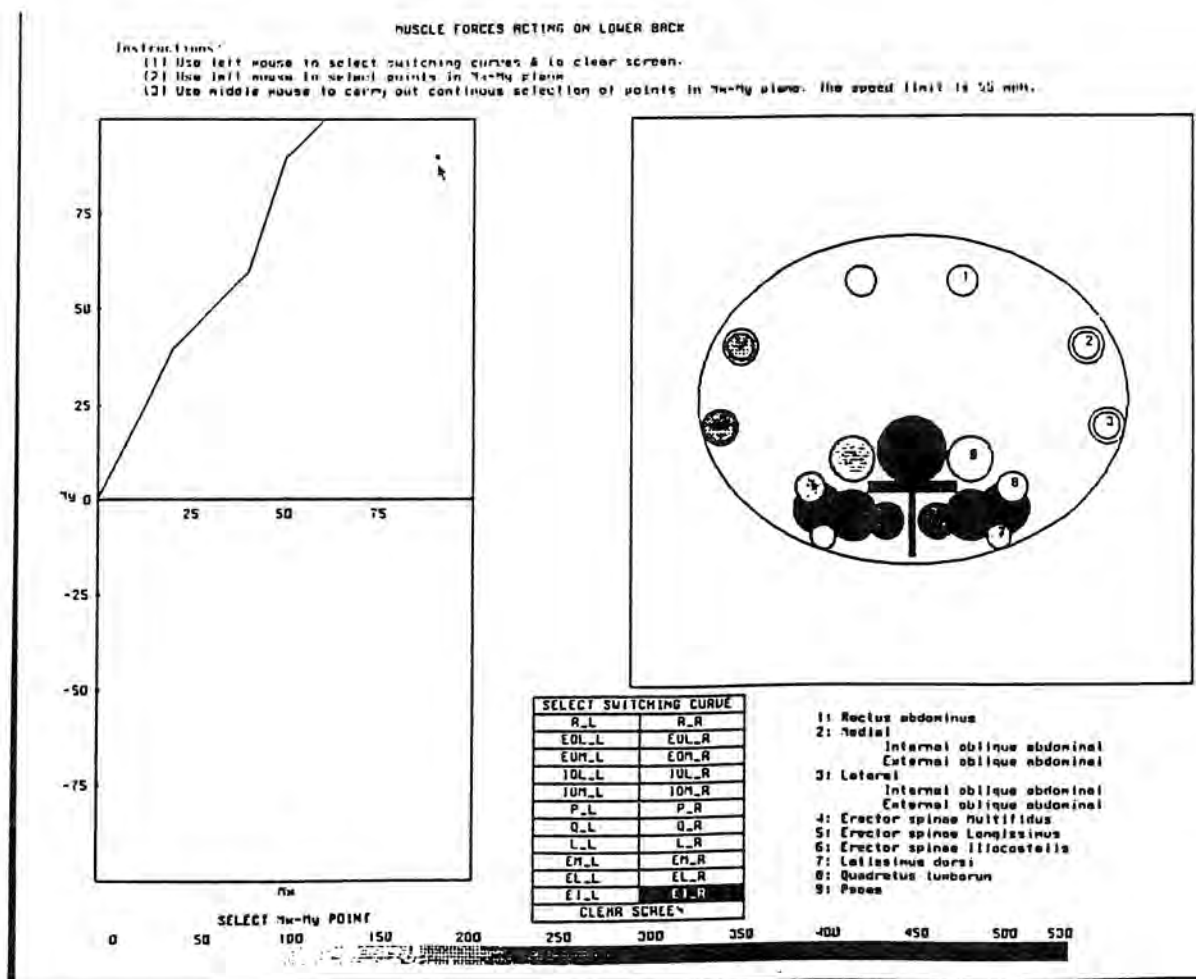


Fig. 1 Screen display of the LUMBARATOR

EVALUATION OF VERTICAL JUMPING TESTS

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INTRODUCTION. Vertical jumping ability in both athletic and nonathletic populations has traditionally been measured by means of Sargent's jump and reach test. In 1980 Bosco introduced a simple way of measuring the rise of height of the body's center of gravity (rHCG) using the jump flight time, which was measured using a thin contact mat and a digital timer. One of the main differences between these methods is the utilization of arm swing during take-off: in the Sargent test the arms, and to some extent, the upper trunk and head swing; giving, if these movements are well coordinated with the leg extension, additional impulse to the take-off. In the Bosco test, on the other hand, the subjects are required to keep their hands on their hips during the entire jump. On the basis of Asmussen et al. (1974), it is possible by comparing jumps performed with and without counter movement (squatting jump, SJ) to evaluate the storage and reutilization of elastic energy of the lower extremity musculature during take-off. Thus if rHCG in the SJ is subtracted from the respective value in the Sargent test, it provides a measure of the effectiveness of the swing and counter movements.

The present study was designed to analyze the results 1) in the Sargent test 2) in the rHCG of the Sargent test 3) in the rHCG of the SJ and 4) in the utilization of swing movements in subject of different sexes and ages, and accustomed to different amounts of daily physical activity.

METHODS. 381 female and 314 male subjects representing age groups of 25, 35, 45 and 55 years were randomly selected from the city of Turku in Finland for the study. On the basis of questionnaire items, 24.1 % of the men and 12.2 % of the women were classified as having physically active life habits while the rest of the subjects were considered as passive in this respect. The subjects were tested for their vertical jumping ability following medical examinations, a warm-up period and preliminary trials. They performed three maximal trials in Sargent's jump and reach test on a contact mat. For the jumps the subjects stood flatfooted with their preferred side adjacent to a calibrated board. The board was contacted with one hand at the highest point of the jump. The score was the difference between the standing reach and the jumping reach. These jumps were followed by three vertical squatting jumps (SJ) performed without a counter movement from a static position of 90° knee angle keeping the hands on the hips. The flight times in the Sargent's and squatting jumps were used to calculate the rHCGs. Two best Sargent's and two SJ's were selected for further analysis. Effects of arm, head and

upper trunk swing and counter movements on jumping height were calculated both in absolute terms by subtracting the rHCG in the SJ from the rHCG in the Sargent's jump and in relative terms (% from the rHCG in the SJ). Reproducibility values for the reaching height as well as for the rHCGs were calculated using the coefficients of variations. Three dimensional analysis of variance was used to calculate the effects of sex, age and everyday physical activity habits on the test values as well as on the reproducibility of the test values.

RESULTS AND DISCUSSION. The coefficients of variation in the Sargent's jump and reach test were between 3.9 and 5.2 for women and between 2.0 and 3.5 for men; in rHCG of the Sargent's test between 3.0 and 5.2 for women and between 2.8 and 3.9 for men; and in rHCG of SJ between 2.7 and 3.7 for women and between 2.2 and 3.0 for men. The analyses of variance showed sex ($p < .01$) and age ($p < .05$) to have statistically significant effects on the reproducibility of the three jumping values; so that men reached more reproducible values than women, and younger subjects more reproducible values than the older ones. Everyday physical activity habits did not have statistically significant effects on the reproducibilities.

The results in the Sargent jump and reach test, the rHCG in the Sargent test and the rHCG in SJ were statistically significantly affected by sex ($p < .001$); age ($p < .001$); and by everyday physical activity habits ($p < .001$); men reached better values than women, younger subjects better than older ones and physically more active subjects better than the less active. Expressed both in absolute and in relative values, the utilizations of swing and counter movements were better for men than for women ($p < .001$); and for the younger subjects than for the older ones ($p < .01$). The youngest male group reached 17.1 % higher jumping values in rHCG of the Sargent's test than of the SJ while for the 45 and 55 year old female cohorts the respective values were 0.8 % and 1.4 %. Everyday physical activity habits did not have a statistically significant effect on the utilization of swing and counter movements.

It was concluded that the Sargent jump and reach test, and rHCG in this as well as rHCG in SJ give highly reproducible results for physically active and inactive urban subjects between the ages of 25 and 55 years. The absolute jumping height is affected by sex, age and physical activity habits, while the utilization of swing movements of arm, head and upper trunk and of counter movements is affected by sex and age.

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DETERMINANTS OF JUMPING AND THROWING ABILITY IN YOUNG ATHLETES

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INTRODUCTION. Throwing and jumping tests are extensively used in sports training in order to evaluate the force-velocity characteristics of athletes. Good results in these tests are combinations of skill and such background variables as, e.g., training, hereditary factors, maturity level, anthropometric characteristics and age (Viitasalo et al. 1988).

The present study was designed to investigate, by multiple regression analysis, the explanatory power of training history, maturational status, anthropometric characteristics and calendar age on vertical jumping height and horizontal throwing velocity among young athletes.

METHODS. Subjects for the study were 174 cross-country skiers, basketball players, ice-hockey players and track and field athletes who belonged to the calendar age groups of 12, 13 and 14 years. The subjects were a sample of child athletes who participated in the Winter Games of the Finnish Sports Federation in 1985.

Skeletal age was estimated from radiographs of the hand and wrist to the nearest 1/4 year by two independent times for each subject according to Greulich and Pyle (1959). For further analysis, the difference between the skeletal and calendar age (Δ age) was calculated to describe maturity level. Body weight and height were measured to within 0.1 kg and 0.1 cm. A plastic flexible tape was used for the determination of shank, thigh and arm circumferences. Vertical jumping height (the height of rise of the body's center of gravity, rHCG) was calculated using the flight time of the jump, which was measured by a contact mat and a digital timer (for details see Bosco 1980). After a warm-up period the subjects performed three to five maximal countermovement jumps (CMJ) keeping their hands behind their necks. The two best results were averaged for further analysis.

Throwing velocity was measured by means of a photocell gate, a contact mat hanging on the wall and an electrical timer. The subjects threw balls of the same diameter (0.32-0.35 m) but of different masses (from 0.3 to 4.0 kg) at maximal speed through the photocell gate to the contact mat. The timer calculated the flight time; which was used together with the distance between the photocell gate and the mat (2 m) for calculation of throwing velocities (for details see Viitasalo 1988). From a parabolic fit between the throwing velocities and the masses of the balls, throwing velocity was interpolated for the mass of 5 % of body weight for each subject.

A structured questionnaire was used to evaluate the frequency, duration and intensity of sports training per week. A stepwise linear multiple regression analysis was used to study the interrelations between the variables. In each analysis only

one of the anthropometric variables in turn was included with the calendar age, training quantity and Δ age to predict results in the throwing and jumping tests.

RESULTS AND DISCUSSION. Calendar age, Δ age, training quantity and body weight together explained 25.2 % of vertical jumping height. Age (15.8 %) and Δ age (8.4 %) were the best predictors in the model. Body weight, training quantity and calendar age together accounted for 36.7 % of horizontal throwing velocity. Body weight had the best explanatory power 29.3 %. The percentage of variability explaining by the different variable factors considered in the present study were rather low when compared to those reported in the literature (e.g. Slaughter et al. 1982). This is most probably explained by the narrow range in the calendar age (36 months) in the present study. In spite of the narrow range in the calendar age, its effects on vertical jumping height were great. This suggests narrow age categories in those sports events where jumping is an important element. Although the results in the throwing tests were calculated for a load which was expressed as percent of body weight, it explained a large proportion of throwing velocity. This means, that muscularity together with the amount of training may be one of the most important factors affecting horizontal throwing velocity. This probably suggests that classifications of young athletes in events where throwing velocity is important, should be based more on body weight than on age.

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EFFECTS OF VARIOUS LOADS AND SPEEDS ON THE KINEMATICS OF CLIMBING

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Changes in movement kinematics have been extensively investigated in walking, running, and cycling based on different loads and/or speeds. There has been a lack of documentation on the kinematic analysis of climbing and climbing under different environmental and work conditions. Therefore, the purpose of the study was to investigate the effects of load and speed upon the kinematics of climbing. A climbing exercise machine (Versa Climber) was used as the experimental apparatus.

Thirty-six physically healthy volunteer subjects (18 males and 18 females), ranging in age from college-age to middle-age (20-40 years), were tested at the University of Illinois Biomechanics Research Laboratory. Each subject was marked at the following anatomical sites with 1.5" white adhesive tape with 1"x1" black rectangles marked over it: left ankle, left knee, left hip, left shoulder, left elbow, left wrist, and 7th cervical vertebrae. The subject wore an elastic belt with styrofoam blocks with black circles to mark the iliac crests and sacrum. Each subject performed under 9 randomly assigned load-speed conditions for 1.5 minutes each, with a rest in between. The 3 load conditions (in total kilograms) for males and females, respectively, were as follows: no load (0, 0), partial (8.7, 6.8), and full (16.9, 12.7). Partial load consisted of weights to the head, waist, and ankles, and full load consisted of weights to the head, waist, ankles, wrists, and back. Weight bands and a backpack were used as loads. The 3 speed ranges (ft./min.) for males and females, respectively, were as follows: low (80-90, 40-50), medium (110-120, 70-80), and high (134-140, 94-100). A split-screen videotape system was used to record the climbing performance in the coronal and sagittal planes. One camera was positioned 3.3 m to the left side of the subject to record the climbing performance on the Versa Climber in the sagittal plane. The second camera was placed 9.0 m behind the subject to record the performance in the coronal plane. A digital timer was used to superimpose real time on each video image. During the videotaped performance, actual climbing speed and step length were recorded from the read-out display on the Versa Climber. Climbing cadence, angle of head and trunk, and range of motion and angular velocity at elbow, hip, and knee joints were measured and recorded from the videotapes by using a transparency superimposed to the screen of the monitor. A protractor was used to measure joint angles at maximum/minimum flexion, and the values were averaged for 10 climbing cycles.

Reliability of data measurement was confirmed by repeat measurement of all data. Based on F-tests (MANOVA's), the following changes were significant at the .05 or .001 level. All kinematic parameters, except the range of motion at the knee, angle of the trunk in the coronal plane, and pelvic rotation in the coronal plane, changed significantly with respect to speed. As speed increased, the following changes took place in the parameters: 13.6% increase in the range of motion at the elbow joint ($p < .05$), 13.7% increase in the range of motion at the hip

joint ($p < .05$), 45.8% increase in the angular velocity at the elbow joint ($p < .001$), 40.4% increase in the angular velocity at the knee joint ($p < .001$), and 13.4% increase in the angle of the head/neck to the left in the coronal plane ($p < .05$). An analysis of cadences (step length, cycle time, and cycle frequency) was conducted to confirm that three distinct actual climbing speeds were obtained, and their averages were 65, 95, and 116 ft./min. for the entire group. There was consistency in the actual speed by the subjects with each of the three loads. Load affected mainly movements of the trunk and head in the sagittal plane. There was a significant increase in the forward movement of the trunk ($p < .05$) and increase in the backward movement of the head ($p < .001$). Another parameter also affected by load was the distance between the sacrum and ankle markers at the midpoint of the climbing cycle. This distance (representative of increased trunk flexion) increased significantly ($p < .01$). Load also contributed to the significant increase in downward pelvic rotation in the coronal plane ($p < .01$).

Speed was the predominant factor causing a change in performance of these subjects, but affected mainly the upper and lower extremities. Movements of the head, trunk, and pelvis in the sagittal and coronal planes were affected by the changes in load. Such changes in the body segments were extraneous and inefficient and added greater stress to the lower back.

Table I Significant Variables ($* < .05$, $** < .01$)

LOAD	SPEED
**Head angle	**Head angle
*Trunk angle	**Trunk angle
**Horizontal distance: ankle to sacrum	**Horizontal distance: ankle to sacrum
**Rotation of pelvis	*ROM: elbow, hip
	**Angular velocity: elbow, hip, knee
	*Lateral head movement
	**Cycle characteristics

THE INFLUENCE OF IMPACT SPEED ON JOINT KINEMATICS AND IMPULSE CHARACTERISTICS OF DROP LANDINGS*

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Landings performed by gymnasts during competition must be performed with a single placement of the feet without excessive joint motion. Gymnasts must satisfy this performance criteria even when landing advanced gymnastics skills at high impact speeds. In contrast, landings performed by recreational athletes are uninhibited by performance standards and thus may reflect a more natural response to increases in impact speed. The purpose of this study was to characterize and compare the adjustment in landing strategies made by gymnasts and recreational athletes when encountering landing impacts over a range of impact speeds. By studying the changes in segment motion and reaction forces under increasing impact velocity conditions, the preferred adjustments in loading characteristics for the gymnasts and the recreational athletes may be identified.

Methods

Six collegiate male gymnasts and six recreational athletes, matched in height, weight, and age, served as subjects. Standardized two foot landings were performed by each subject from three heights (.32m, .72, 1.28m) in a randomized order. The segmental movement patterns demonstrated during each of the three trials performed at each height were recorded using high speed film at a rate of 202.4 frames per second. The external forces applied to the human link system at the foot-ground interface were quantified simultaneously using a force plate at a data collection frequency of 1000 Hz. The data for each coordinate of the digitized points were digitally filtered independently by using a second order Butterworth Filter (Saito & Yokoi, 1982) with a cut-off frequency derived by the method of Jackson (1979). The ground reaction force data for each subject were normalized by body weight of the individual and expressed as a multiple of body weight (BW). The interval of interest, identified as the landing phase, was defined as the time required to bring the vertical velocity of the total body center of mass to zero. Statistical comparisons between the ground reaction force-time histories and the ankle, knee, and hip joint kinematics were made between subject groups across the three velocity conditions by using ANOVA ($p < .05$).

Results and Discussion

The reaction force-time data acquired during the landings reveal that the Gymnasts tended to decrease the landing phase time when approaching landing impacts at greater speed where as the Recreational Athletes tended to increase the landing phase time (Table 1). The tendency to increase the landing phase time by the Recreational Athletes may reflect the natural tendency to distribute the higher forces incurred at higher impact speeds over a longer period of time. This strategy may be responsible for the lower mean impact peak experienced by the Recreational Athletes (9.12 BW) under the high impact condition, as compared to the mean impact peak of experienced by the Gymnasts (10.96 BW). Although, the impact peak magnitudes were significantly different over impact speeds, no significant differences were noted. The Recreational Athletes also demonstrated joint flexion motions which appeared to be more sensitive to impact speed than those of the Gymnasts (Table 2). Although both groups increased joint flexion at the knee and hip with increases in impact speed, the adjustments in minimum joint flexion angles from low to high impact speeds were greater for the Recreational Athletes (Knee: 109°-64°; Hip: 129°-49°) as compared to the Gymnasts (Knee: 96°-73°; Hip: 93°-61°).

Conclusion

The adjustment in landing strategies used by gymnasts and recreational athletes in response to increases in impact speed appear to be significantly different. The landings performed by the gymnasts appeared to be less sensitive to increases in impact velocity, which may enhance the ability of gymnasts to meet the performance criteria required for competitive gymnastics landings.

Table 1: Mean vertical reaction force-time characteristics of landings performed by Gymnasts (GYM) and Recreational Athletes (REC).

GROUP	SPEED	PHASE	IMPACT	TIME OF	
		TIME (s)*+ #	PEAK (BW)*	IMPACT (s)*	PEAK (% phase)*+ #
GYM	Low	0.302	3.93	.049	(16.2)
	Medium	0.285	6.26	.038	(13.8)
	High	0.273	10.96	.030	(11.2)
REC	Low	0.168	4.16	.061	(38.4)
	Medium	0.259	6.38	.041	(16.5)
	High	0.292	9.12	.034	(11.9)

* Statistically significant difference between impact velocity conditions ($p < .05$)

+ Statistically significant difference between groups ($p < .05$)

Statistically significant interaction effect between impact velocity and groups ($p < .05$)

Table 2: Mean minimum joint angle and time of minimum joint angle demonstrated by Gymnasts (GYM) and Recreational Athletes (REC) during the landing phase.

MINIMUM JOINT FLEXION ANGLE							
GROUP	SPEED	DEGREES(°)			TIME OF MINIMUM (s)		
		ANKLE*	KNEE*	HIP*#	ANKLE	KNEE#	HIP*#
GYM	Low	80	96	93	.24	.28	.29
	Medium	81	87	81	.21	.25	.28
	High	81	73	61	.23	.24	.27
REC	Low	83	109	129	.16	.17	.17
	Medium	80	83	87	.19	.23	.26
	High	78	64	49	.24	.25	.30

* Statistically significant difference between impact velocity conditions ($p < .05$)

Statistically significant interaction effect between impact velocity and groups ($p < .05$)

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THE EFFECTS OF INTERNAL TIBIAL DEROTATIONAL OSTEOTOMIES ON
GAIT IN CHILDREN WITH CEREBRAL PALSY

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Cerebral palsy spastic diplegia patients have multiple gait abnormalities which are caused by muscle spasticity and imbalance. During growth, the imbalance in muscle forces leads to contractures and eventually rotational deformities. These imbalances can be addressed surgically through muscle lengthening and transfers. Bony deformity can be corrected by fusion and derotational osteotomy. One common problem seen in spastic diplegia is the excessive external rotation of the foot secondary to an external tibial rotation. External tibial torsion is aggravated by an inability to clear the foot during swing. Perry¹ hypothesized that a transfer of the rectus femoris to the posterior aspect of the knee along with hamstring lengthening would eliminate clearance problems by decreasing crouch in stance and augmenting flexion in swing. Gage², however, noted poor results with this procedure were associated with a foot progression angle of 8 or more degrees greater than normal.

The purpose of this study was to compare pre-operative and post-operative gait analyses in a group of patients who underwent supramalleolar derotational osteotomies for excessive external tibial torsion. Five children (7 limbs) with excessive external tibial torsion as defined by a lateral twist of the tibia on its longitudinal axis were included in the study. This was measured by the angle between the foot and the thigh with the knee at 90° (normal =10°) and by the foot progression angle during gait. All patients had undergone previous lower extremity surgery to improve sagittal plane motion previous to the derotational osteotomy. Pre and one year post-operative gait analyses were performed including 3D kinematic and kinetic analyses. The gait analysis system used had been described in detail by Tashman³.

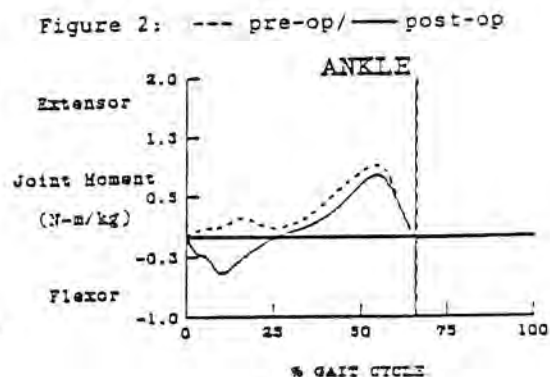
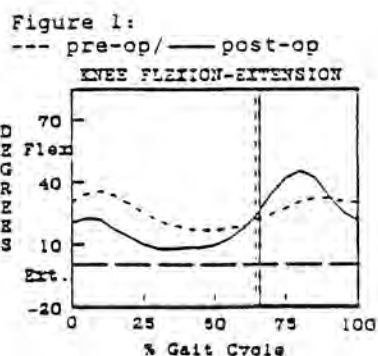
Comparison of the average pre-operative (31.8 degrees) and postoperative (11.5°) external foot rotation demonstrated that the surgery did affect the desired change. Kinematic sagittal

plane differences can be seen in figure 1. Ankle dorsiflexion and plantarflexion showed an increase in all cases but significant differences were found in an increase from 13 to 17 degrees ($P < 0.02$). Sagittal plane motion of the knee showed an increase in extension in stance (9.5 degrees) and an increase in swing (4.0 degrees) and a significant increase in range of motion (13 degrees, $P < 0.08$) post-operatively. Pre-postoperative ankle kinetics of one patient are presented in figure 2. Postoperatively there is better modulation with a typical dorsiflexion moment after initial contact.

It is clear from the results that in order to improve sagittal plane motion in children with cerebral palsy spastic diplegia the excessive external tibial torsion must also be addressed. The Supramalleolar derotational osteotomy brings the externally rotated foot closer to the line of progression resulting in improvement in sagittal plane kinematics and kinetics at the knee and ankle following surgery.

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A Biomechanical Analysis of Changes in Running Technique
during a Treadmill Run to Volitional Exhaustion

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INTRODUCTION

A vast amount of research has attempted to identify the mechanisms and manifestations of fatigue as it occurs in human performance (Gibson & Edwards, 1985), but there have been relatively few attempts to assess changes in motor performance during prolonged activity in basic motor tasks such as running. The purposes of this investigation were to describe the compensations runners make in order to maintain a given running velocity and to compare fast and slower runners with regards to the magnitude and relative timing of these compensations.

METHODS

Nineteen male runners were divided into two groups: a fast (F) group ($n=9$) whose most recent 10 km time was less than 36 min and a slow (S) group ($n=10$) whose most recent 10 km time was between 44 and 50 min. Each subject completed a treadmill run to volitional exhaustion at a speed approximating his current 10 km pace. If the time run prior to volitional exhaustion (TT) was not within 30% of his most recent 10 km time, the subject was given a few days to recover and the run was repeated at an adjusted speed.

Coordinate data representing the head of the fifth metatarsal, heel, ankle, knee, hip, shoulder, elbow, and wrist were collected after the initial five minutes on the treadmill, and every third minute thereafter, until the subject indicated he was within five minutes of volitional exhaustion. At that time, coordinate data were collected every minute until the test was terminated. Coordinate data were sampled at 100 Hz by a WATSMART optoelectric device using standard two-dimensional motion imaging techniques. Variables analyzed statistically included stride length, maximum hip flexion, maximum hip extension, range of motion at the hip, maximum knee flexion, maximum knee extension, range of motion at the knee, maximum plantar flexion, maximum dorsiflexion, range of motion at the ankle, and mechanical work done per stride.

RESULTS

The statistical analyses ($p<.05$) suggested that stride length, maximum hip flexion, range of motion at the hip, and maximum knee flexion were significantly increased at various times during the run to volitional exhaustion (see Figures 1, 2, 3, and 4). Conversely, the maximum knee extension angle demonstrated a significant decrease (see Figure 5). They further indicated that the two groups did not differ with regards to the nature of the compensations identified.

In general, the magnitude of changes was small, but the nature of the changes was consistent. For instance, the maximum increase in stride length was 2.8 cm for the S group and 8.3 cm for the F group. Similarly, the largest increases in maximum hip flexion were 1.4 deg and 1.8 deg for the F and S groups, respectively. The largest changes in the range of motion at the hip (2.4, 2.0 deg), maximum knee flexion (1.6, 1.7 deg) and maximum knee extension (-2.9, -1.2 deg) also were small. The decreased knee extension was reflected as a reduction in the maximum amount of knee extension observed prior to foot contact. While these data represent group trends, it was apparent from the data that some subjects were considerably more sensitive to the effects of fatigue, as they demonstrated pronounced compensations during the run to exhaustion.

DISCUSSION

The nature of the changes identified above indicates that submaximal force production was unimpaired during the run to volitional exhaustion. This seems to be inconsistent with the

conception of fatigue as a reduced ability to produce force. These changes may represent a conscious attempt to increase stride length or they may be associated with a progressive inability to maintain the cycle rate. Research with isolated muscles has indicated that the development of fatigue causes a slowing of contractile velocity and relaxation, producing a reduction in the rate at which cross-bridges can cycle without diminishing force production (Bigland-Ritchie & Woods, 1984). If this can be extended to the intact system, it would support the latter possibility.

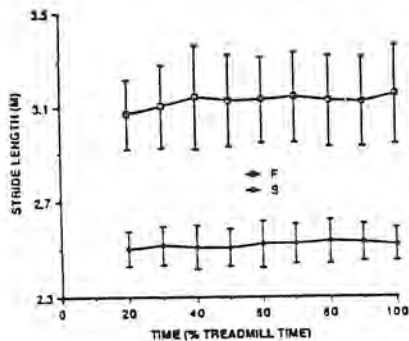


Figure 1. Mean stride lengths interpolated at various points of the run to volitional exhaustion.

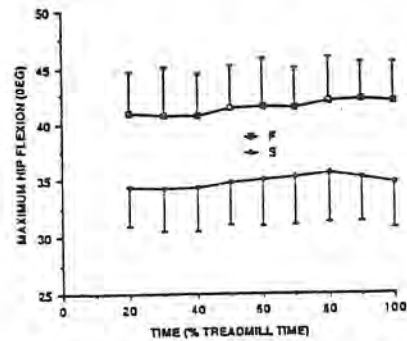


Figure 2. Mean maximum hip flexion angles interpolated at various points of the run to volitional exhaustion.

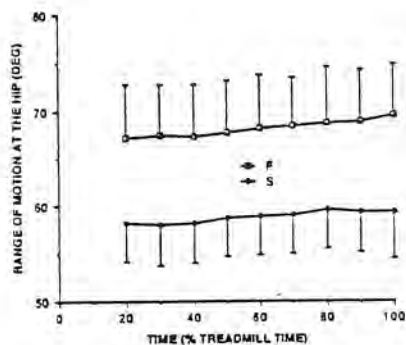


Figure 3. Mean range of motion at the hip interpolated at various points of the run to volitional exhaustion.

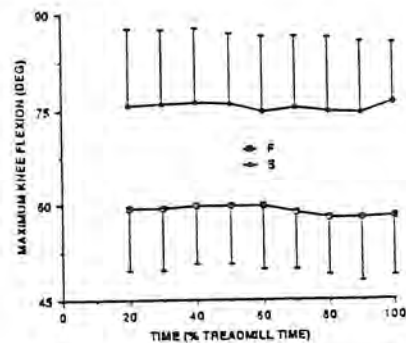


Figure 4. Mean maximum knee flexion angles interpolated at various points of the run to volitional exhaustion.

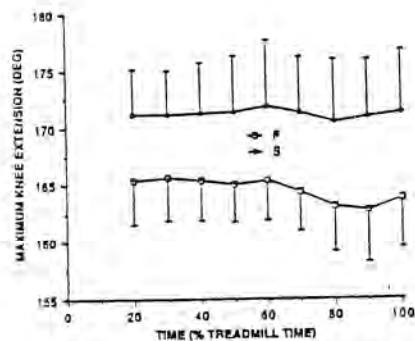


Figure 5. Mean maximum knee extension angles interpolated at various points of the run to volitional exhaustion.

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FINITE-ELEMENT-STUDY OF THE STABILITY OF A FEMUR PLATE OSTEOSYNTHESIS. INFLUENCE OF THE SCREW DIAMETER AND THE SCREW PLATE CONNECTION

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INTRODUCTION

The E-module of steel is about ten times as high as the E-module of cortical bone. Stress concentrations in contact areas result, especially in the screw-threads and in the screws. As opposed to the function of the screw for plate fixation we were interested in the screw as part of the static structure. Of special interest was the screw diameter and a fixator like rigid screw-plate connection. Such internal fixators are of great value in instable spine fractures(2).

METHOD

A quarter-section of a plated human femur was modelled (fig.1) by half of a 75mm tube of 14mm radius and 7mm cortical thickness with attached quarter of a 6-hole-plate fitted at its convex undersurface. Plate and tube were generated separately, so no friction was assumed, all load transfer took place over the screws. 3-dimensional 20-node elements were chosen for tube and plate. Beam elements were used for the screws. The material properties are given in table 1. Calculations were done for 3mm, 5mm and 8mm screw core diameter, and in each case simulating a screw-plate junction with allowed or fixed rotation. Single-legged-stance loading conditions were used. Forces were applied on the center of an assumed stiff star at the end of the tube, given in table 1 related to a coordinate system fixed in the transverse symmetry section at the middle of the plate.

Table 1 : Material and load parameters(4)

E-module cortical bone	18000 N/mm ²
Poissons's ratio	0.11
Maximal pressure	100 N/mm ²
E-module stainless steel	210000 N/mm ²
Poisson's ratio	0.3
Maximal yield stress(σ 0.2)	300 N/mm ²
Loads: bending moment Mx	55200 Nmm
shear force Fy	-90 N
axial force Fz	460 N

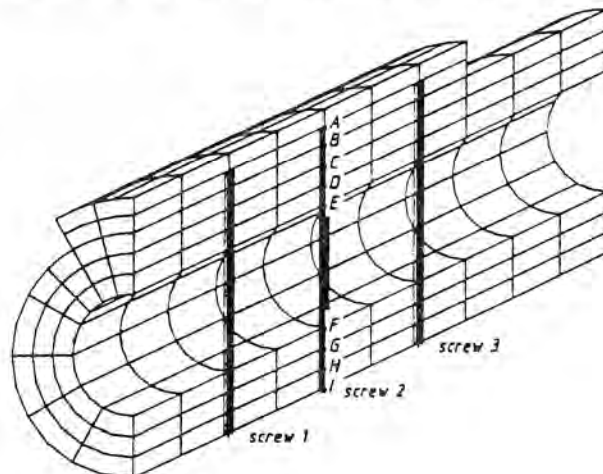


Figure 1: FEM-Model

RESULTS

Pressures on the face of the screwholes were maximal directly under the plate (B, see fig.1), decreasing in the cortical bone (C,D,E), while the values in the opposite (F,G,H,I) thread were one order of magnitude less

with an opposite direction. Maximal values are summarized in table 2. Bending stresses in the screws behaved in a similar course. The maximum values in the cases of a fixed screwhead were found at the screw-plate-junction, however. Maximal values are given in table 3. End-plate bone and screws (screw 3, see figure 1) were loaded in excess of their material limits when inner diameters of 3mm were used. Axial stresses in the mid-plate cross-section, which can be thought of as a healed transverse fracture, were calculated and the results summarized as the location of the neutral fibers on the hemicircular surface (table 4). A definitive shift towards the plate with larger screw-diameters could be demonstrated, while in this model, with an intact bone, relatively small changes were due to a fixed screw-plate-connection.

Table 2: maximal pressure on the face of the hole(N/mm²) Table 3: maximal bending stress in screws(N/mm²)

	screw	3mm	5mm	8mm		screw	3mm	5mm	8mm
head	1	45.2	21.5	8.8	head	1	122	35	9
rotation	2	105.5	52.2	22.6	rotation	2	284	86	29
possible	3	178.3	96.2	47.9	possible	3	479	161	52
head	1	39.7	18.2	8.3	head	1	131	29	3
fixed	2	94.6	45.0	21.6	fixed	2	308	71	8
	3	166.8	86.4	46.4		3	529	129	13

Table 4: location of neutral fiber(degrees, 0:directly under plate)

	3mm	5mm	8mm
head rotation possible	75	62	50
head fixed	72	62	50

DISCUSSION

Plate-tube-friction influences the stress distribution with maximum error when a fixed plate-tube surface is assumed (1), screw- and plate-pretension will give higher stability in early fracture healing but decrease later (3). So, neglecting friction and pretension seemed appropriate in a simple model. We concluded that the end-plate screws are most prone to failure, screws with 3.2mm core (typical cortical screw) may brake or cause bone resorption. Increasing screw diameter in a femur osteosynthesis should be further investigated, as a possibility to increase the bending stability of the system without the necessity of an axially stiffer plate(5). When no osteal defect is present a rigid screw-plate connection has little influence.

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Effects of rate of force development on ligament-muscle synergies in normal knees.

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The presence of mechanoreceptors in the human anterior cruciate ligament (ACL) has been established (1). Indirect clinical evidence that mechanoreceptors may serve to provide the central nervous system with afferent information important in the control of joint forces and motion appear to support the neurosensory hypothesis of ligaments (2). Investigation of the neurosensory role of the ACL have primarily focussed upon the ACL-hamstrings response to anterior drawer forces. Previous research has demonstrated that the normally functioning knee joint does not elicit the proposed anterior drawer ACL-hamstrings synergy during constant force isometric contractions (3,4) but subjects having a conservatively or surgically treated ACL injury have significantly different synergies than those established as normal (5). The purpose of this experiment was to determine if manipulation of the rate of isometric force development could elicit measurable ligament-muscle interactions in the ligamentously normal knee.

METHODS

Subjects performed isometric knee extension contractions at two knee joint angles (15 and 85 degrees) and at three rates of force development (force rate). The ACL is strained by anterior drawer at 15 degrees flexion but at 85 degrees flexion the drawer component is posteriorly directed. The maximum force rate and the magnitude of the maximum force was determined for each subject as the mean of four trials. Five trials at two submaximal force rates (1.0 and 0.5 Hz) were collected. Criteria were established to determine the acceptability of each of the submaximal trials: peak force within 15 percent of the previously established maximum value and elapsed time to the peak force within 20 percent of the required target time. Electromyographic signals were collected from four components of the quadriceps femoris, the long head of the biceps femoris (BF) and a muscle mass considered representative for both the semimembranosus and semitendinosus (Semi). Analysis of the data consisted of integrating the EMG signals in three consecutive 50 ms windows, the first of which began at the onset of isometric force. Consecutive windows were used so that reflexive excitation of the hamstrings following the onset of force (and ACL strain at 15 degree flexion) could be observed. The EMG signals of the hamstrings were normalized using the mean of ten 50 ms integrals during a maximum knee flexion contraction.

RESULTS AND DISCUSSION

No evidence of force rate-dependent increased excitation indicating reflexive activity was observed for either muscle

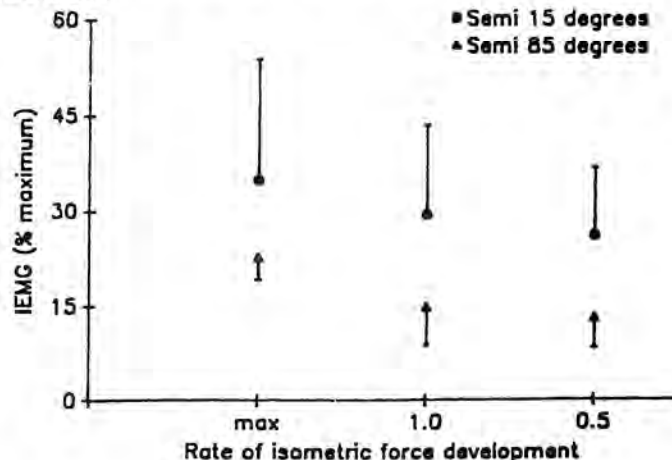
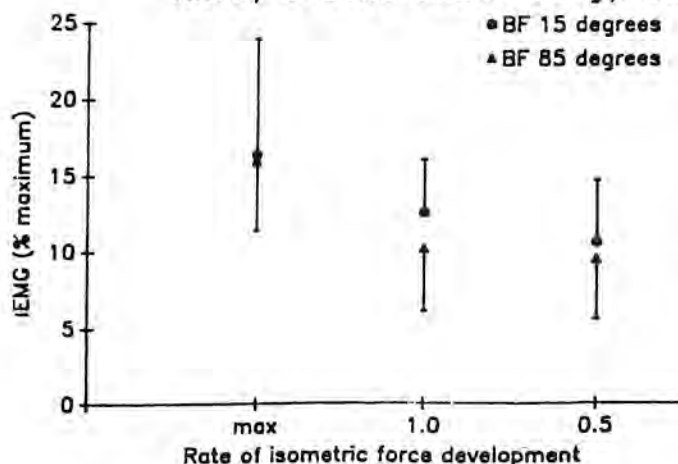
at either knee angle during analysis of the consecutive 50 ms integrals. The largest difference between integrals for either muscle at any force rate was 2.9%. The summed integrals were subsequently treated as a single 150 ms variable (see Figure). Force rate and knee joint angle had no meaningful effect on BF excitation. The mean increase in BF excitation due to force rate was 6.7%. The BF excitation at the 15 degree position was a mean of only 1.3% higher than that of the 85 degree position. Force rate was associated with a mean 9.2% increase in Semi. Knee joint angle had a larger effect on Semi excitation than did the force rate. The excitation of Semi at 15 degrees flexion was a mean of 13.2% higher than that at the 85 degree flexion.

CONCLUSIONS

The results of this study lead to the conclusion that the rate of isometric knee extension force development does not elicit a ligamentously mediated reflexive contraction of the hamstrings in the ligamentously normal knee. However, the results of this experiment suggest that knee joint mechanoreceptors are important in the planning and execution of motion. The data suggest that BF and Semi excitation is selectively mediated as a function of knee joint position to restrict anterior drawer, internal tibial rotation, or a combination of both. Furthermore, the excitation mediation occurs prior to the onset of force generation.

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DESIGN VALIDATION AND USE OF TWO INDOOR-OUTDOOR SPROCKET-MOUNTED PORTABLE CYCLING COMPUTERS FOR EVALUATION AND MODIFICATION OF PEDALING TECHNIQUE.

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The purpose of this investigation was to design and validate portable cycling computers capable of providing instant feedback for the improvement of individual pedaling technique during training and competition.

The negative torque indicator (NTI) was designed to provide instant feedback to the cyclist in the form of an audible sound in headphones installed within a bicycling helmet, such that on occasions when the cyclist produced a counterpropulsive or negative tangential or shearing crank arm force sound intensity registered in the helmet would be directly proportional to the magnitude of the applied negative torque(s).

The assumption is made that a cyclist who can reduce or eliminate negative torque will increase both his propulsive power output and improve his mechanical efficiency.

A strain gauge bridge on each crank arm senses the negative torque produced by each leg. Output from the strain gauges is converted to alternating current and amplified. A precision rectifier removes the signal caused by positive torque. The rectified output from the left and right crank arms is summed and transmitted to a receiver mounted on the bicycle frame without the use of a brush assembly. The signal is amplified by the receiver and transmitted to a headphone or a speaker.

Prior to testing each crank arm is zeroed by rotating each potentiometer to a position where sound cannot be detected in the headphones. The static loading calibration procedure is complete when the technique described is observed for a negative force magnitude range of 0-135 N using 24N increments.

The pedal stroke analyzer system (PSA) was designed to be a dual-purpose portable training device for the possible enhancement of individual pedaling and cycling performance. Specifically, it has been used to provide the cyclist with real-time visual feedback both indoors and outdoors of total and individual tangential force mechanical power outputs (W), average leg speeds (r.p.m.), gear ratios, negative torque values (Nm.), impulse patterns (N-sec.) indoors, heart rate (BPM), breath frequency (bpm), that can be simultaneously displayed on a liquid crystal display as an extension of the handlebars, and enabled or disabled by the cyclist at will.

More detailed post-event analyses of cycling patterns matched with mechanical power output variations during a race (leg speed x gear size combinations), have been incorporated using the computer dump mode.

For indoor testing and training sessions appropriate software has been developed to enable documentation and visual feedback of 16-point impulse

patterns generated for each leg in an attempt to reduce what appears to be mechanically power dependent magnitude(s) of negative torque(s) pattern(s) produced by elite riders.

The system connects to any stationary bicycle, bicycle ergometer, or trainer, and any IBM compatible computer with a serial port. The PSA has a strain gauge bridge mounted on each crank arm to sense left and right torque outputs. Crank arm position is determined using a sensor. The signal from the strain gauges is fed into four high gain chopper stabilized differential amplifiers. The output from each amplifier, the position sensor, and battery voltage sensor is modulated and transmitted to a receiver on the bicycle frame. A receiver mounted on the bicycle frame demodulates the signal, and a microprocessor converts the demodulated signal into a 4800 baud RS232 compatible signal. The signal is then relayed to the host computer and selected parameters displayed real-time.

Results:

At leg speeds of 90-105 rpm, power outputs of 300-385W, and gear ratios between 53:17 and 52:14 (84-102 inch) for 45 minute rides indoors, 11 of 15 subjects were able to reduce their magnitude(s) of negative torque by 10-18 percent using the NTI, without altering bicycle posture.

Significantly greater negative force magnitudes existed for the group at the higher average leg cadence range of 100-110 rpm.

Ten of 15 subjects produced similar total work magnitudes during a 45 minute outdoor time-trial, but total negative work values varied considerably between individuals tested, and appeared to occur when the total average net power asymmetry per individual was significant ($\geq 15\%$ bilateral difference(s)).

Force patterns at power outputs of 300-420W were replicable for all subjects tested; they achieved this by increasing the propulsive force magnitude(s) between 45 and 157.5 degrees past top-dead-center, rather than altering the distribution of the impulse pattern(s) at the higher power outputs.

A direct relationship existed for all subjects between increase in leg speed (rpm) and increase in the magnitude of negative torque (Nm) generated for gear sizes eliciting power outputs in the 320-449W range.

The most significant increase in average power output for the group (40W), occurred with increases in leg speed from 104-109 rpm in the 53:17 to 56:14 gear ratio range.

Conclusions:

Within the limitations of this study, both computers developed by the group provided accurate, repeatable measurement of the magnitudes of negative torque (Nm) and average net power (W) generated by elite riders during indoor and outdoor training and competition.

It appears that use of the NTI in conjunction with weight training enhances hamstrings involvement and contribution to pedaling, which may improve pedaling efficiency.

Assuming that smoothing impulse patterns is the key to a reduction in performance times for the individual, more research is needed to determine the combination of mechanical and biomechanical parameters that will achieve the desired goal.

WHAT IS THE ROLE OF BIARTICULAR MUSCLES DURING JUMPING?

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INTRODUCTION Previous investigators [1, 2] have conjectured that biarticular muscles contribute significantly to jumping performance by transporting power from proximal to distal joints in an attempt to increase ankle power output prior to lift-off. In fact, the importance of biarticular muscles to jumping has been punctuated by the hypothesis that jump height would decrease by as much as 50 percent if only uniarticular muscles were available for propulsion [1]. With the overall aim of understanding how intermuscular control and musculotendon dynamics coordinate a complex human motion, an optimal control model was constructed to study maximum-height jumping [3]. A specific objective of this work is to assess the functional role of biarticular muscles.

METHODS The human body was modeled as a four-segment, planar, articulated linkage, joined together by frictionless revolute joints. The lower extremity musculature was modeled using eight musculotendon actuators: soleus (SOL), gastrocnemius (GAS), other plantarflexors (OPF), tibialis anterior (TA), vasti (VAS), rectus femoris (RF), hamstrings (HAMS), and gluteus maximus (GMAX). Muscle was represented as a three-element entity: a Hill-type contractile element, a series-elastic element, and a parallel-elastic element. An elastic stress-strain curve was used to describe the mechanical behavior of tendon. The optimal control problem was to maximize the height reached by the body's center of mass, starting from a known initial state [4]. To quantitatively assess the model's response, experimental force-plate, limb motion, and EMG data were collected from five adult males, each instructed to perform a maximum-height squat jump with hands held at their hips.

RESULTS AND DISCUSSION There was good agreement between the model's response and experimental results. With the exception of the trunk, each subject's limb-segment displacements compared well with those predicted by the model. The absence of trunk counter-movement in the model prior to upward propulsion, while representing a qualitative difference, does not detract from the model's ability to quantitatively study the dynamical response of the whole body. The model's vertical and horizontal ground reactions compared well with those generated by each subject, and its on-off muscle excitation times were confirmed by recorded EMG results. In addition, jump height and lift-off time were both within 20 percent of the values calculated from force-plate data.

Fig. 1 shows how power is instantaneously transferred by RF alone. We computed *muscle power* by taking the dot product of muscle moment and segment angular velocity and found that *muscle power* is transported *from* the trunk (dotted line in Fig. 1(a)) *to* the tibia (dotted line in Fig. 1(b)). We also computed *joint power* by taking the dot product of muscle's contribution to an intersegmental force and the linear velocity of that joint. Since the *joint power* transported by RF *to* the tibia via the ankle joint was found to be almost zero (solid line in Fig. 1(b)), nearly all the *muscle power* delivered to the tibia by RF is transferred *into* the femur through the knee joint (dashed lines in Fig. 1(b) and (c)). Also, since femoral power output was nearly equal to femoral power input (Fig. 1(c)), almost all *muscle power* generated by RF is transported through the knee joint, femur, and hip joint, *to* the trunk. A similar trend was found for GAS, which instantaneously transports *muscle power* through the ankle joint, tibia, knee joint, and hip joint, in a *proximal* direction, back *to* the trunk.

We have found that the uniarticular muscles VAS and GMAX dominate total power delivered to the trunk for almost the entire jump, with RF, GAS, and HAMS contributing no more than 8

percent. During the final 10 percent of the jump, however, the ankle plantarflexors (SOL, OPF, and GAS) deliver substantial power to the trunk (as much as 30 percent of the trunk's total instantaneous power). It is particularly important that GAS contributes approximately one-third of the total ankle plantarflexor power delivered to the trunk, thereby influencing jumping performance appreciably. However, this biarticular actuator is the final muscle to be activated prior to lift-off (a result consistent with experiment), and it contracts to assist SOL and OPF in accelerating the foot, and therefore the trunk's center of mass, upwards. In this light, GAS plays a role that is no different from any of the uniarticular ankle plantarflexors.

Finally, to test the dependence of jumping performance on biarticular muscle function, we removed all biarticular actuators from the musculoskeletal model, and recomputed the optimal control solution. With only uniarticular muscles available for propulsion, the model was able to deliver a fully coordinated jump (e.g., physiologically realizable joint-angle variations). More importantly, we found performance to be degraded by only 20 percent. This was due entirely to a decrease in power delivered to the trunk in response to GAS's absence. We found, however, that performance could be restored (and in fact increased) with the addition of a uniarticular ankle plantarflexor that is structurally equivalent to GAS.

To conclude, our optimal control results indicate that biarticular muscles (like uniarticular muscles) transport power to the trunk. However, their function is less significant than previously suggested; biarticular actuators increase jump height by 25 percent rather than 100 percent [1]. Moreover, this increase in performance is almost entirely the result of power delivered to the trunk by GAS just prior to lift-off.

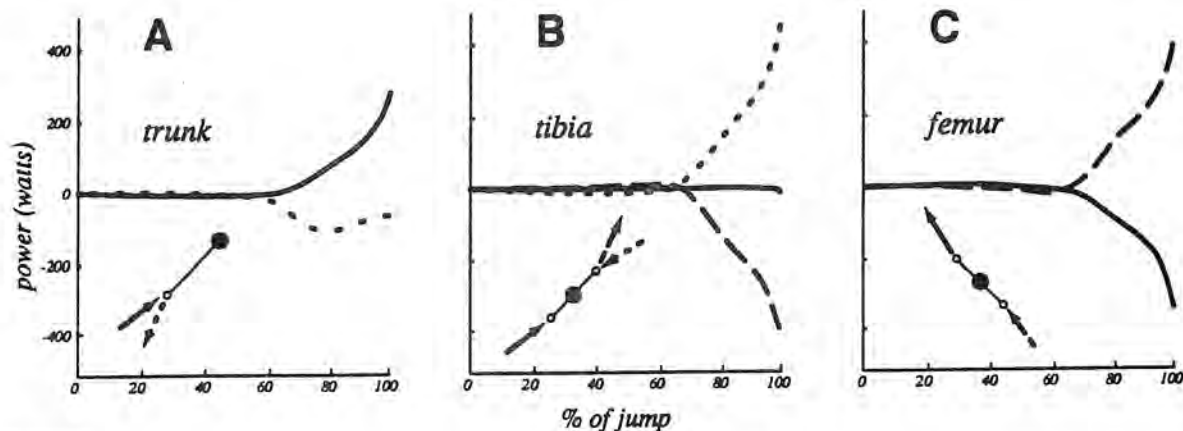


Figure 1

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SENSITIVITY OF SIMULATED HUMAN GAIT TO NEUROMUSCULAR CONTROL PATTERNS

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INTRODUCTION Learning to walk is a trying process. Balance, coordination, strength, and timing must come together simultaneously in order to enable the developing toddler to take a few steps. In the case of the spinal-cord injured, there is a possibility that controlled electrical stimulation of lower extremity muscles might someday enable them to walk again. While clinical research groups continue to develop methods of eliciting contractions from paralyzed muscle, we desired to explore the more global issue of coordinating these contractions in order to obtain gait-like motions. Our focus was to find a minimal set of muscles which, when controlled rather crudely, would still enable level and unsupported gait to be maintained at a normal cadence.

Forming a 3-D, 8 degree-of-freedom (DOF) model of the musculoskeletal system suitable for simulating both single-leg stance (SLS) and double-leg support (DLS), we found a reduced set of 9 muscles that allowed us to simulate normal gait. Coarsely modulated, near-optimal controls in the form of muscle activation levels were determined through dynamic programming, but were found to be very tenuous in nature. Key characteristics of the normal walking cycle were found to be particularly vulnerable to slight changes in the pattern of muscular activations. These findings confirm that neuromuscular control of multilinked dynamic systems is exceedingly difficult, and that artificial devices functioning in place of the central nervous system must quickly, and finely adjust musculotendon forces in order to react to body-segment trajectory errors.

REVIEW AND THEORY Dynamic programming is an alternative to static and variational optimization methods that can be used to determine dynamic optimal decision sequences for coordinated musculoskeletal motions¹. In contrast to optimal methods based on the calculus of variations, linearization of the equations of motion about an assumed trajectory or operating point is not required. Rather, in dynamic programming the equations are used exactly, but the controls are determined in combinations of M fixed, discrete levels instead of being allowed to vary continuously. Also, optimal control decisions in response to state feedback are applied only at specified intervals of time to minimize a criterion function, here based on body-segment trajectory tracking errors and muscle exertion². Since the method is computationally intensive, fewer control levels are allowed as the number of muscular controls (m) and DOF (n) increases. This results in controls that are coarsely adjusted, and updated only at discrete intervals of time.

METHODOLOGY For our purposes, $m=10$ and $n=8$ were determined to be the maximum numbers of musculotendon actuators and DOF that could be reasonably handled by our laboratory computer. The 3-D musculoskeletal model developed is suitable for simulating both the SLS and DLS phases of a single step, since a "soft" constraint under the swing-leg heel was modeled in lieu of hard kinematic constraints. Passive joint constraints were included at the knees and ankles as damped, nonlinear, torsion springs. Defining the step from 14% (left toe off, or LTO) to 62% (left foot flat, LFF) of the gait cycle enabled us to treat the right leg always as the stance leg, and the left as the swing leg. A lumped parameter model was used to replicate the tension-length and force-velocity properties of muscle and the compliant property of tendon.

Optimal muscle activation patterns were obtained by first deriving the discretized control law, and then applying it to simulate a step. A choice between a high and a low (typically zero) activation level ($M=2$) was given to each muscle. These two levels were adjusted in increments of 10% activation at gait cycle intervals of 4% (47 msec) via trial-and-error, until a near-normal step was obtained.

RESULTS AND DISCUSSION A set of 9 muscles was found to be sufficient to generate quasi-normal walking gaits. Figure 1 lists these muscles and the discretized set of neuromuscular controls. Activation of *gluteus maximus* was found to be undesirable, as it tended to pull the single-segment trunk into a backward fall. *Gluteus medius* and *minimus*, acting without the benefit of passive tension in the iliotibial band, was excessively activated

for most of SLS. EMG signals reported by others are shown to be similar to the muscle activations found here in some cases, but do not compare favorably overall, likely because the means and goals of artificially controlled and natural walking are different.

We found that simulated gait was extremely sensitive to small changes in the neuromuscular controls. The difficulty of obtaining a solution was compounded by inertial coupling among the links, which meant that trajectory errors at one link significantly affected the motions of the other links. For instance, the following is a list of problems requiring near-simultaneous solutions: stance leg collapse, lack of lateral stability, hyperextension of the swing-leg knee, lack of toe-heel clearance in the swing leg, and improper step length. If the stance knee flexed, or the stance ankle dorsiflexed about 10 degrees too much during early stance, the large moving mass of the head, arms, and trunk (HAT) led to rapid collapse. Heavily weighting trajectory errors in the cost function for these segments, plus the addition of an orthotic ankle support helped to solve this problem. The ankle orthosis, which consisted of a torsional spring exerting plantarflexion moments at the ankle when excessively dorsiflexed, also simplified the control of swing-leg foot orientation. Since the foot in free swing has little mass compared to potential dorsiflexion moments, it tended to dorsiflex to the limit (20 deg.) imposed by the passive joint constraints. Since this caused the heel of the swing leg to hit the ground prematurely (at ~38% of the gait cycle) as the leg was swinging through, the step length was often grossly shortened. Because activation of the dorsiflexors even at the lowest quantized control level (10%) led to this problem, the orthotic spring simplified the control by offering a measure of resistance without requiring coactivation by a crudely controlled, antagonist muscle. Once proper clearance of the swing leg was achieved, step length needed to be carefully regulated. If too long, the HAT tended to fall backward, because the upward and posteriorly directed joint reaction force at the swing-side hip resulting from foot-floor contact was absent. Interestingly, knee extension moments tended to decrease step lengths, as the strong extensors (eg. *vasti*) tended to decelerate, and in some cases reverse, the angular velocity of the swinging thigh, even when the hip flexors were active. As the resulting hyperextension of the knee was abrupt, and therefore undesirable, activation of *rectus femoris* alone provided the minimal levels of extensor moment required.

CONCLUSIONS It is possible to achieve quasi-normal, unsupported walking gaits using a reduced set of muscles, ankle orthoses, and crudely modulated control. However, the resulting gaits are highly sensitive to small perturbations in the controls, and probably to external disturbances as well. It is doubtful that such a control scheme could be made to work in practice. These results emphasize that artificial methods of controlling muscular contractions need to be capable of finely modulating activation levels in many muscles in order to enable the disabled to walk again.

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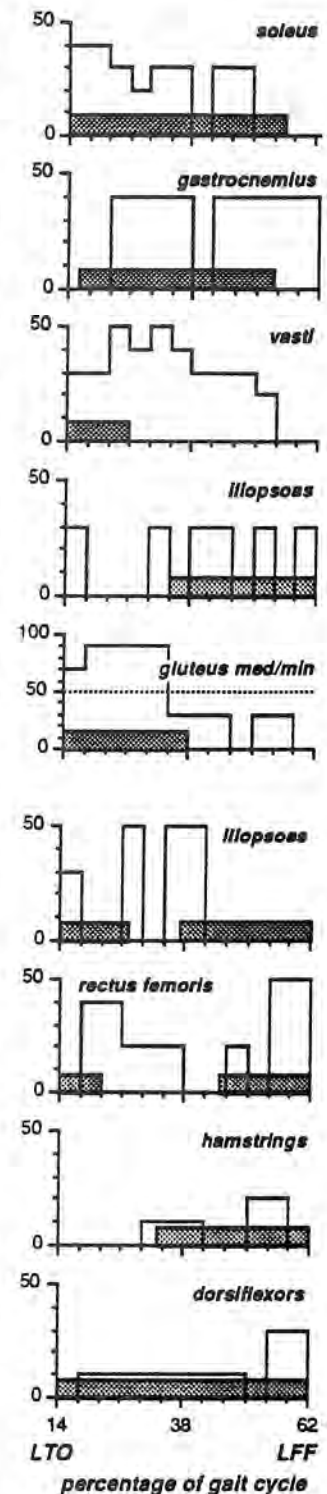


Figure 1. Percent muscle activations found in this study compared with reported EMG activity (shaded regions). The top five plots show activity in the right (stance) leg, while the bottom four depict activity in the left (swing) leg.

FEMORAL BONE ARCHITECTURE COMPUTED FROM 3-D MODELS RELATING BONE REMODELING TO STRESS HISTORIES

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Introduction: The ability to predict bone remodeling due to changes in physical activity or to the implantation of a prosthesis is an important goal of orthopaedic biomechanics. We have been developing a remodeling theory which uses the finite element method to calculate bone apparent density based on the stress history of the bone tissue(1). Recently, we have developed a time-dependent remodeling algorithm which takes into account the amount of internal bone surface area on which osteoblasts and osteoclasts can operate(2). Since bone resorption and formation can take place only on bone surfaces (3), internal remodeling, like external remodeling, should be considered as a surface phenomenon. Using the finite element technique, we have applied this theory to 3-D models of an idealized long bone and a proximal femur for a variety of loading histories.

Theory: The stress stimulus for remodeling, ψ , is defined using a daily load history approach, i.e., $\psi = \left[\sum_{\text{day}} n_i \sigma_i^m \right]^{1/2m}$, where, n_i is the number of cycles of load type i , σ_i is the peak energy stress associated with load i and m is an empirical constant. The remodeling technique incorporates two basic assumptions. 1) There is a single functional relationship between bone surface remodeling and the mechanical stress history(1). 2) When homeostasis exists, the bone mass is distributed in such a manner that the ratio of the stimulus, ψ_H , to the density is constant throughout the bone tissue. When homeostasis does not exist, the net linear rate of surface deposition or resorption ($\dot{r}$) is a function of the stimulus, ψ , the difference between the actual stimulus and the homeostatic stimulus, $\Delta\psi = \psi - \psi_H$, and the available bone surface area on which remodeling can take place.

Carter (4) proposed that bone was "lazy" — that significant remodeling would not begin until a certain level of over or under-load occurred. The relationship used in the present work was a tri-linear rule with transitions at $(1 - w)\psi_H$ and $(1 + w)\psi_H$,

$$\dot{r} = \begin{cases} c_1(\Delta\psi/\psi_H) + (c_1 - c_2)w; & \psi < \psi_H(1 - w) \\ c_2(\Delta\psi/\psi_H); & \psi_H(1 - w) \leq \psi \leq \psi_H(1 + w) \\ c_3(\Delta\psi/\psi_H) + (c_2 - c_3)w; & \psi > \psi_H(1 + w). \end{cases} \quad (1)$$

In this equation, which is applicable to both internal and external remodeling, c_1, c_2, c_3 are empirical rate constants and w is the non-dimensional half-width of the lazy zone. The internal bone surface area available for remodeling has been measured as a function of tissue apparent density(3,5). For internal remodeling, the rate of change of density, $\dot{\rho}$, is then given by

$$\dot{\rho} = \rho_t S_v \dot{r}, \quad (2)$$

where ρ_t is the true density of the bone tissue and S_v is the bone surface area density.

Methods: Discretized forms of Eqs. 1 and 2 have been applied to three-dimensional models of an idealized long bone and a proximal femur. Figure 1 shows the finite element mesh of the proximal femur. In the long-bone model, loading histories included contributions from bending, axial compression and torsion. In the proximal femur, loading histories were composed using *in vivo* pressure distributions reported in the literature(6) for the various portions of the

walking cycle. Both models were initially assumed to consist of solid bone tissue of constant density throughout. The rate constant, c_2 , was chosen to be equal to one-fifth the value of the other two rate constants ($c_1 = c_3$). Progressive remodeling of the internal bone structure was accomplished by iteratively changing the density and material properties as calculated from finite element solutions in which time-dependent remodeling algorithms were embedded.

Results: In the long bone, the combination of compression, bending and torsion in multiple planes led to the progressive remodeling of a solid cylinder into a bone structure consisting of a contiguous cortex of dense bone surrounding a minimally dense medullary region. This did not occur for bending in only one plane. With bending in a single plane, the addition of a torsional load component was necessary to generate an annular cortex. In the proximal femur, using multiple loads to represent the walking cycle, several characteristics of a natural femur were evident from the remodeling solutions. A region of dense trabecular bone emanating from the superior articular surface, Ward's triangle, and a contiguous cortex of cortical bone distally were apparent. Figure 2 shows the calculated apparent density distributions for a coronal section of the remodeled femur. If only one loading case was applied, all of these characteristics were not evident.

Discussion: In the present study a time-dependent bone remodeling theory which accounts for the amount of internal surface area has been applied to three-dimensional models. The success of this theory in emulating the morphogenesis of the normal bone structure of the femur is extremely encouraging. Preliminary applications of the theory to bones with orthopaedic implants are also encouraging and suggest that the functional adaption of bone to mechanical stresses can be effectively emulated using computerized numerical techniques.

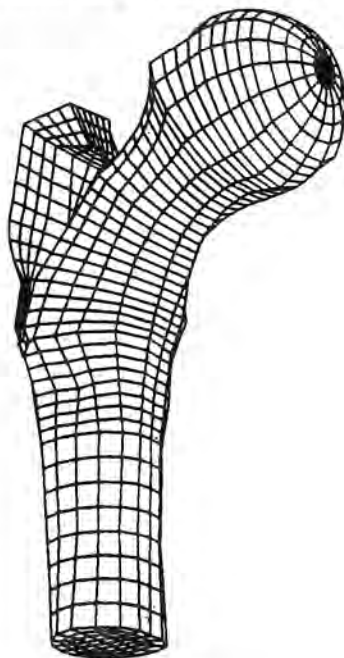


Figure 1

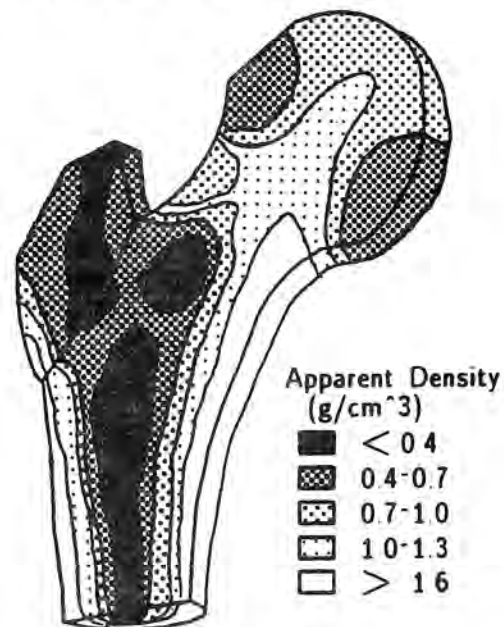


Figure 2

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INTERPRETATION OF MUSCLE ACTIVITY DURING CYCLING BASED ON A DYNAMICAL, TWO-LEGGED BIOMECHANICAL MODEL

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INTRODUCTION

Biomechanical studies of bicycle pedaling have had limited success in explaining muscle function and coordination during cycling. Despite the vast experimental data on bicycle pedaling, a suitable theoretical framework is needed to interpret these experimental findings. We have developed a dynamical bilateral model of seated bicycle pedaling to interpret activity of muscles based on the angular accelerations they induce on the crank and limbs. Assuming that each muscle should be activated so as to overcome friction by positively accelerating the crank, we found predicted muscle activity to compare very favorably with EMG data found in the literature^{2,5,6}, except for uniaxial hip muscles. We found that hip musculature contributes insignificantly to crank acceleration but rather provides for intralimb coordination.

REVIEW AND THEORY

Although biomechanical models of the human lower extremity have been employed, for example, to compute net muscle joint torques from pedal reaction force data, no fully satisfying interpretation of observed muscle EMG activity has yet been proposed (consider the problem of Lombard's Paradox, for example^{1,3}). Since the dynamical matrix equations of motion describe how a system will move under the influence of all applied forces, we can use these equations to relate muscle joint torques to the accelerations they induce⁸. The equations are:

$$M(\theta)\ddot{\theta} = T(\theta) + V(\theta, \dot{\theta}) + G(\theta) + F \quad (1)$$

where $M(\theta)$ = mass matrix
 $T(\theta)$ = muscle contribution
 $V(\theta, \dot{\theta})$ = velocity contribution
 $G(\theta)$ = gravity contribution
 F = friction contribution

Accordingly, muscle, velocity, gravity, and friction contributions to the acceleration of the bicycle/rider system can be found by multiplying (1) by $M^{-1}(\theta)$.

$$\ddot{\theta} = M^{-1}(\theta)T(\theta) + M^{-1}(\theta)V(\theta, \dot{\theta}) + M^{-1}(\theta)G(\theta) + M^{-1}(\theta)F \quad (2)$$

Notice that the angular acceleration of each generalized coordinate is the sum of the accelerations caused by muscle, velocity, gravity, and friction terms. Thus, a single muscle's contribution to acceleration can be found by entering its joint torque(s) into $T(\theta)$ in (2) while ignoring all other terms. In general, a muscle will induce angular accelerations in all the coordinates due to the nature of $M^{-1}(\theta)$.

To interpret how a muscle will act to accelerate the crank and limb segments per unit of muscle force, we need only specify the muscle's moment arm trajectory over the crank cycle. In this way, we can investigate whether a muscle would positively or negatively accelerate the crank and limb segments if it was active and generating a unit of force.

METHODOLOGY

We derived the equations of motion for a two-legged, three degree-of-freedom planar cyclist on a stationary bicycle. Each side of the cyclist was considered to be a five-bar linkage coupled through the crank. We used anthropometric and bicycle parameters along with nominal kinematic crank and pedal data reported in the literature for a bicycle ridden on rollers at 80 RPM⁷. We computed hip, knee, and ankle joint trajectories from crank and pedal angle data. Muscle moment arms were computed from the human musculoskeletal model developed by Hoy *et al.*⁴ Ergometer flywheel inertia and friction parameters were chosen to simulate the dynamic response of a seated cyclist riding a bicycle on rollers.

To gain confidence in our model, we computed total muscular contribution to the equations of motion, which is the $T(\theta)$ vector in (1), by two approaches. Given a reported nominal cycling trajectory which assumes zero crank acceleration at each instant⁷, one can find $T(\theta)$ over the crank cycle from (3):

$$T(\theta) = M(\theta)\ddot{\theta} - V(\theta, \dot{\theta}) - G(\theta) - F \quad (3)$$

Comparison of this result with $T(\theta)$ derived directly from net muscle joint torque estimates based on experimental data⁷ indicated that crank acceleration must vary periodically throughout the crank cycle.

This acceleration variation is, however, extremely small and caused the 80 RPM cadence to vary by only $\pm 1\%$.

RESULTS AND DISCUSSION

Velocity-induced and gravity-induced crank accelerations were found to be small and more or less in opposition over the crank cycle. Thus, to achieve an approximately constant crank angular velocity, crank acceleration resulting from all the muscles must be positive to balance the constant negative crank acceleration resulting from friction. On this basis alone, we would only expect muscles to be active when they generate positive crank acceleration.

Based on regions where muscles induce positive crank acceleration, we predicted when we would expect individual muscles to be highly active. Figure 1 compares our predictions with previously reported experimental regions of EMG activity⁵. Notice that there is reasonable agreement for all muscles except gluteus maximus (GMAX). For muscles crossing the ankle or knee (hamstrings - HAMS; rectus femoris - RF; vastus medialis and lateralis - VAS; gastrocnemius - GAS; tibialis anterior - TA), experimentally observed activity commences and terminates sooner than predicted (compare filled bars with open bars). These differences may be due to activation and musculotendon contraction dynamics.

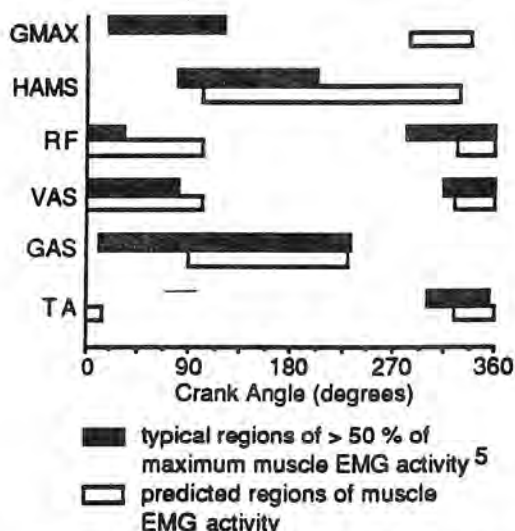


Fig. 1. Comparison of predicted and reported regions of muscle EMG activity.

The reason for good agreement at the ankle and knee and bad agreement at the hip may be due to the relative effectiveness of ankle, knee, and hip muscles

in accelerating the crank. That is, we found that a unit of hip torque produces roughly 1/20th as much crank acceleration as a unit of ankle torque and roughly 1/8th as much crank acceleration as a unit of knee torque. Hence, given the choice between ankle, knee, and hip muscles, we would expect the body to activate ankle and knee muscles when they produce positive crank acceleration and to not concern itself with the small negative crank acceleration resulting from hip muscles. At this point, it is also interesting to note that although RF and HAMS both span the hip and knee, their predicted muscle activity agrees quite well with reported EMG activity^{2,5,6}. Because of the ineffectiveness of a unit of hip torque in producing crank acceleration, we believe RF and HAMS are active not because of the larger joint torques they produce at the hip but because of the positive crank acceleration produced by their smaller joint torques at the knee.

Since hip muscles in subjects are not active at times when they produce positive crank acceleration, they must be active for some other reason. Once the crank acceleration is controlled, only one degree of freedom is left to control in each leg. This is because each side of the cyclist was modeled as a five-bar linkage coupled through the crank. In addition, our model showed that for cycling on rollers, muscles on one leg induce only negligible accelerations on the segments of the other leg, so that cross-coupling acceleration effects between the two legs can be ignored. Finally, we found that net muscle hip, knee, and ankle torques are capable of producing limb accelerations of the same order of magnitude. Hence, we submit that uniarticular hip muscles on either leg act to produce the additional acceleration control needed to provide for intralimb coordination of that leg during the pedaling cycle.

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A Computer Graphic System to Study Human Movement

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Introduction

Because computer graphics systems are such an effective means to visualize and help understand complex technical problems, they have been under vigorous development for the last two decades. Nevertheless, there are still very few computer graphics programs for biomechanical analysis. We have developed a set of computer graphics tools to study human movement. We present here systems to animate the results of computer simulations and biomechanical experiments, and to model the musculoskeletal system.

In the animation system, we create windows containing, for instance, a three-dimensional skeleton that moves when driven by the experimental data or the simulation results. Each of the windows can easily be moved and resized, using a mouse and screen cursor, so that experimental results can be compared with an animated simulation shown in an adjacent window. We also create windows containing plots of the data and bargraphs showing each muscle's contribution to the acceleration of the body segments³.

The Animation System

A Silicon Graphics Iris 2400T workstation is the host for our computer graphics applications. We use the workstation to animate the results from both biomechanical experiments and dynamical simulations of human movement (jumping, walking, bicycling, and postural control). The simulations are generally run on either a Vax 11/780 or a Microvax 3200. During experiments, kinematic, force plate, and EMG measurements are recorded with a motion analysis system. Both the simulation results and the motion analysis data are first formatted and then transferred to the animation system for display and analysis (Fig. 1)

We have found that using computer graphics to visualize and interact with biomechanical data and models is helpful to insure the integrity of analyses, to provoke insights, and to communicate those insights to others. For example, while a simulation may stay within mathematically imposed constraints, animating the simulation has, at times, revealed obviously unrealistic movement. In our animations of walking, subtle gait abnormalities, not apparent from examining joint angle/velocity plots, became evident. Comparing animated results of our experiments with animated simulation results has also been helpful in defining our future modeling efforts. The most common use of the animation system is, however, as a tool for communication with our colleagues and visitors.

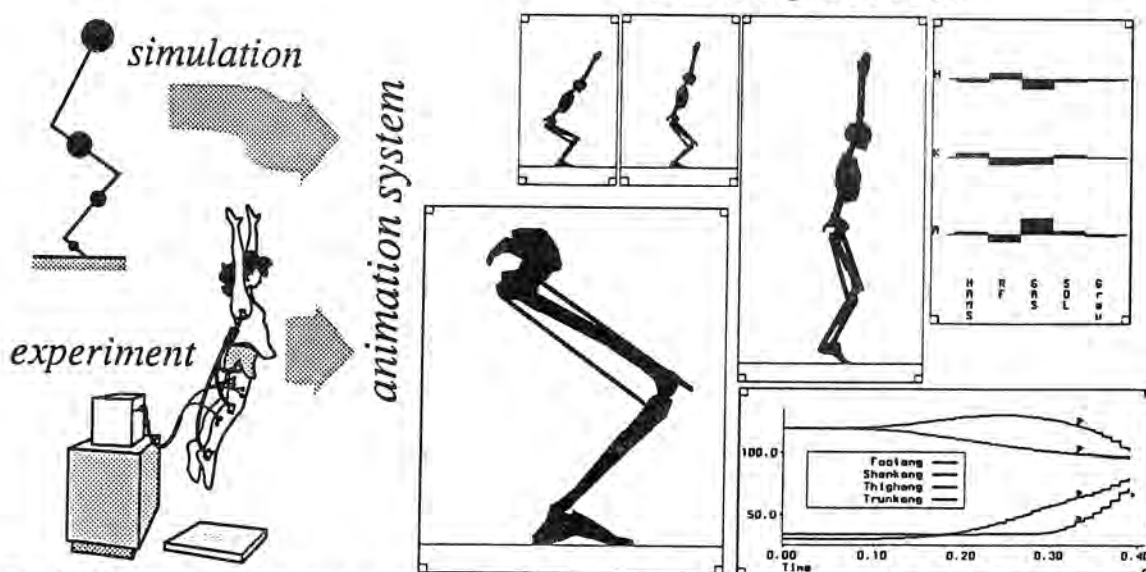


Fig. 1. We use computer graphics to animate and interact with experimental and simulation results.

The Musculoskeletal Model

We also use computer graphics to study the effects of musculoskeletal geometry (i.e., the geometric relationship of muscles and bones) on muscle-tendon function. To acquire the coordinates needed to display the bone geometry we digitized the surfaces of lower-extremity bones with a Polhemus three-dimensional digitizer. Using the graphics system, we translated and rotated the digitized bones into an anatomical configuration. The coordinates representing the pelvis, femur, and tibia were then scaled to fit origin and insertion coordinates reported by Brand *et al.*¹. We display some muscles as straight lines (Fig 2). However, when a muscle's path is altered by bony or soft tissue constraints, we display it as a series of line segments. Since the muscle coordinates are defined with respect to the joints, muscle moment arms and origin-to-insertion lengths can be computed. Also, by combining a mathematical model of muscle and tendon⁴ with the musculoskeletal geometric model, we can compute muscle force and its moment about the spanned joint(s) for any skeletal position⁴.

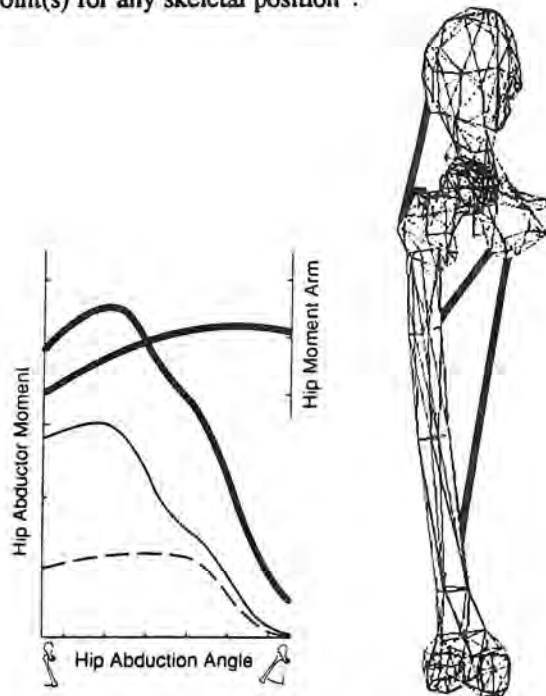


Fig. 2. We use a computer graphic model of the lower extremity to study how properties of muscle, tendon, and skeletal geometry affect joint moments.

We use the musculoskeletal model to study orthopaedic surgical procedures. For example, to study the effect a Chiari osteotomy on hip abduction moment, we adjusted the hip geometry according to a set of surgical parameters². We then used the musculoskeletal model to determine how the geometric changes affect hip abduction moment. Currently, we are using the system to simulate an Achilles tendon lengthening with concomitant anterior transfer of the tibialis posterior. To simulate this surgery we adjust the model's Achilles tendon parameters and, on the computer graphics system, detach the tendon of tibialis posterior, and transfer it to the dorsum of the foot. The musculoskeletal model then computes a set of post-surgery muscle forces and moments, and displays the resulting plantarflexion/dorsiflexion moments as a function of ankle angle.

The advantage of using computer graphics to simulate surgery is that it allows us to visualize the complex three-dimensional geometric relationships among the muscles and the bones. By rotating and zooming-in on the geometric model of the lower extremity we can visually verify its accuracy. Also, since we can change a muscle's parameters (attachment location, strength, fiber length, and tendon length) and receive immediate visual feedback, we have gained insight into the sensitivity of computed muscle forces and moments to the parameters of the model and of the surgery.

Acknowledgement

This work is supported by the Veterans Administration, a NSF fellowship to SLD, and NIH grant NS17662. We thank Gary Yamaguchi for the simulations of human walking.

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MUSCLE'S ACTION TO PRODUCE JOINT ANGULAR ACCELERATIONS WITH APPLICATION TO STANDING

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INTRODUCTION

Understanding how force in a muscle acts to rotate joints is fundamental to the treatment of motor disorders and to analysis of athletic performance. The anatomical location of a muscle determines unequivocally the directionality of its moment(s) about the spanned joint(s). Action of a muscle is often inferred from the directionality of its moment(s); e.g., a knee flexor muscle, one that develops a knee flexor moment, is assumed to act to flex the knee. If the same muscle should be biarticular and also generate a hip extensor moment (e.g., hamstrings), the muscle is assumed to act to extend the hip.

Muscle action on joint rotation cannot, in general, be inferred from anatomy alone.^{1,2} For example, a biarticular muscle, like hamstrings or gastrocnemius, that generates a flexor moment about the knee may act to extend (*sic*) the knee. We present a general method, based on dynamics, for finding the action of a uni- or multi-articular muscle during a motor task. We have applied the method to develop general principles of muscle action during multiarticular movement, and to understand the specific action of the lower-limb musculature during standing.

GENERAL METHODOLOGY AND RESULTS

For a multi-jointed body, the relation between muscle force and angular acceleration depends on joint and body-segmental properties, and the interactions among segments and objects external to the body (e.g., the ground). Thus muscle action is task-dependent. However, the dynamical equations of motion applicable to a specific task can be derived and analyzed to find how muscle force accelerates each of the joints during the task.

Assume that the body can be represented as a set of rigid segments connected by frictionless pin or ball-and-socket joints, and that only one segment (or part of one) is stationary (e.g., as in standing or walking). The dynamical matrix equations are:²

$$M(\theta)\ddot{\theta} = T_{mus} + T_{vel} + T_{grav} \quad (1)$$

This equation is a matrix torque-balance equation, written in joint-space. θ is the vector of joint coordinates. The joint-torques from angular accelerations of the joints ($M(\theta)\ddot{\theta}$) balance the sum of the joint-torques developed by muscles (T_{mus}), the angular velocity of the joints (T_{vel}), and gravity (T_{grav}). Solving for the angular accelerations ($\ddot{\theta}$), we have:

$$\ddot{\theta} = M^{-1}(\theta) (T_{mus} + T_{vel} + T_{grav}) \quad (2)$$

Each joint angular acceleration in the vector $\ddot{\theta}$ is the sum of the accelerations caused by the muscle, joint angular velocity, and gravity joint-torques.

The contribution of one muscle to the angular acceleration of the joints is found by substituting its joint-torque(s) into (2) and setting all other terms to zero. A muscle generates joint-torques only at the joints it spans. If a spanned joint is a ball-and-socket joint, for example, the muscle will generate three joint-torques, one for each of the three degrees-of-freedom (DOF) of the joint.³ The three joint-torques are given by the three components of the moment of muscle force about the joint. Importantly, anatomy specifies the direction of a muscle's moment about the joint (e.g., flexor vs extensor moment). Thus anatomy alone specifies the sign of a muscle's joint-torque(s).

Assuming that the body is constrained so that only one-joint can move, and in only one direction, then anatomy alone determines the direction of muscle action. For example, if only one joint can flex or extend, then any uni- or multiarticular muscle crossing the joint and developing a flexor torque accelerates the joint into flexion (i.e., acts to flex the joint). Thus, in such highly-constrained, single-joint motor tasks, a flexor muscle will indeed act to flex the joint, and an extensor muscle to extend the joint.

In multi-articular movement, the direction of muscle's action is more complex. We can show that:

1. A uniarticular muscle crossing a pin-joint will always accelerate the spanned joint in the same direction as its joint-torque. As in one-DOF, one-

joint movements, anatomy alone determines the action of a muscle to produce angular acceleration of the spanned joint.

2. A uniarticular muscle crossing a ball-and-socket joint may accelerate the joint in directions that are opposite to some of its applied joint-torques. Thus, anatomy does *not* by itself specify how a uniarticular muscle will accelerate a spanned ball-and-socket joint.

3. A uniarticular muscle crossing a pin or ball-and-socket joint accelerates unspanned joints as well. This action can be more powerful than that on the spanned joint. The directionality of the induced angular acceleration of the unspanned joint depends on all the joint angles, all the inertial parameters, and the task being performed (i.e., $M^{-1}(\theta)$). Thus, anatomy alone does *not* specify how a muscle acts on unspanned joints.

4. Multi-articular muscles also cause angular acceleration of the unspanned joints.

5. A biarticular muscle crossing two pin-joints can accelerate one of the two spanned joints in a direction opposite to its applied joint-torque. Again, anatomy does *not* by itself specify how a biarticular muscle accelerates the joints it spans.

APPLICATION TO STANDING

The equations of motion were derived for a planar, three-segment model of a human standing flat-footed. Pin-joints were assumed. Body segmental parameters of adults⁴ and children⁵ were used. A muscle's contribution to the angular acceleration of one joint relative to its contributions to the other joints was computed from (2). The net contribution of a biarticular muscle was found by summing the contributions caused by each of its two joint-torques. Since a biarticular muscle's net contribution depends on the ratio of its moment arms at the two joints, we studied how its moment arm ratio affects the joint angular accelerations it induces. Muscle's contribution to the linear acceleration of the body's center of mass (c.m.) was also computed.

We found for postures near the vertical that:

1. Uniarticular extensor (flexor) muscles crossing the ankle, knee, or hip accelerate all joints towards extension (flexion).
2. Uniarticular ankle, knee, and hip muscles all accelerate the joints in about the same proportion. The amount of angular acceleration of the ankle relative to the knee to the hip caused by each muscle is about 1 : 2.3 : 1.6.
3. Uniarticular ankle and hip extensor muscles accelerate the body's c.m. posteriorly, and knee extensor muscles anteriorly. Flexor muscles produce the opposite action. Assuming equal joint-torques,

the linear acceleration of the c.m. induced by ankle, knee, and hip muscles is about 1 : 1.3 : 0.4.

4. Gastrocnemius (GAS) accelerates all joints towards flexion when its knee to ankle moment arm ratio is greater than 0.49; for just slightly less ratios, it accelerates all joints towards extension.

5. Hamstrings (HAM) accelerates all joints towards flexion when its knee to hip moment arm ratio is greater than 0.52; for just slightly less ratios, it accelerates all joints towards extension.

6. Rectus femoris (RF) accelerates all joints towards flexion when its knee to hip moment arm ratio is less than 1.1; for just slightly higher ratios, it accelerates all joints towards extension.

7. GAS, HAM, and RF's action to flex or extend a joint is sensitive to its moment arm ratio, since its moment arm ratios are near these critical values.

8. GAS, HAM, and RF's action is also sensitive to deviations of the body from the upright position.

9. Angular accelerations induced by uniarticular muscles are relatively insensitive to changes in inertial body-segmental parameters occurring during ontogeny.

DISCUSSION

Muscle's action to produce angular acceleration of the joints of the body is complex for multiarticular motor tasks. The complexity arises because the body is articulated and forces at one end of the linkage are transmitted through the joints to the other end. Thus, forces generated by muscles crossing one joint cause angular acceleration of all the joints, and the accelerations of unspanned joints can be higher than the accelerations of the joints the muscle spans. Unfortunately, muscle's action is task-dependent because the dynamical equations describing the relation between muscle force and acceleration depend on the interactions among the body segments and the environment (e.g., the ground). Since anatomy does not alone specify the dynamical equations, it is, in general, impossible to infer even the directionality of muscle action from just the anatomical location of a muscle.

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DEVELOPMENT OF A WEIGHTED VERTICAL JUMP TEST FOR ICE HOCKEY

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Introduction:

The purpose of the study was to examine the relationship between on-ice skating performance and a vertical jump test completed both with and without a weight vest. Successful execution of a vertical jump demands a pattern of hip, knee, and ankle extension similar to that in the hockey stride yet most of the attempts to measure strength in hockey players have used isolated joint testing. In addition, both the skating stride and the vertical jump utilize active and passive potentiation from the prestretch during the concentric phase of the movement. There is a need for a test that reflects changes in performance-related multi-joint strength patterns following training.

Methods:

The subjects were 24 male, Bantam (14 and 15 years of age) ice hockey players. One group (n=14) participated in a 7-week on- and off-ice training program specific to hockey while a second group (n=10) served as a control group. Both groups were tested before and after the 7-week training period.

For the vertical jump test, each subject completed 4 jumps from a force platform. The subject began the trial standing motionless on the force platform and was instructed to jump as high as possible without taking any preliminary steps. For the last 2 jumps, the subject wore a vest weighing 25% of their body weight. The force platform output signal representing the force along the vertical axis was amplified and sampled at 500 Hz for a 4 second period. Vertical velocity at take-off was determined according to the impulse-momentum principle with the subject's mass calculated from the force platform output while the subject was stationary. The vertical displacement of the center of gravity was derived from the take-off velocity. The mean value for the 2 jumps under each condition was used in the analyses.

On-ice performance was assessed by 3 tests. The top speed test is a measure of the time to skate 15.25 meters given a moving start. The acceleration test measures the time to skate from 3 to 27.5 meters from a stationary start. To test cornering ability, the time to complete an S-shaped path around the face-off circles is determined. All tests were electronically timed and the mean score of 2 trials was recorded.

Results:

A repeated measures analysis of variance was used to determine the effect of the 7-week training program on vertical jump height and on-ice skating speed. For the training group, significant pre- to post-training differences ($p < .01$) were observed for the weighted jump scores and the three on-ice measures. Significant test date by group interactions were present for the three on-ice measures. An analysis of pre- to post-training scores for the two groups is presented in Table 1.

Table 1
Mean Pre- and Post-Training Scores

Variable	Training Group (n=14)		Control Group (n=10)	
	Pre	Post	Pre	Post
Jump Ht (cm)	29.85	33.49	30.91	31.12
Wt. Jump Ht. (cm)	19.24	21.41 ^a	18.94	19.68
Top Speed (s)	1.80	1.72 ^a	1.83	1.80
Acceleration (s)	3.74	3.65 ^a	3.74	3.74
Cornering (s)	6.80	6.40 ^a	6.78	6.65

^a p<.01

Pearson-product moment correlation coefficients were calculated to examine the relationship between the on-ice measures and the vertical jump scores. Data from all 24 subjects were included in the analysis (Table 2)

Table 2
Correlation of Vertical Jump Heights and On-ice Measures (n=24)

On-ice Test	Pre-training		Post-training	
	Jump	Jump w/Vest	Jump	Jump w/Vest
Top Speed	-0.79	-0.71	-0.70	-0.75
Accel.	-0.81	-0.79	-0.76	-0.83
Cornering	-0.36	-0.39	-0.57	-0.69

The results indicate that on-ice performance was significantly improved for the training group as was the weighted vertical jump score. The correlations between the jump scores and the on-ice top speed and acceleration tests were unchanged as a result of training while the correlations between the jump scores and the cornering test were significant (p<.01) only following the 7-week training period. This may be due to an increased heterogeneity of the cornering test scores following the training program. The results would suggest that a weighted vertical jump test may be capable of reflecting changes in hip, knee, and ankle extension strength and recoil potentiation related to ice hockey as a result of training.

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DYNAMIC TRANSFER OF BODY MASS FROM BIPEDAL TO SINGLE LIMB STANCE

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INTRODUCTION

The dynamic transfer of body mass laterally is a necessary prerequisite to the performance of single leg flexion movements while standing, as well as the initiation and ongoing execution of locomotion. Although the former task lacks the forward propulsion component of the latter, both movements require the transfer of body mass from bipedal to single limb support. The initiation of gait has been found to be accompanied by 1) an initial increase in the vertical component (F_z) of the ground reaction force (GRF) beneath the stepping leg [2], and 2) an initial displacement of the center of pressure towards the swing limb [3]. These observations have led to the proposal that an initial displacement of the body center of mass (CM) towards the flexing (fl) limb resulted in the increase in F_z needed for the subsequent lateral transfer of the CM towards the single stance (st) leg [3]. This view has been challenged [1] based on the predictions of a dynamic model. Moreover, such a theory of lateral weight transfer was not supported by the direct measurement of the trajectory of the body CM, and the corresponding horizontal component (F_y) of the GRF. The purpose of this investigation was to examine the GRFs and the trajectory of the body CM in the frontal plane during standing leg flexion movements performed at different speeds.

METHODS

Eight healthy subjects (22 to 39 years) stood on two separate force platforms (AMTI) and withdrew one lower limb from the support surface through a combined flexion movement. Five to ten trials were performed at each of three speeds: 'as fast as possible' (F), natural speed (N), and 'as slowly as possible' (S). A motion analysis system (Watsmart) was used to locate the positions of the body landmarks, and a five-segment model (feet-shanks, thighs, head-arms-trunk) used to determine the instantaneous location of the body CM in the frontal plane.

RESULTS AND DISCUSSION

The earliest decreases in F_{zfl} for fast and natural speeds were always preceded by an earlier abrupt increase in F_{zfl} over quiet stance (Fig.), while such early increases did not regularly occur for slower speeds (Fig.). The means and standard deviations of the peak vertical forces beneath the flexing limb prior to the onset of unloading were $322.1 \pm 81.8N$, $134.9 \pm 29.6N$, and $20.4 \pm 10.6N$ for fast, natural, and slow speeds respectively. While the data suggested that the magnitude of the early increases in F_{zfl} was proportionate to the speed of movement, there were numerous trials during slow speeds in which this initial increase in F_{zfl} did not occur (Fig.). In all cases, the earliest changes in the resultant horizontal component (F_y) of the GRFs acting on the body in the frontal plane preceded the onset of unloading and were exerted in the direction of the stance limb (Fig.). These results indicated that the horizontal component of the GRF was directly responsible for the linear displacement of the body CM towards the stance limb. For F and N movements, the horizontal force recorded beneath the flexing limb (F_{yfl}) contributed a much greater

proportion ($79 \pm 12\%$ and $80 \pm 9\%$ respectively) to the resultant F_y than F_{yst} measured beneath the stance limb for F ($21 \pm 8\%$) and N ($20 \pm 8\%$) speeds. In contrast, slower movements showed a marked change in the GRF distribution whereby F_{yst} and F_{yfl} became $48 \pm 17\%$ and $52 \pm 17\%$ of the resultant horizontal force.

Examination of the motion analysis data also indicated that the displacement of the body CM in the frontal plane was directed immediately towards the single stance limb regardless of the speed of leg flexion. Therefore, the early increases in F_{zfl} were not associated with a shift in body weight to the flexing leg prior to its removal from the stance support surface as suggested previously [3]. Rather, the latter likely reflected the need to create an angular acceleration to rotate the body laterally in order to withdraw the limb from the floor.

Overall, the results indicated that the transfer of body mass laterally during more rapid speeds of leg flexion movements were generated by a coupling of both an initial increase in the pressure (F_{zfl}) and the frictional (F_{yfl}) components of the GRF beneath the flexing limb. During very slow speeds of leg flexion a different control strategy was adopted whereby the frictional force beneath the stance limb was increased in the presence of a marked reduction and/or absence of an initial increase in the vertical force beneath the flexing leg. The speed related differences in the kinetic profiles suggested different control strategies for the displacement of body mass laterally, which appeared to reflect the requirements necessary for overcoming the inertial force of the body mass during transfers from bipedal to single limb stance.

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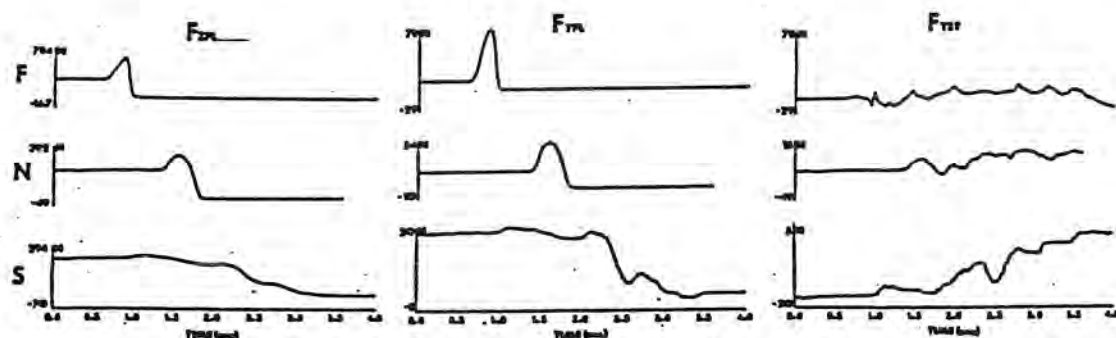


Fig. Representative example from one subject showing the vertical (F_z) and horizontal (F_y) components of the ground reaction forces (GRFs) recorded beneath the flexing (fl) and stance (st) limbs during leg flexion movements performed at fast (F), natural (N) and slow (S) speeds. The positive F_y indicates the flexing limb to stance limb direction.

PREDICTING TRABECULAR BONE STRENGTH AND MICRO-STRAINS USING
HOMOGENIZATION THEORY: Application to a Vertebra

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INTRODUCTION: Using a recently developed elasticity theory for the mechanics of microstructured materials (homogenization theory) the dependence of trabecular bone strength on volume fraction of bone was calculated. The results showed a quadratic dependence for loadings along the principal strength axes, and showed a cubic dependence of strength on solid volume fraction for loading in shear. These results are consistent with the theoretical predictions of other investigators and with experimental results (see 1). The assumed failure mechanism was tensile tissue rupture rather than the more commonly assumed trabecular buckling failure.

The homogenization approach also provides the means of approximating the micro-strains in the trabeculae by post-processing the results of standard finite element analyses. Using an axi-symmetric finite element model of a lumbar vertebra it was demonstrated that the pattern of peak strain energy density (SED) in the trabeculae is different from the pattern of apparent SED calculated using the standard finite element method. Hence, peak trabecular micro-SED (micro-strain) is not related in a straightforward way to the apparent SED (strain) calculated by standard finite element analysis.

PREDICTION OF TRABECULAR BONE MICRO-STRAIN AND STRENGTH: In the homogenization method, trabecular bone is approximated as a periodic cellular structure. The basic cell used in this analysis was a 1mm cube with a spherical void inclusion (Fig. 1) similar to the finite element model of Beaupre and Hayes (2). For solid volume fractions less than 0.4 the cells were open (the sphere penetrated the sides of the cube) and for volume fractions of 0.4 or greater the cells were closed. The basic theory of this method (e.g., 3) and the details of developing the micro-cell for trabecular bone (4) are not reported here. Fig. 1 shows the predicted micro-SED pattern in the cells of a porous material ($vf=0.21$) subjected to uniaxial stress. The predicted maximum energy occurs at the rim of the hole, as expected. It is important to note that this is not a standard finite element result. The program used to calculate the micro-SED pattern for the 192 elements ran in 2 minutes on an 80286 based personal computer. Similar analyses were run for cells with volume fractions between 0.1 and 0.9. Using a maximum principal tensile strain failure criterion for the trabecular tissue, the strength of the structure relative to the strength of the tissue was predicted for uniaxial tension, compression and pure shear (Fig. 2). The compressive and tensile strengths were proportional to volume fraction squared. Shear strength was proportional to volume fraction cubed.

ENERGY PATTERN IN A LUMBAR VERTEBRA: An idealized axisymmetric finite element model was developed from measurements of a lumbar vertebra taken from a 49 year old male. The material properties of the cortical bone and disc were taken from the literature. The trabecular bone had a volume fraction of 0.21 and had cubic

material properties calculated using homogenization theory of $E=538.8\text{MPa}$, $G=83.6\text{MPa}$, $\nu=0.0863$. The apparent SED calculated using a standard finite element model from a uniform displacement of the end of the disc is shown in Fig. 3. In Fig. 4 the peak predicted trabecular (micro) SED is plotted. The peak micro-SED is much larger than the apparent SED and the pattern of the micro-SED differs from that of the macro-SED. In the standard model the maximum SED occurs near the center of the end plate. In the micro-SED pattern the maximum trabecular energies are away from the center, nearer to the cortex.

DISCUSSION: The results of this study demonstrate that the spherical void model has a quadratic dependence of tensile and compressive strength on volume fraction without assuming trabecular buckling failure. This finding suggests that trabecular bone may fail by tensile rupture of the mineralized tissue. Since the shear strength has a cubic dependence and will be less strong, experimental determination of the tensile and compressive strengths will be difficult. Unless the specimen is tested exactly along a principal strength axis the shear failure mode will dominate.

The difference between the patterns of the peak micro-SED and the apparent SED may be relevant to trabecular bone remodeling. Since trabecular bone probably remodels in response to the distribution of micro-stress (strain) in the underlying tissue, a microstructural model may be necessary to determine accurately the bone remodeling pattern. This method may be a valuable tool for understanding the link between applied stress and trabecular bone remodeling.

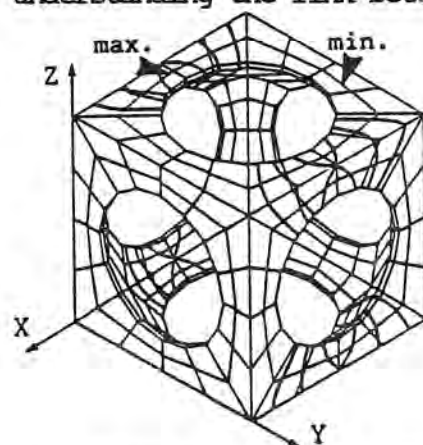


Fig. 1: micro-SED in cell. Stress in X direction.

Fig. 2: Predicted relative strength.

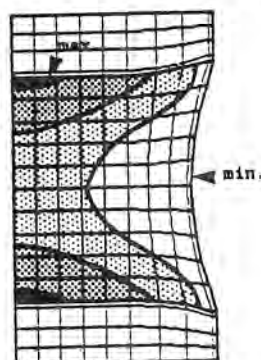
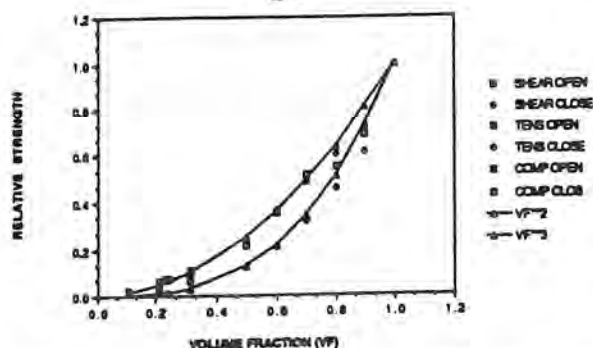


Fig. 3: macro-SED in lumbar vertebral model. White is least value.

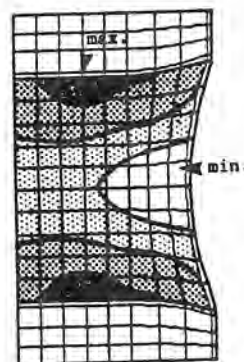


Fig. 4: micro-SED maxima in vertebral model.

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JOURNAL OF BIOMECHANICS

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THE EFFECTS OF SEAT POSITION AND LOWER EXTREMITY KINEMATICS ON AUTOMOBILE BRAKE PEDAL FORCE CAPABILITY

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The purpose of this study was to investigate the relationships between seat position and lower extremity kinematics on automobile braking forces. The braking capabilities of 45 females were tested with a KINCOM isokinetic dynamometer while subjects were seated in a mock-up of a mid-sized car. Subjects were tested in five (5) seat positions of varying distances from the brake pedal, which altered lower extremity joint angles. Testing was done with both a movable pedal (isokinetic mode) and a stationary pedal (isometric mode). Results of this study suggest: 1) there is a significant difference ($p=0.0063$) between forces obtained during static and dynamic testing ($w=30$ deg/s); 2) subjects typically perform best in a position closer to the brake than in their seat position of comfort; 3) there is no optimum lower extremity position for highest braking forces; and 4) automobile design can optimize operator pedal force production by allowing lower velocity pedal travel. In addition, the force distributions suggest that current limitations of braking force requirements are reasonable for the U.S. female driving population, and designed for a geriatric target population, might not necessitate different braking requirements from the standard.

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BIOMECHANICAL ASSESSMENT OF EXERCISE INDUCED MUSCLE DAMAGE.

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INTRODUCTION

Mechanical models of muscle have been used extensively in motor control theory (1,2,3), biomechanics (4), and muscle physiology (5). The mechanical components of muscle are thought to include series elastic (SEC) and parallel elastic components (PEC). In more recent models these components have been associated with active and passive muscle respectively. SEC's are important in the storage of potential energy when an active muscle is stretched (5). A considerable portion of the SEC is thought to be in the actino-myosin cross bridges. PEC's give passive muscle its resistive tension when stretched, and consist of sarcolemma and the connective tissue components. The active behavior of the limbs attached to the elastic muscle system has been modelled after a non-linear mass-spring system (1,2,3,4). The nature of the oscillation depends upon the damping and stiffness parameters of the spring (6).

Damage to SEC's or PEC's following induced soreness may be accompanied by changes in the oscillation as a result of increased mechanical stiffness or damping of these elements. Moreover, since SEC's are thought to be relevant only in active muscle, it is possible to parse out their effects with an experimental paradigm which places passive stretch on non-contractile tissues. The purpose of this study was to determine if delayed onset muscle soreness results in damage to the non-contractile tissues of muscle as measured by changes in the frequency, number of oscillations to equilibrium, and decay of oscillations in the passively swinging limb.

METHODS

Six college age men performed eccentric and concentric contractions of the muscles of the leg by means of a standard step-test. Delayed onset muscle soreness (DOMS) was confirmed 48 hours later by means of blood Ck levels and subjective reports. Prior to the exercise testing session and 48 hours following, criterion measures of EMG amplitude, limb oscillation frequency, number of oscillations, oscillation decay and settling time were recorded. Subjects sat upright in a chair with their upper leg supported to approximately 10cm proximal to the knee. The lower leg was supported and extended to 135 degrees of extension, released and allowed to oscillate until coming to rest. Data were collected from an electrogoniometer located at the lateral epicondyle of the knee and EMG from the rectus femoris and biceps femoris. Five trials pre- and post-induced soreness in which EMG was essentially silent were used for analysis.

RESULTS

To assess the patterns of decay of the oscillations in terms of amplitude, a regression analysis was performed using values for peak amplitudes in each cycle. Beta coefficients for the two conditions were generated for each subject and compared using a paired T-test. Results are presented below.

<u>BETA COEFFICIENTS</u>	<u>PRE</u>	<u>POST</u>	<u>t value</u>
intercept	21.33	20.69	.050
linear	-6.81	-6.91	.020
quadratic	.607	.669	.39

Significant ($p < .01$) linearity was found under both conditions. Differences between conditions were not significant. The interval between peaks was highly consistent within a trial, within a subject, between conditions and between subjects (mean = .26, sd = .0054 seconds), indicating a highly linear damped oscillator system. Frequency, settling time and number of oscillations to equilibrium were compared in the pre- and post- conditions using a repeated measures within subject design (BMDP2V). Overall means and standard deviations () are presented below.

	<u>PRE</u>	<u>POST</u>	<u>F(df=1,5)</u>
Frequency (Hz)	1.962 (.031)	1.967 (.045)	.100
Settling time (sec)	2.65 (.299)	2.64 (.494)	.948
# Oscillations	5.33 (.540)	5.30 (.890)	.893

All comparisons failed significance at $p < .05$.

DISCUSSION

The linearity of the decay in oscillation and the constant interval between peaks support a linear damped oscillator model for the passively swinging leg. This contrasts with the experimental findings which show active limb movements to behave as a mass-spring (1,2,3). It can be concluded that the non-linearity and spring-like behavior of active limbs is in likelihood derived from the change in properties of muscle at different lengths. In the passive paradigm used in this experiment such influences are minimized. The criterion measures of limb frequency, settling time, and number of oscillations to equilibrium did not change between the two conditions. Thus the damping and stiffness parameters of the passive muscle-joint system were unchanged as a result of induced soreness. The PEC's including the epimysium, perimysium and endomysium are largely unaffected when a person has muscle damage. These findings lend support to the notion that the major impact of muscle damage is to the active contractile elements of the muscle (actino-myosin cross bridges). Furthermore, this paradigm and a similar one in which active oscillations are used may provide simple non-invasive measures of muscle damage.

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LEG DOMINANCE AND BIOMECHANICAL SLIP MECHANISMS

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As more statistics become available, there is no doubt that slips and falls are hazards to workers and to older people even when not working. Falls caused 23% of worker's compensation cases in U.S. workplaces [1], but in certain occupations a larger proportion of compensation claims comes from slips and falls. Current demographics indicate a trend toward an older work force, and since the consequences of falls are potentially more severe for older workers [2], slip and fall prevention are more urgently needed.

Two separate mechanisms require investigation: 1) how an unexpected displacement of the shoe/foot complex occurs with respect to the support surface (the slip mechanism), and 2) what body reactions occur as a result of this unexpected displacement (the fall prevention mechanism). The slip mechanism has been studied tribologically by measuring coefficients of friction [3], and biomechanically by observing kinematics and ground reaction forces of the walking human [4,5]. Fall prevention mechanisms have been probed by perturbing walking [6] and by perturbing stance [7]. Motor control factors affect the fall prevention mechanisms, but no previous biomechanical studies of slip mechanisms have included motor control considerations. The purpose of this research was to determine the effects of leg dominance on biomechanical parameters related to the slip mechanism.

METHODS: Four young male volunteers participated in this study. Each subject underwent one session of acclimation and two days of testing. After signing the informed consent document, each subject's height and weight were determined (82.44 $\pm$ 6.06 kg, 182.22 $\pm$ 4.32 cm). Two subjects were left leg dominant (LDM), and two were right leg dominant (RDM). Subjects were instructed to walk at a comfortable pace along a carpeted runway onto a vinyl surface attached to a force plate (AMTI Model SGAG-3) and to the remaining walkway surface. Subjects wore a backpack harness which was attached by a life-line to the ceiling above the force plate to prevent falls. Two slipperiness conditions were used on the force plate landing area: 1) dry vinyl floor, and 2) vinyl floor lubricated with a liberal coating of glycerol. Each subject performed twenty successful trials on each landing surface each day; the conditions were balanced over days. Subjects all wore rubber overshoes and landed on the force plate with the right foot. The first 10 trials were considered practice in each condition; only the final 10 trials are reported here.

Kinematic data were collected in the sagittal plane via a Motion Analysis System (MotionAnalysis Corp., VP-110 video processor, Sun 3/160 workstation) using two cameras (NAC MOS V-14) at their 200 Hz frame rate. One camera was positioned to capture the stride immediately preceding force plate contact, the other collected kinematic information centered on the force plate. Reflective markers were placed on the lateral malleolus, calcaneus, head of fifth metatarsal, greater trochanter, shoulder, elbow, and zygomatic process of the right side. Ground reaction force data were sampled at 200 Hz, synced to the cameras. The criterion measure derived from the kinematic data was the horizontal velocity of the trunk center of gravity (CG). Ground reaction force parameters included the braking and the propelling impulses from the antero-posterior ground reaction force.

RESULTS: All variables were stable over days, and hence were pooled. The order of surface presentation had no effect on velocity. Subjects did not change velocity for the different surfaces:

Mean (+/- S.D.) Velocity of Trunk CG (m/s)

	<u>Dry</u>	<u>Lubricated</u>
LDOM	1.4080 (.2329)	1.4168 (.2460)
RDOM	1.4102 (.3806)	1.4151 (.3805)

Ground reaction force parameters were normalized for body mass:

	<u>Dry</u>	<u>Lubricated</u>
<u>Mean (+/- S.D.) Braking impulse (N*s/kg):</u>		
LDOM	.3106 (.0901)	.1670 (.1795)
RDOM	.3257 (.0543)	.1942 (.0758)

<u>Mean (+/- S.D.) Propelling impulse (N*s/kg):</u>		
LDOM	-.2825 (.0572)	.0146 (.1886)
RDOM	-.2406 (.0314)	-.1123 (.0850)

Both groups had significantly smaller braking impulses on the lubricated surface. The LDOM group had problems propelling on the lubricated surface, as indicated by the positive propelling impulse.

DISCUSSION: Walking has been demonstrated to be a relatively symmetrical activity [8]. However, one manifestation of the biomechanical slip mechanism appears to be its dependence upon leg dominance in lubricated conditions. The LDOM subjects in this study were more susceptible to slips on the right foot. In kicking, the dominant leg is used as the accuracy leg while the non-dominant leg is normally the support leg. This support function was adversely affected when the LDOM subjects landed on the lubricated floor. The greatest implication of this investigation is that biomechanical studies of slip mechanisms must take into account leg laterality, unlike studies of unperturbed gait.

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MODELLING OF THE SHOULDER MECHANISM

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In upper limb motions, the functioning of the shoulder mechanism is of major importance. It consists of clavicle, scapula and humerus, and is connected to the thorax by the sternoclavicular and scapulo-thoracic joints. Movements of the shoulder complex are the sum of synchronous participation of all joint units. Using non-invasive techniques it is not possible to describe the kinematics of the separate segments and muscles. Some attempts have been made to design a physical model of the shoulder (Mollier, 1899; Shiino, 1913; Hvorslev, 1927). Since then the scope of descriptions of the shoulder mechanism has been essentially two-dimensional or has been limited to one joint (Poppen & Walker, 1978). However such models are inadequate to describe or simulate complex movements, as encountered in rehabilitation (e.g. glenohumeral arthrodesis) or movement analysis (e.g. wheelchair propulsion). For studying complex movements a three-dimensional model of the shoulder complex is necessary. This requires a description the three-dimensional orientation of muscle origins and insertions, ligament attachments and (shapes of) joint surfaces. In addition, characteristics of muscles related to force development are needed, preferably based on the same population. However no database on these parameters of the shoulder complex in a three-dimensional orientation is available (with the exception of data of Hogfors et al, 1987). Information on muscle characteristics, involving muscle architecture, indices of length, mass and physiological cross-section (PCSA) is scattered and incomplete. It was our aim to initiate a database on the geometry and properties of the shoulder complex. In the current study data on muscle mass and PCSA of muscles of the shoulder complex are presented.

Methods

To collect data described above, a cadaver study was performed focussing on the collection of three-dimensional spatial and muscle characteristics, to be implemented in a biomechanical model of the shoulder complex. Of seven embalmed cadavers (5 ♂, 2 ♀, age 80.0 $\pm$ 7.0, height 171.1 $\pm$ 6.92 cm, mass 76.1 $\pm$ 16.43 kg) anthropometric data were collected and both shoulders measured. Estimates of segment parameters (volume, mass, mass position, moments of inertia) were calculated using the regression equations of Clauser et al. (1969). Relevant structures were measured in a three-dimensional orientation with a 'palpator' (Pronk & van der Helm, 1989), and individual muscles were dissected and weighed. PCSA was determined by measuring the muscle surface of a cross-section perpendicular to the muscle bundle orientation.

Results

Anthropometric and estimated segment parameters of the cadavers agreed reasonably well with data of Clauser et al. (1969). PCSA data (Figure 1) were generally lower than those published earlier (Shiino, 1913; Poppen & Walker, 1978), but agreed with indirect measurements (Basset, 1983), based on muscle volume and bundle length. Muscle mass (Figure 2) correlated well with PCSA ($P < 0.01$), but generally no statistical differences between left and right were found. Comparison with earlier publications indicate that our results showed generally lower values. However a good correlation was generally found between present results and data of earlier publications (obtained from younger subjects).

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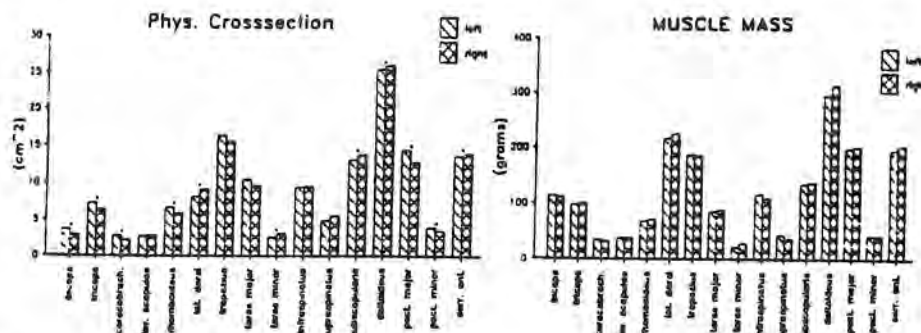
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LEFT: Figure 1. Physiological cross-sections of muscles. The * indicates a significant correlation with muscle mass ($P < 0.01$; m.triceps only the long head).
RIGHT: Figure 2. Mass of muscles. Of m.triceps only the long head was measured.

PROPERTIES OF TENDINOUS STRUCTURES OF RAT EDL MUSCLE-TENDON COMPLEX

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Introduction

The series elastic component (SEC) of skeletal muscle consists of a part located in the cross-bridges and of tendinous structures. Within the tendinous structures, tendon and aponeurosis (i.e. tendon plate on the surface of the muscle belly) can be distinguished. In this study an attempt was made to compare elastic properties of the tendinous structures with the properties of the entire SEC.

Methods

The EDL muscle-tendon complex of 5 anaesthetized male Wistar rats were prepared in situ. Compliance of SEC was determined at different force levels by imposing quick length decreases (0.2 mm within 3 ms) during isometric tetanic contractions at different muscle lengths. Force-extension relationship of SEC was determined by means of integration of the force-compliance data with respect to force. Length of tendinous structures was measured before and during contraction by means of macro-photography. SEC extension, length differences of tendinous structures in active muscle, from near-slack muscle length to optimum muscle length (l_0), and extension of tendinous structures during a contraction at optimum length were compared.

Results

In figure 1 extension of SEC, and length of tendon and aponeurosis with respect to force are shown. It appears that the force-length relationship of the tendon is similar for active and for passive muscle. The data point of the tendon in passive muscle at l_0 (circle) lies on the same curve describing the data points in the active condition. Tendon extension at maximal isometric force is 3.1 ± 1.0 %. Extension of aponeurosis at l_0 from passive to active condition is 2.3 ± 0.6 %. Extension of all tendinous structures at l_0 is less than the calculated SEC extension, which is expected since part of SEC is located in the cross-bridges. However, if the data of the aponeurosis in active muscle are considered, a complete different length-force relationship of aponeurosis is obtained (Fig. 1B). In that case extension of the aponeurosis itself exceeds extension of entire SEC.

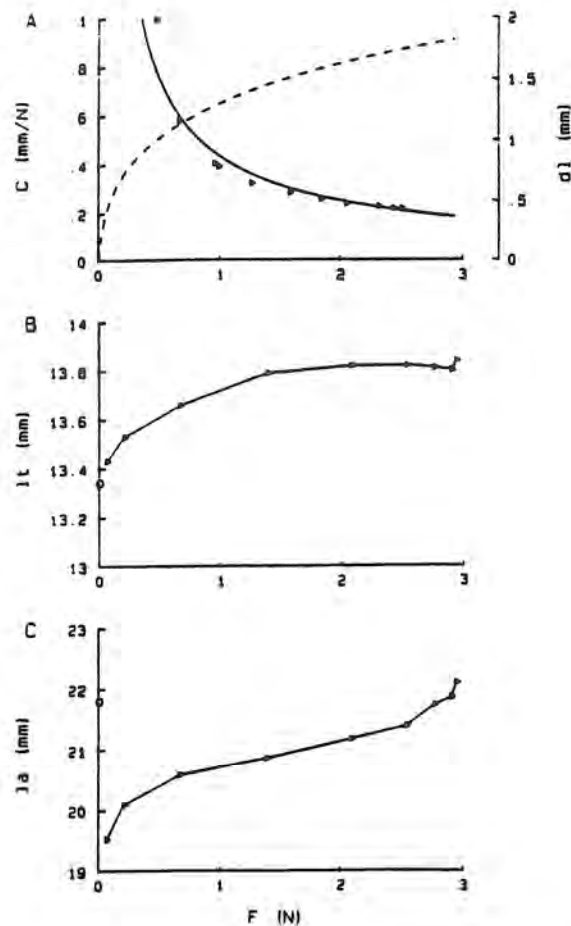


Fig. 1 Compliance (solid line) and extension (dashed line) of SEC (A), and length of tendon (B) and aponeurosis (C) as a function of force. Triangles represent data for active muscle. Circles (B and C) represent lengths in passive muscle at l_0 .

Discussion

It is concluded that aponeurosis force-length relationship is not a unique one, but is influenced by muscle length. It is proposed that this influence concerns a shift of the force-length relationship of aponeurosis along the length axis, leaving the slope of the curve (i.e. compliance) unaltered. This means that compliance is only force dependent. This way it can be explained that compliance measurements are in agreement with extension of the tendinous structures at l_0 , even though compliance was measured at different muscle (and thus aponeurosis) lengths.

The characteristics found for aponeurosis may effect muscular energetics, by means of elastic energy storage and working range of the muscle fibres in dynamic contractions.

Influence of muscle geometry on force-velocity characteristics of gastrocnemius medialis of the rat.

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Introduction

It is generally accepted that the force-velocity relation exhibits the same characteristics over a wide range of different muscles. Relatively little work has been done comparing force-velocity curves for different muscle lengths. Especially in more pennate muscle, it is expected that muscle geometry has a considerable effect on force production given a shortening velocity of the muscle-tendon complex. If this holds true, the force-velocity curve determined at muscle optimum length (l_0) can not simply be corrected for effects of muscle length to give force-velocity curves for lower muscle lengths.

Methods

The muscle-tendon complex of unipennate gastrocnemius medialis (GM) was exposed in situ in three male Wistar rats. Using a muscle ergometer the muscle force-length curve was determined isometrically. During the experiment, l_0 was measured using a pair of compasses. Subsequently, force-velocity characteristics were determined at three different muscle lengths (circa 78%, 85% and 100% l_0). After isometric tetanic activity the muscle-tendon complex was released 5 mm and shortened with constant velocity. After 2.5 mm shortening, force and velocity were calculated from their continuously registered records. This was done for velocities ranging from 5 mm/s to 110 mm/s. Total force was corrected for passive force at appropriate muscle length, yielding active muscle force (F_{ma}). Curves for muscle force against shortening speed of origin and insertion (V_{oi}) were plotted. To correct for the influence of muscle length on force production, the actual force measured at each velocity was normalized with respect to the isometric force at that muscle length (relative force production). To correct for the amount of shortening of the series elastic components (SEC), a previously determined force-compliance relationship for GM was used (unpublished observation). Shortening velocity of the contractile element (V_{ce} ; $V_{ce} = V_{oi} - V_{sec}$) in the direction of the line of pull of the muscle was calculated.

Results

For all three rats similar absolute results were obtained. At each velocity force produced was highest for l_0 and decreased with lower muscle length (fig.1). However, if force was normalized with respect to the isometric force at that length, the force-velocity curves show an inverse order (fig.2). For any velocity the muscle was able to produce relatively more force at low muscle length than at high muscle length. That this effect was not caused by an increasing amount of shortening of the SEC at low muscle length, is shown in figure 3 as a similar order is observed as in figure 2.

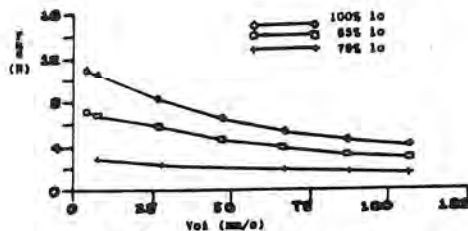


Fig. 1. Force-velocity curve:
Voi is velocity of origin and insertion.

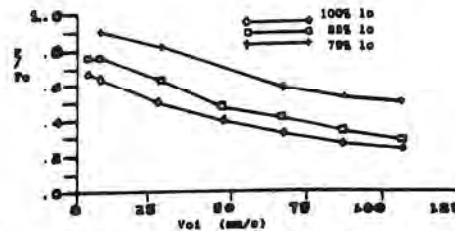


Fig. 2. Relative force-velocity curve.

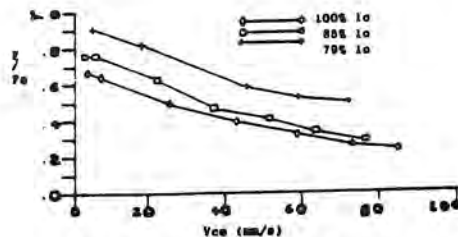


Fig. 3. Relative force-velocity curve:
Vce is velocity of contractile element.

Discussion

It is suggested that the higher relative force production at lower muscle length is an effect of geometry. At low muscle length the angle between the fiber and the line of pull of the muscle can increase considerably. At lower muscle length the increase of fiber angle accounts for a large part of the muscle shortening. At lower muscle length the contribution of fiber shortening to muscle shortening is therefore less than at higher muscle length. Therefore, for a given shortening velocity of the muscle-tendon complex, the fiber shortening velocity decreases with decreasing muscle length. This results in a higher relative force production of the fiber, and consequently the muscle, at lower muscle lengths.

APONEUROSIS CHARACTERISTICS OF RAT GASTROCNEMIUS MUSCLE.

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Introduction

As a muscle tendon complex is made to contract isometrically, muscle length will change as a consequence of increases of tendon length. Within the muscle belly fiber length and angle as well as aponeurosis length and angle will increase. The effects of changes of fiber length and angle will have to be equal to the sum of muscle length changes and effects of changes of aponeurosis length and angle. The difference between muscle length changes and length changes of the projection of fiber length on the line of pull of the muscle should reflect changes related to aponeurosis. These changes will be compared to the aponeurosis length force characteristics for different age groups of rats.

Methods

The right gastrocnemius medialis muscle of 6 male Wistar rats of each of three groups (10, 14 and 16 weeks old) were used in isometric length-force measurements. With the help of data obtained using markers in the muscle and macro-photography, muscle geometry was reconstructed for the passive and fully active state. The length change of the projection of fiber length on the muscle line of pull was calculated and compared with the change of muscle length for both the 10 and the 16 weeks groups. Length-force characteristics of the proximal aponeurosis were determined, by plotting total muscle force versus aponeurosis length, and compared to that of the muscle. Energy storage in aponeurosis is scaled with respect to maximal work of the muscle (i.e. area below muscle length force curve).

Results

Typical examples of differences between projected fiber lengths in the passive and active state, as well as changes of muscle length are shown in Fig. 1. For a given muscle length the vertical difference between these variables is due to the lengthening of the projection of the aponeurosis on the line of pull. In both groups this lengthening is smallest above 95 % of muscle optimum length (l₀). At short muscle lengths the absolute value of lengths changes of aponeurosis projected length was almost equal to fiber projection length changes indicating that changes related to the aponeurosis predominated those related to tendon.

From Fig. 2 it is concluded that normalized length-force characteristics of aponeurosis in the active muscle did not differ significantly for the age groups. At muscle active slack length aponeuroses were approximately 10 % shorter than at l₀. No difference could be shown between passive length force curves of the age groups (Fig. 2). However passive and active curves differed substantially: For a given low level of force much longer aponeuroses were found in the passive than in the active muscle.

As a consequence of aging, length of the aponeurosis increased from 21.67 ± 1.05 mm to 25.51 ± 1.12 mm (i.e. 15 %). Energy stored in the aponeurosis increased (from 8.51 to 14.89 mJ) because absolute length range (i.e. l₀ to active slack length) and force increased during aging (from 2.17 to 2.31 mm and from 10.76 to 14.13 N respectively), while maximal work of the muscle increased (from 67.28 to 72.80 mJ).

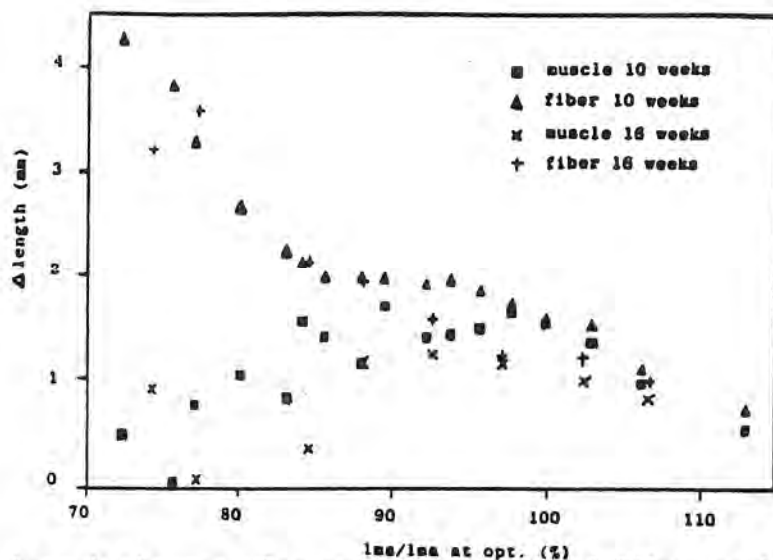


Figure 1. Length changes of muscle (■ and x) and fiber projected length (▲ and +) between passive and active states, plotted against normalized active muscle length.

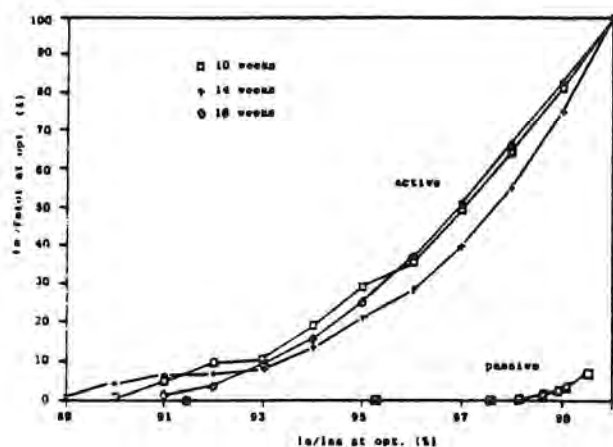


Figure 2. Normalized total muscle force plotted against normalized aponeurosis length in passive and active muscle.

Conclusions

For isometric contraction of the muscle tendon complex, increasing differences between fiber projection length changes and muscle length changes were found between the passive and active state. This indicates a relatively increased contribution, of aponeurosis compared to tendon, to geometrical changes taking place in the muscle at short muscle lengths. Among factors to be considered as possible contributors to this difference are: 1. Differences of aponeurosis length force characteristics in passive and active muscle (shown to occur, Fig. 2), 2. different slack behavior of tendon and aponeurosis.

As normalized aponeurosis length force characteristics in active muscle were unchanged during aging, aponeurosis intrinsic properties did not change. Increased energy storage in the aponeurosis does have implications for work the fiber has to perform in an isometric contraction.

ANKLE ARTHRODESIS
Kinematic and Kinetic Effects in Locomotion

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A problem in some areas of gait analysis lies in delineating cause and effect in an abnormal gait pattern where more than one joint or muscle group is affected. It is difficult to decide which deviations from normal are the primary result of the pathology, and which alterations are a secondary effect. Although a number of authors have attempted to determine cause and effect in 'multi-joint pathologies', their conclusions can only be considered tentative. Another problem stemming from pathologies affecting multiple joints is to determine a sequelae of effects on the locomotor system. For example, if limitation of ankle motion results in a compensation at the knee joint, does that predispose the knee joint to possible injury at a later stage?

We examined the sagittal plane kinematic and kinetic characteristics of 5 subjects, each with a fully united unilateral ankle arthrodesis and no other gait limitation. Each subject was initially examined by an orthopaedic surgeon to confirm the above requirements, and suitable subjects then participated in a session during which locomotion data were collected. Subjects were photographed with high speed 16 mm cameras while walking at their preferred pace across a hidden force platform. A minimum of three trials with ground reaction force (GRF) data from their stance on the pathological limb were recorded. Data from the film were re-constructed using the Direct Linear Transformation algorithm (Marzan and Karara, 1975). Joint and segment kinematics were also calculated for the sagittal plane motion, and with the inclusion of GRF and anthropometric data, joint kinetics were determined. Time and mass normalised averaged data were compared to Winter (1987).

A noticeable kinematic result was the similarity to normal of the angular displacement-time pattern formed between the knee, ankle and head of the first metatarsal joint (the 'ankle joint'). This indicated that considerable dorsi- and plantar-flexion was taking place in the small bones of the foot. The only alteration to a normal pattern was a lack of plantar flexion immediately after foot strike. Subsequent "dorsi-flexion" was achieved from flattening of the arch of the foot.

Compensation for ankle arthrodesis was noted in the ipsilateral knee and hip joint displacements, primarily at heel strike and prior to toe-off. As already noted by Mazur et al. (1979), the contralateral ankle joint appeared to mimic the displacement pattern of the arthrodesed limb, and a greater range of motion than normal was seen at the contralateral hip joint.

Calculation of the moment necessary to produce the motion seen between the foot and shank on the fused limb showed a pattern virtually identical to normal. The knee joint displayed a similar pattern to normal, although the extensor resultant

joint moments (RJMs) after heel strike were reduced, and a greater flexor impulse produced. The hip joint RJM was similar to normal for 3 subjects, but the 4th subject produced large hip flexor moments during stance.

We noted that minor modifications to the displacement pattern of the contralateral ankle joint were made to achieve bilateral symmetry. However, adaptation to unilateral ankle arthrodesis was achieved primarily by increasing the movements required in the small bones of the ipsilateral foot. Minimal changes were seen in the resultant joint moment patterns at ankle, knee and hip, or in the contralateral leg.

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3D Muscle Activation Surfaces - Linear Relationships

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Most torque/force-angular velocity-angular position muscle activation surfaces, or torque/force-velocity graphs have shown characteristics consistent with theoretical muscle mechanics. However, since most data have been derived from muscle groups consisting of several muscles with different resting lengths, architecture and mechanical advantages which act at a mechanically complex joint, this agreement may be coincidental.

We examined the changes which resulted in the three dimensional (3D) angular muscle activation surface from inclusion of patellar moment arm and excursion information. Thus, torque-angular velocity-angular position data were transformed to linear force-velocity-length data.

Isokinetic angular data for maximal knee extension trials at 11 speeds were collected on six volunteer subjects using a BIODEX dynamometer, and the results presented as 3D surfaces. Radiographs were made of each subject's knee joint in seven positions from 90 degrees to full extension, with quadriceps contracted. Displacement of the superior aspect of the patella, and the instantaneous center of rotation of the joint (using the method of Frankel and Burstein, 1971) were determined. Patellar excursion and moment arms were then calculated for movement from 90 degrees of flexion to full extension. The values calculated for patellar displacement and moment arms agreed well with literature from studies using the same technique. These data were averaged across subjects, and expressed as functions of the angular position of the knee. They were then used to transform the angular data to their linear equivalents. 3D surfaces of the force-velocity-length data were drawn. Figure 1 shows the results from the averaged data.

The linear 3D surface displays the force-velocity and tension-length characteristics in general, but at moderate speeds shows less variation in muscle force with changes in length than for the equivalent angular surface. This plateauing has been noted previously by Fuglevand (1987). A number of factors could contribute to these results. Several authors have suggested antagonist activity or neural inhibition may be involved in the plateauing. We did not record electromyographic activity, and so cannot comment on either of these suggestions.

Another possible reason for the flatness of the surface may be the effects of the contributions from the individual muscles of quadriceps. Each muscle may have an optimum length and mechanical advantage at a knee flexion position which differs from the other knee extensors. The effect of this would be to decrease the peak force produced, but to minimise the force decrement over a larger range of motion.

Peak force was achieved earlier in the range of motion than was evident in the angular 3D surface. Also in contrast to the angular data, peak linear force was generated at the slowest speed.

Our study has looked at the effects of two aspects of knee mechanics on the shape of muscle activation surfaces. These results suggest that the agreement frequently found for torque/force-velocity relationships and theoretical muscle mechanics is likely to be coincidental.

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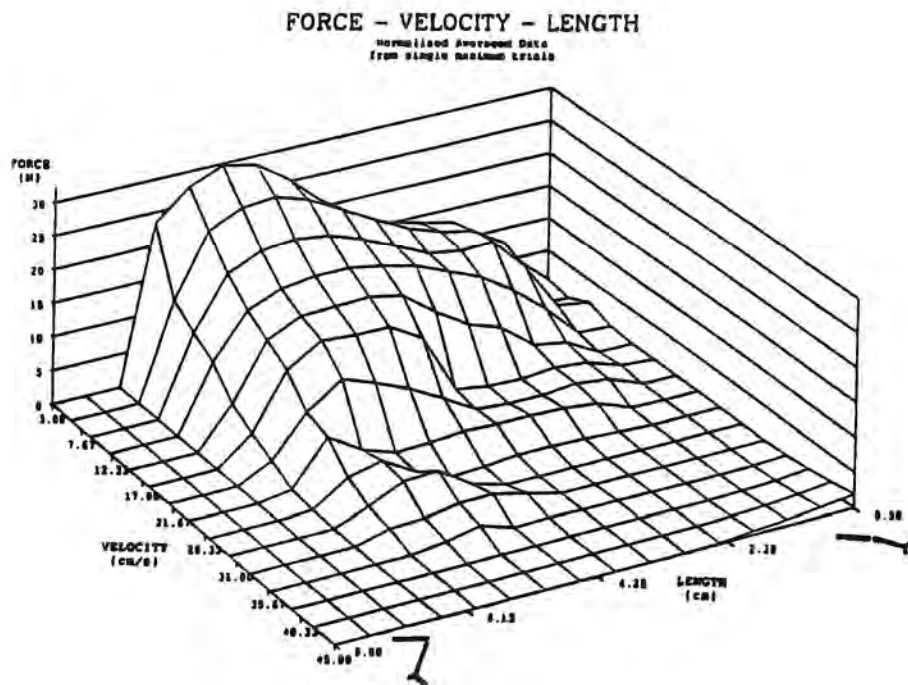


FIGURE 1

3D MUSCLE ACTIVATION SURFACES - ANGULAR RELATIONSHIPS

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Most isokinetic data on maximum muscle contractions have reported either torque/force at a specific angle, or peak torque/force regardless of position, for a variety of speeds. In either case the data represent discrete points on the three dimensional (3D) force-velocity-length muscle activation surface and provide little information about the actual shape of the entire surface.

A number of studies have shown differences between 'in-vivo' and 'in-vitro' muscle strength patterns; thus the 3D graphs of 'in-vitro' muscle preparations may not be representative of human 'in-vivo' contractions. Fuglevand (1987) presented a muscle activation surface for human knee extensors, derived from Cybex II data, and showed movement of the location of the peak torque toward full extension, as well as plateauing of the torques with increased speed of movement. Neither of these results is compatible with traditional length-tension theory. This study gave us useful information, but the curves and surface provided epitomise the oversmoothing characteristic of Cybex II data.

The intention of our study was to generate a 3D surface of muscle activation for knee extensors. Six volunteer subjects were tested at 11 randomly assigned speeds (30, 60, 75, 120, 165, 210, 255, 300, 330, 400 and 450 degrees/second) on a BIODEX isokinetic dynamometer. Testing took place over two days, at the same time of day. After a warm-up, each subject performed six maximum quadriceps extension trials at each speed, and the last three trials were recorded at 100 Hz. Each subject determined their flexion speed, up to a maximum of 450 degrees/second. For each speed, the trial with the maximum extensor torque was used for subsequent analysis. Data for each subject were windowed to remove torques not generated within 5% of the isokinetic speed. The 3D graphics facilities of SAS (SAS Institute Inc.) were then used to plot muscle activation surfaces for each subject (see Figure 1). Each subject's data were normalised to their maximum torque, and these results were averaged and plotted.

The 3D surfaces illustrated the general characteristics of force-velocity and tension-length relationships. Peak torque was not always recorded at 30 degrees/second. The location of the peak torque moved toward extension (ie. from 65 to 54 degrees), with increases of speed up to 255 degrees/second. At speeds higher than this, peak torques were found at greater angles of flexion (68 to 77 degrees).

In general, subjects achieved the target velocity before reaching 90 degrees. With increases in speed, they did not maintain target velocities for equal ranges of movement, and the surface shows that torque generation decreases rapidly from full extension with increases in speed. This may be due to an inability to sustain the speeds, or it may be indicative of the need for earlier 'braking' prior to full extension.

Three of the six subjects were unable to achieve speeds of 400 or 450 degrees/second. However, those who did displayed consistent torque production over a larger range of motion than at 300 or 330 degrees/second. Two reasons are suggested for this anomaly. At higher speeds the quadriceps may act eccentrically to slow and reverse the flexion movement, thus producing larger torques and speed than would be expected. Also, increased extensor speeds may produce sympathetically mediated increases in flexor speeds, with the result that an increase in pre-stretch speed is experienced by the extensors.

Fuglevand, A.J. (1987) Resultant muscle torque, angular velocity and joint angle relationships and activation patterns in maximal knee extension. Biomechanics X-A (ed. B. Jonsson). Human Kinetics Publishers, Illinois. pp. 559-566.

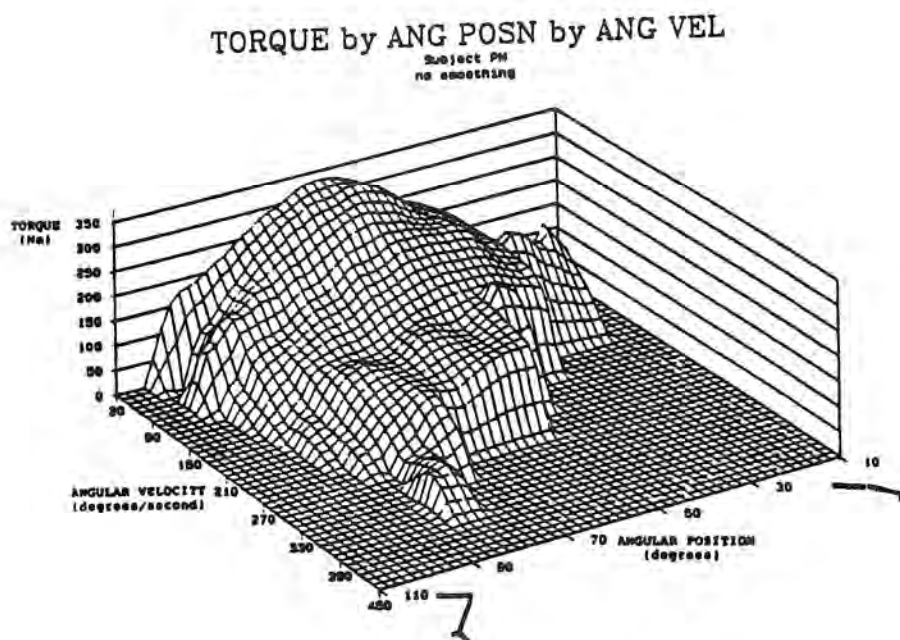


FIGURE 1

THE GAIT OF SUBJECTS WITH ANTERIOR CRUCIATE LIGAMENT DEFICIENCY.

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Injury to the anterior cruciate ligament (ACL) of the knee has been of interest to the orthopaedic community for many years. While considerable research has been conducted to compare the functional disability associated with this ligament injury to the results of a physical examination, few studies have examined the gait of subjects with this knee condition. Evaluation of the gait of subjects with an ACL deficient knee may provide useful information to surgeons and clinicians concerning the compensatory techniques used and provide insights into factors which are important in ACL deficient gait.

Ten subjects with unilateral ACL deficiency participated in this study. Each subject answered an extensive questionnaire outlining their knee injury and their return to sports activity. Subjects also underwent a physical examination by an orthopaedic surgeon. A inter-limb comparison was then undertaken in several areas. Isokinetic muscle testing was carried out at joint angular velocities of 60, 180 and 300 deg/sec using a BIODEX dynamometer. Each subject was also videotaped at 200 Hz while running on a treadmill at a recreational jogging pace. Kinematic analysis determined the angle of knee flexion at footstrike and the nett amount of stance phase flexion. EMG data were collected from the vasti lateralis and biceps femoris muscles during running. EMG data were demeaned, high pass filtered (10Hz) to remove movement artifact, rectified about their zero line, and normalised to a submaximal anti-gravity contraction.

The questionnaire results showed ACL deficiency was associated with considerable functional disability, particularly during sports which involved pivoting, sudden stops and jumping activities. While subjects had decreased their level of sports participation they were still playing at levels where they experienced "giving way" episodes periodically.

At the physical examination seven subjects were noted to have effusion of the knee joint. All subjects had a positive pivot shift test indicative of anterolateral instability. No subjects had anteromedial instability.

Isokinetic muscle testing revealed a significant ($p < 0.05$) quadriceps deficit in the ACL deficient limb. The presence of effusion in seven subjects may explain some of the quadriceps deficiency noted, as Kennedy et

al.(1982) have shown that a knee joint effusion as small as 60 mls to lead to 30 to 50 percent decreases in reflex evoked quadriceps contraction. An insufficient rehabilitation program following injury could also account for the observed quadriceps muscle deficiency.

Normalised EMG profiles showed increased quadriceps activity during a 200 ms period centred about footstrike. This increased activity would not at first appear to be of benefit to subjects with ACL deficiency. As the ACL is the primary restraint to anterior displacement of the tibia on the femur excessive repetitive quadriceps activity in the ACL deficient limb would place increased stress on the secondary ligamentous structures. Furthermore, increasing the anterior displacement of the tibia on the femur would predispose the knee to giving way. However, the observed EMG activity of vasti lateralis may not be excessive particularly in light of the observed strength deficits in the quadriceps muscles. It is plausible that the quadriceps muscles of the ACL deficient limb may contract at a greater percentage of their maximum to generate tension similar to that of the uninvolved limb in an effort to maintain bilateral kinematic symmetry. In doing so increased EMG would be observed in the quadriceps of the ACL deficient limb.

It was concluded that even during straight line running, an activity not usually thought to affect individuals with ACL deficiency, marked compensatory techniques may be observed.

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MUSCLE TORQUE AND VELOCITY RELATIONSHIPS IN ELITE
POWER AND ENDURANCE ATHLETES.

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At present little information is available concerning dynamic force characteristics of athletes. This project aimed at providing quadriceps force-velocity data for power and endurance athletes using a BIODEX dynamometer.

Elite male power ($N = 6$) and endurance athletes ($N = 7$) were selected to participate in the project (mean age = 23.6 years, $SD = 3.0$). Selection criteria was based upon achieving at least provincial representation in one of the following sports: sprint running, olympic and power lifting, volleyball, distance running and cycling. Subjects performed maximal, preferred leg, quadriceps contractions at each of twelve angular velocities: 30, 60, 90, 105, 120, 135, 150, 180, 210, 240, 285, 300°/s.

The experiment was undertaken using a BIODEX dynamometer and software (BIODEX Corporation) to control angular velocity and measure torque. Testing for each subject took place at the same time of day over three days. Following a training session (day 1), each subject performed six repetitions at each of the randomly assigned angular velocities (days 2 and 3). Three minutes rest was permitted between each set of contractions.

Torque measurement was calibrated using known masses attached to the lever arm. Torque was independently validated against that obtained in another laboratory using a dynamometer with force transducers attached to an isokinetically controlled lever arm.

Data were averaged for the last five contractions and treated as absolute torque and torque normalized to body mass. Data were analysed using Hotelling's T-squared analysis to isolate differences between athlete groups at each of the angular velocities, and analysis of covariance to study differences between the slopes and intercepts of power-velocity relationships.

Data from both groups were first pooled for analysis. When referenced to full extension (0°), the angle at peak torque decreased as a linear function of angular velocity increment. Peak torque occurred at a mean angle of 59.5° ($SEM = 0.8$) in agreement with Thorstensson *et al.* (1976).

Peak torque decreased curvilinearly with increased angular velocity. From 30-300°/s this decrement averaged 83 Nm, or 40.6% of the absolute torque obtained at 30°/s. Normalized torque fell from 2.60 to 1.53 Nm/kg, or 41.2%. The magnitude of these changes is quantitatively similar to the

observations of Perrine and Edgerton (1978) and Tesch (unpublished data), but is less than reported by Thorstensson et al. (1976), Lesmes et al. (1978), Coyle et al. (1979), and Wyatt and Edwards (1981).

Clear separation occurred between absolute and relative torque-angular velocity curves for both subject groups, with power athletes generating greater torque at each angular velocity ($p < 0.05$), as reported by Johansson et al. (1987). Proportional decrement in peak torque with angular velocity increase was approximately equivalent for both subject groups.

Instantaneous power increased as a linear function of angular velocity for both power ($r^2 = 0.839$) and endurance groups ($r^2 = 0.728$), in close agreement with the power changes of runners reported by Johansson et al. (1987). The slopes of these lines diverged significantly between the groups, with increments in angular velocity ($p < 0.05$).

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SEGMENTAL CONTRIBUTIONS TO TOTAL BODY MOTION IN SIT-TO-STAND

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INTRODUCTION

Although the sit-to-stand (STS) movement requires the transfer of the body mass in both the horizontal and vertical directions, we have recently reported [1] the directionally specific differences with respect to the motion of the body center of mass (CM). Specifically, an increase in the speed of ascent was accompanied by a disproportionately smaller increase in the magnitude of the maximum linear momentum of the body CM in the horizontal direction than its vertical counterpart. Additionally, the time to reach the maximum linear momentum was relatively constant in the horizontal direction compared to that in the vertical direction. The linear momentum of the CM was initially equal to zero. Therefore, its maximum value was equal to the propulsive impulse required to accomplish the STS. Moreover, the time to reach the maximum linear momentum was the time when the propulsive impulse became the braking impulse. It was thus postulated that such a relative invariance in the horizontal motion was associated with a neuro-muscular control strategy that may simplify the process of balance control in the horizontal direction from the total task of standing from a seated position.

Since the motion of the CM is dependent on the motion of the segments, such differences in the control over the horizontal and vertical motions of the CM must be reflected in the motion of the individual segments. The purpose of this study was to examine whether the directionally specific differences existed in the contributions of the segments to the motion of the CM during the STS.

METHODS

Ten healthy adult subjects each performed five trials of the STS at a natural speed. A four segment model including foot, shank, thigh and head-arms-trunk (HAT) segments was employed. A motion analysis system (Watsmart) was used to locate the positions of the landmarks for these segments, and this information was used to compute the linear momentum of the CM based on the segmental parameters [2] and the following equations:

$$m_g V_g = \sum (m_i V_i) \quad (i=1,4) \quad (1)$$

The non-zero terms on the right-side of the Eq (1) can be written:

$$m_s V_s = m_s \Phi_s \times r_s$$

$$m_t V_t = m_t (\Phi_t \times r_t + \Phi_s \times l_s)$$

$$m_h V_h = m_h (\Phi_h \times r_h + \Phi_t \times l_t + \Phi_s \times l_s)$$

Thus, the linear momentum contributed by the i th segment (L_i) is defined based on the rotational motion of this segment:

$$L_s = \Phi_s \times (m_s r_s + m_t l_s + m_h l_s) \quad (2)$$

$$L_t = \Phi_t \times (m_t r_t + m_h l_t) \quad (3)$$

$$L_h = \Phi_h \times m_h r_h \quad (4)$$

Finally, $m_g V_g = L_s + L_t + L_h$

Where m is the mass; l and r , the position vectors of the proximal joint center and the CM of the segment with respect to the distal joint center; V , and Φ , the linear and angular velocities. Subscripts g , s , t and h indicate the body CM, the shank, thigh and HAT segments, respectively; the

bold characters, the vector quantities. Both the horizontal and vertical components of the Eqs (2)-(4) were computed.

RESULTS AND DISCUSSION

In the horizontal direction, L_h ($21 \pm 5 \text{ kg m.s}^{-1}$) contributed 64% of the maximum linear momentum of the CM ($33 \pm 6 \text{ kg m.s}^{-1}$). In contrast, in the vertical direction, L_h ($4 \pm 2 \text{ kg m.s}^{-1}$) was only 9% of the maximum linear momentum of the CM ($44 \pm 6 \text{ kg m.s}^{-1}$) while the contribution of the L_t ($34 \pm 7 \text{ kg m.s}^{-1}$) was as great as 77%.

Although the thigh segment had a dominant influence on the momentum-time history of the CM in the vertical direction (Fig 1), the motion of this segment tended to increase the forward motion of the CM while this forward motion was reduced to bring the body to a stop upon the completion of the STS (Fig 2). Consequently, the overall reduction in the forward motion of the CM was accomplished by the combined contributions from the shank and the HAT. Since the location of the CM was confined within a region of few centimeters along the anteroposterior direction upon the completion of the STS, the control over the counteractions related to the thigh and to the HAT and shank during the braking phase were critical to the control of balance in this task.

In conclusion, the findings indicated the existence of the directionally specific contributions of the individual segments in the STS, which would support the postulation that different neuro-muscular control strategies were employed with respect to the horizontal and vertical motions the CM. It remains to be seen, however, whether a progressive variation in the speed of ascent is mainly accomplished by regulating the motion of the segments that impart dominant input to the total body motion in the horizontal and vertical directions.

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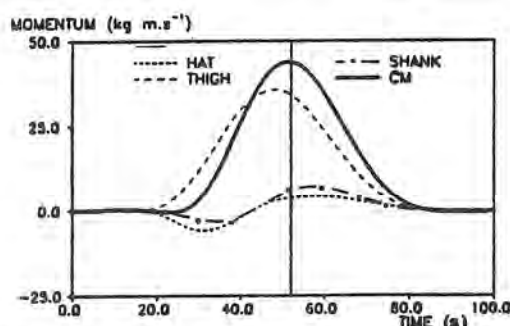


Fig. 1. The averaged vertical linear momentum of the CM and the contributions from shank, thigh and HAT segments in the STS for ten subjects. Time was normalized as a percentage of the total duration. The time to reach maximum linear momentum of the CM is indicated.

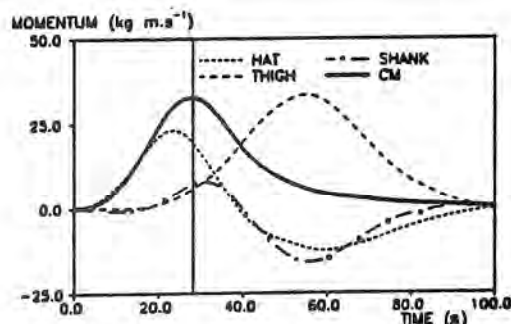


Fig. 2. The averaged horizontal linear momentum of the CM and the contributions from shank, thigh and HAT segments in the STS for ten subjects. Time was normalized as a percentage of the total duration. The time to reach maximum linear momentum of the CM is indicated.

On the Stationary Region of Fluid Resistance
Curve in Swimming

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ABSTRACT This paper investigates the varying properties of resistance for an athlete and his body-like model swimming with fully extending and floating forms by dragging experiments in a water basin. Results show that there exist a certain velocity region within which resistance increases slowly, which is practically meaningful for middle and long distance swimming athletes.

In order to investigate the characteristics of fluid resistance in swimming, an athlete and his body-like model (the angles of neck, hip, knee, shoulder and elbow can be adjusted arbitrarily.) in a fully extending and floating form are tested by dragging experiments in a water basin. For floating forms with different head lifting angles, hip angles and knee angles, 8 kinds of dragging speeds (0.8--2.1 m/s) are chosen. Over 30 Resistance-Velocity (R--V) varying curves are measured, and comparison study is done for resultant R--V curves from model test and athletic test.

In fluid dynamics, the general mathematical expression for resistance is given as follows,

$$R = 1/2 \cdot \rho \cdot S \cdot C_d \cdot U^2$$

Usually, we just roughly regard the resistance as being proportional to velocity square. However, the above variables ρ , C_d , U are not constant except water density in swimming. As a human body swims in water, the total resistance consists of shape, frictional and wave-making (include wave-breaking) resistance. The proportions of above components are studied by the authors. As swimming velocity increases, in ordinary case it is shown that all kinds of resistances become greater, but their proportions in

the total resistance vary with velocity. The proportions of shape and frictional resistances are smaller at higher speed, and the proportion of wave-making resistance becomes greater. In tests of human body-like model, both the absolute value and proportion of wave-making resistance decreases apparently when dragging speed increases up to 1.6--1.7 m/s, the optimal efficiency is achieved, this is "the Stationary Region of Resistance".

The appearance of stationary region is a result of reduction of wave-making resistance. It is concerned with the characteristic length of a moving body. Thus for different athletic body shapes and swimming forms, the optimal velocity extents occurring stationary region are not the same. As an athlete swims in water, the cause of appearing stationary region is that the wave series produced by head and shoulder interferes profitably with that by hip and two legs, i.e., for two wave series caused by former and latter parts of human body, the wave crest and trough superpose with each other, which makes the total wave-making resistance smaller.

Experiments show that the stationary region of resistance exists objectively for human body floating on water and swimming with increasing velocity. How to make use of stationary region effect of resistance? This needs detailed tests and researches for each athlete and his characteristic swimming forms and moving techniques, thus the corresponding optimal velocity extents of stationary region should be determined. Because the stationary region is associated with head lifting angle and hip angle when floating on water, it's no doubt that the theory of stationary region of resistance is important for finding optimal swimming technical plan and for developing bearability at optimal velocity, which is practically meaningful for middle and long distance swimming athletes.

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A MODEL OF ANTERIOR CRUCIATE LIGAMENT INJURY IN ALPINE SKIING

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INTRODUCTION

Knee ligament injuries are the most prevalent alpine ski-related injury and have consistently been reported as 20 to 25 per cent of all skiing injuries (e.g. Johnson et al., 1979; Howe and Johnson, 1982; Bechtel et al., 1984 and Weaver et al., 1987). However, during the last decade the nature of these injuries has changed. There has recently been a dramatic increase in the incidence of anterior cruciate ligament (ACL) injuries (Howe and Johnson, 1982; Bechtel et al., 1984; McConkey, 1986; Feagin et al., 1987 and Hauser and Schaff, 1987). These ACL injuries appear to occur most often in advanced and competitive skiers and seem to be associated with changes in equipment that have occurred over recent years such as higher, stiffer ski boots (Bechtel et al., Hauser and Schaff, 1987). One of the particular mechanisms of ACL injury that has been reported occurs when a skier lands off a bump in a sitting back position, and then tries to recover (McConkey, 1986 and Feagin, 1987). It has been speculated that a violent contraction of the quadriceps muscles is instrumental to injury of the ACL (McConkey, 1986 and Feagin et al., 1987). Furthermore, it has been suggested that increased strength of the hamstrings muscle group may aid in prevention of the injury (Feagin, 1979; Feagin et al., 1987).

The purpose of this study was to simulate the force response of the ACL to a variety of different load sharing assumptions using a two dimensional model of the knee joint and selected force carrying structures crossing the knee. These assumptions were specifically related to the amount of activity and geometry of the acting quadriceps and hamstrings muscle forces. The model was applied to a specific movement in alpine skiing which sometimes results in rupture of the ACL.

METHODS

The inverse dynamics approach was used to obtain the resultant external forces and moments about the knee joint. The human body was modeled as a mechanical system composed of six rigid bodies connected by smooth and slightly-loose joints. For each segment, the Newtonian equations of motion were solved to obtain the intersegmental resultant forces and moments. The system equations were calculated starting with the upper body (shoulder joint) and progressing to the thigh (knee joint). Alpine skiers were filmed during training and competition at a location where the falling back and subsequent recovery movement was likely to occur. Film data was digitized and transferred to a PDP 11/44 computer for analysis. The raw displacement data was smoothed and linear and angular accelerations of the rigid body segments were calculated. Segmental weights and transverse centroidal moments of inertia were obtained using Dempster's (1955) cadaver data. An estimate of the force due to air resistance was obtained from reported studies on wind tunnel testing of skiers (Luethi and Denoth, 1987). The system equations of motion were solved for the resultant external force and moment at the knee joint during the movement as a function of knee joint angle.

In order to calculate the forces acting on the ACL, the resultant forces and moments at the knee joint were distributed to the individual anatomical structures in the vicinity of the knee joint using the joint equipollence equations. The muscles, ligaments and bones included in this model were the quadriceps femoris and hamstring muscle groups, cruciate and collateral ligaments and the tibia and femur bones, respectively. Lines of action and

moment arms of the muscles and ligaments were obtained as a function of knee joint angle from cadaver specimens using a three dimensional digitizer. These data were subsequently transformed to the mid-sagittal plane of the lower limb.

The following five assumptions were made to reduce the unknowns to a determinate system: (a) the knee joint was assumed frictionless, therefore, the contact force acted perpendicular to the joint surface; (b) ligament moment arms about a transverse axis through the knee joint were assumed negligible compared to muscle moment arms; (c) antagonistic activity was initially assumed to be zero; (d) force in the posterior cruciate ligament was assumed to be zero as long as the resultant force was directed in the anterior direction and (e) force in the collateral ligaments was assumed to be 15 per cent of the force in the ACL. The three scalar joint equipollence equations were then solved for the three unknowns: patellar tendon force, articular contact force and anterior cruciate ligament force.

The two dimensional mathematical model was implemented on a SUN 3/280S computer for simulating the effect of changes in selected input variables of the forces calculated for the ACL. In particular, antagonistic activity of the hamstring muscles, initially assumed to be zero, was varied from zero to 90 per cent of the force in the quadriceps muscles.

A pilot computer simulation was performed where an arbitrary resultant joint moment of 100 newton-meters was to be satisfied. Lines of action and moment arms for the range of knee flexion from zero to 120 degrees were estimated for the quadriceps and hamstring muscles, based on data from the literature and from radiographic slides. Additionally, a constant value (20 degrees from the horizontal) for the line of action of the anterior cruciate ligament was assumed. Variations in antagonistic activity, lines of action and moment arms were simulated and the force response of the ACL was calculated.

RESULTS AND DISCUSSION

The results presented here are from the pilot simulation. An increase in hamstring activity was found to decrease the force in the ACL for all simulations. Force in the ACL varied slightly with changes in moment arm, but was very sensitive to changes in the line of action of the muscles, particularly the quadriceps muscle (patellar tendon). A patellar tendon line of action corresponding to full knee extension, resulted in the highest forces in the ACL and a line of action corresponding to full knee flexion, corresponded to the lowest ACL forces. From the initial findings, it appears that ACL forces are highly sensitive to the geometry of the knee joint and its associated force carrying structures.

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Differences in EMG-activity between patients with shoulder instability and normal controls

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Shoulder instability with recurrent dislocation and no history of previous significant trauma often occurs in individuals with general joint laxity (3). Altered skeletal geometry or altered activation of the shoulder muscles have been discussed as possible etiological factors. Surgical procedures are most often aimed at stabilizing soft tissues surrounding the joint but the clinical outcome is uncertain (6).

The aim of the present study was to record muscle activity from shoulder muscles in order to determine whether patients with shoulder instability showed different activation of their muscles compared to normal controls. If so, such data may improve the approach used in soft tissue surgery or the design of a muscle training rehabilitation program.

Methods

EMG recordings from twelve shoulders in six patients suffering from recurrent anterior shoulder dislocations were compared with recordings from ten shoulders in five normal controls. Normalized, full-wave rectified, low pass filtered and time-averaged EMGs were recorded in parallel from the subscapularis, supraspinatus, infraspinatus, the sternoclavicular part of the pectoralis major, the anterior, middle and posterior part of the deltoid and from the latissimus dorsi muscles. Intramuscular fine-wire electrodes (1) were used for the subscapularis (5), supraspinatus and infraspinatus muscles. For the other muscles, skin electrodes were used and placed over the muscle belly in the main direction of the muscle fibres (4).

The subjects used a weight-and-pulley apparatus loaded with 25 N (women) or 35 N (men) when performing internal or external humeral rotation. During abduction and flexion of the shoulder the subjects held a 20 N weight in their hand.

Results and conclusions

During *external rotation* with the upper arm abducted 90 degrees the patients with shoulder instability showed increased activity in supraspinatus and in some part of the movement also in infraspinatus.

During *abduction* the patients showed increased activity in supraspinatus but a decreased activity in infraspinatus and anterior part of the deltoid. During *internal rotation* subscapularis was more active in the patients than in controls. During *flexion* the anterior and middle parts of the deltoid were less active in the patients than controls and infraspinatus showed an increased activation at about 60-90 degrees of flexion.

Our results indicate that patients with shoulder instability show different activity in some shoulder muscles compared to normal controls. This is probably in order to maintain stability and prevent dislocation. The results indicate that muscle training programs should focus on the rotator cuff muscles and the coordination of deltoid. Surgical procedures should not only focus on subscapularis but also on the infra- and supraspinatus muscles.

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CHANGES IN LENGTH OF HUMAN RECTUS FEMORIS MUSCLE FIBERS AS A FUNCTION OF CHANGES IN KNEE JOINT ANGLE

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INTRODUCTION Force production of a muscle appears to be highly dependent on the length and rate of change in length of its contractile elements. These two variables may be approximated with the fiber length and rate of change in fiber length. Muscles have typically been modelled as homogeneous structures with a single muscle fiber length and a single velocity of contraction. However, pilot studies have indicated that muscle fiber lengths are not homogeneous within a muscle and current models may fail to explain experimental findings about muscular force production of intact human skeletal muscles. The purpose of this study was to quantify some non-homogeneities of intact human skeletal muscles, in particular, the dependence of changes in muscle fiber length during movement on the location of fibers within a muscle.

METHODS Fiber lengths were determined on 14 rectus femoris muscles of seven unembalmed human cadavers. Origins and insertions of one fiber (-bundle) from the proximal, middle and distal third of each muscle were marked using sutures. Fiber lengths were measured along the paths of the fibers between the two sutures using a tape measure. Each fiber was measured three times for a given joint configuration and the mean value of these three measures was recorded as the fiber length. Measurements were performed throughout the full range of knee joint motion at typically six different knee joint angles. If muscle fibers were damaged before full knee flexion was reached (i.e. longest muscle fiber length) the experiment was stopped.

Changes in fiber lengths were calculated from one knee joint angle to the next as the difference of the corresponding fiber lengths. Errors in the fiber length measurements were estimated to be maximally ± 2 mm.

Knee joint angles were defined as the included angles between the thigh (greater trochanter and lateral femoral condyle) and the shank (lateral malleolus, fibular head). They were measured using a goniometer. Errors in the angle measurements were estimated to be ± 2 degrees.

RESULTS AND DISCUSSION The results of this study are summarized in Table 1. The first column shows the lengths of fibers from the proximal, middle and distal portions of the rectus femoris muscle corresponding to an average sarcomere length of $2.72 \mu\text{m}$ (Abrahamse et al., 1988). The second column shows the mean change in length of rectus femoris fibers for a change in knee joint angle of 140° (i.e. from full knee extension to full knee flexion). The third and fourth columns show the range of active force production and the maximal velocity of fiber shortening, respectively. The values in column three are based on the experimental findings of Gordon et al. (1966), and those in column four are based, in principle, on the results reported by Hill (1938).

The mean lengths of the fibers of the proximal part of the rectus femoris are statistically longer than those of the middle and distal parts ($\alpha = 0.05$). Therefore, the range of active force production and the maximal velocity of muscle shortening are also larger for the proximal than the middle and distal fibers, if expressed in terms of fiber length and rate of change in fiber length, respectively.

The mean change in lengths of the proximal fibers throughout the range of knee joint motion is larger than that of the middle and distal fibers. That means, for a given change in knee joint angle and for a given rate of change in knee joint angle the change in fiber length and the rate of change in fiber length are larger for the proximal than the middle and distal fibers. Because of this the range of active force production and the maximal velocity of muscle shortening are smaller for the proximal than the middle and distal fibers if expressed in terms of the knee joint angle (Table 2). This result implies that the small middle and distal fibers of the rectus femoris muscle are better suited to produce large forces for fast movements which cover a large range of motion than the long proximal fibers. This result is in direct contrast to the result that would have been obtained, if changes in fiber lengths within a muscle were assumed to be homogeneous.

Based on this study it appears that a proper understanding of the force production of intact human skeletal muscle is highly dependent on the changes in length of muscle fibers during movement, a variable which has been ignored or has been assumed to be constant for all fibers within a muscle in the past.

	Fiber Length	Change in Fiber Length	Range of Active Force Production	Max. Velocity of Fiber Shortening
proximal	94 mm	62 mm	90 mm	940 mm/s
middle	84 mm	46 mm	81 mm	840 mm/s
distal	81 mm	43 mm	78 mm	810 mm/s

Table 1: Mean results for fibers of human rectus femoris muscles.

	Range of Active Force Production	Max. Velocity of Fiber Shortening
proximal	203 degrees	2123 deg./s
middle	247 degrees	2557 deg./s
distal	254 degrees	2637 deg./s

Table 2: Range of active force production and maximal velocity of muscle shortening expressed in terms of angular displacement and angular velocity of the knee joint, respectively.

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A Generalized Three-Dimensional Six Segment Model of the Ankle and the Foot

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INTRODUCTION

Participation in physical activity has changed during the last two decades: physical activity has become a part of every day life for an increasing number of people. This development started with the "jogging trend" in the late sixties and early seventies. A few years after the start of mass participation in running, other sports have become attractive with increasing numbers of participants. The most popular ones are swimming, cycling, tennis, and aerobics (Steinbrueck, 1987). The latter two sports frequently involve movements in medio-lateral direction, such as side shuffles, as well as movements in anterior-posterior direction, which are dominant in running. The frequency of sport injuries involving movements in medio-lateral direction is high. The most common injuries are ankle ligament sprain or rupture involving the lateral ligamentous complex and joint distortions. It has been suggested that the frequency of sports injuries during lateral movements can be reduced by changing external conditions such as shoe construction and surface properties (Nigg et al., 1986). However, quantitative analyses of the influence of external conditions on the loading of internal structures during lateral movements do not exist. Internal forces can be estimated using mathematical models. Several models are available to estimate internal forces during anterior-posterior movements (Burdett, 1982; Procter et al., 1982). These models can not be used to estimate forces during lateral movements since the relevant anatomical structures and functional divisions are not specified.

The purpose of this study was to develop a mathematical model to estimate forces and moments in joints, and forces in ligaments and muscles of the foot in a generalized three-dimensional situation.

METHODS

Anatomical model: The foot and ankle complex was modelled as a six segment system (segment 1: phalangeals of 3 medial rays, segment 2: phalangeals of 2 lateral rays, segment 3: metatarsals of 3 medial rays and cuneiforms, segment 4: metatarsals of 2 lateral rays and cuboid, segment 5: calcaneus, segment 6: talus; Figure 1), taking the medio-lateral functional subdivision of the foot into account (Mann, 1986). The joints connecting these segments and the tibio-talar joint were represented as hinge joints (Morris, 1977). The 12 long foot muscles were considered, being the important active force structures. Four ligamentous structures were selected based on their importance for the functional anatomy of the foot and their injury frequency (plantar aponeurosis, talo-fibular, calcaneo-fibular, and deltoid ligaments). The equations of motion for this system were derived using an inverse dynamic approach with external forces acting on each segment which is part of the plantar surface of the foot.

Muscle and control model: Each muscle-tendon unit was assumed to consist of three different components which are arranged in series: the fiber elastic component within the contractile element, the contractile component of a muscle fiber, and the tendon as elastic structure (Pierrynowski et al., 1985). Muscle fibers were categorized into three different types (SO, FO, FG). The maximal possible force output of a muscle at a certain time was calculated considering the geometry, the elasticity of the serial elastic components, temporal phenomena (prior activation), muscle fiber type, the force-length, and the force velocity relation. The stimulation of a muscle and, therefore, the actual force output at a certain time was calculated with a neurophysiological, central pattern generator (CPG) based control model (Pierrynowski et al. 1985). Each joint allows for a one degree of freedom movement (hinge joint) and is controlled by one CPG. This CPG facilitates muscles

producing a moment in the direction of the required net joint moment and inhibits all other muscles. Muscles crossing more than one joint receive input from the CPGs of all joints they are crossing. This model therefore includes the influence of co-contraction for the bone-to-bone contact force in a joint.

Application: An EMED pressure distribution insole system with 72 sensors was used to determine the external forces acting at each segment. Each sensor was assigned to a segment by means of a dorsal X-ray picture. The resultant external force and its point of application was then calculated for each segment. The location of each segment with respect to certain skin markers (above bony landmarks) through the full range of motion of all considered joints, was determined in an in-vitro study. One cadaver specimen, matched with the test subject by means of 28 anthropometric measurements, as well as a lateral and a dorsal X-ray picture, was used in this study, since it was shown to be not appropriate to use averaged cadaveric information (Engsberg, 1987). The length of the ligamentous structures of interest through the full range of motion was also determined in the cadaver study and the length of these structures was related to force with the help of a force-length relation, derived from the same structures. The results of the model at the tibio-talar joint level will be compared to a simple one segment model of the foot. A lateral side shuffle movement, executed under various conditions, will be the first application, to investigate the influence of different shoe constructions on internal loading. EMG information will be used to compare the predicted muscle stimulation pattern with the actual stimulation.

CONCLUSIONS

The model developed in this study is not restricted to lateral movements but can be applied to all kind of movements. It is formulated in a truly three dimensional way and is the first model available, which considers the medio-lateral functional subdivision of the foot and local forces.

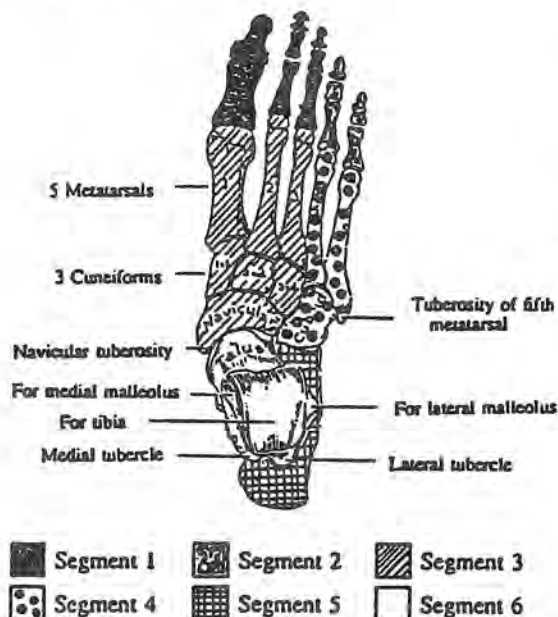


Figure 1: Dorsal aspect of the bones of the right foot. The segments of the model are indicated.

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Biomechanical and CT Analysis of Triplanar Facial Strut Trauma Following Steering Wheel Impact

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Introduction

The steering assembly has been implicated as a causal agent inducing facial injuries in European countries where seat belt requirements have been in effect for a number of years. National Crash Severity Studies indicate that the steering assembly is responsible for 53% of facial injuries in seatbelt restrained occupants. Zygomatic fractures routinely occur because of the prominence of the mid-facial structure. Despite some early biomechanical studies to determine the fracture thresholds of the zygoma, limitations in the experimental design prevents a complete understanding of its dynamic characteristics (1,2). Injury biomechanics of the zygomatic complex with steering wheel impacts has not been explored. In this study, we investigated the injury biomechanics and the probability of facial bone fracture following dynamic contact with steering wheels (Energy absorbing - EA, and standard).

Materials and Methods

Twenty-two fresh human cadavers (age 57-76 years, height 155-196 cm, and weight 45-109 kg) were used in this study. The heads were isolated at OC-C1 level and sealed in plastic bags at -2°C. Two-dimensional computed tomography (2-D CT) was done using a General Electric scanner (GE 9800) with 1.5 mm sections. Radiographs were taken in the anteroposterior, lateral and occipitomeatal (waters) views prior to test. Three-dimensional (3-D) CT's were also done with the Dimensional Medicine Model 3200 system.

Quasistatic experiments were conducted using the MTS testing device. Isolated specimens were attached to the piston and oriented suitably so that the junction of the left lower spoke and rim (JLSR) of either steering wheel contacted the zygoma. Studies were also done to determine the quasistatic mechanical response of the steering wheels at the JLSR.

Nine specimens were impacted on the EA wheel with velocities of 6.93 - 3.58 m/s and six specimens were impacted on the standard wheel at velocities of 3.13 - 2.24 m/s. The specimens were rigidly mounted in a neurosurgical halo ring with fixation into the skull. The specimen was attached to a fixture of a specially designed 7.6 m vertically supported monorail. A cart with precision bearings was designed to traverse the monorail. The positioning fixture was designed to accept specimens, and allowed 0-360° movement in two mutually orthogonal directions for impaction of either zygoma.

A triaxial accelerometer was rigidly attached to the posterior parietal skull area opposite to the side of impact. The specimens impacted the JLSR of the steering wheels placed 30° to the horizontal. A six-axis load cell was placed directly underneath the hub of the steering wheel to record the generalized force histories. Steering wheel deformations were recorded with a combination of linear and rotary potentiometers placed below the steering wheel at the impact site. The recorded generalized force histories of the hub

were transformed to the impact site using the potentiometer data and transformation principles. All biomechanical data were collected at 8kHz sampling using SAE-J211b specifications. Sixteen mm films at 1000 fps were taken to document the kinematics. After test, the specimens were evaluated using x-rays, 2-D CT, 3-D CT and defleshing techniques. Facial trauma was assessed using the method of triplanar facial struts (3).

Results and Discussion

Forces to induce zygoma fracture in quasistatic studies ranged from 1172-2352N with piston displacements of 32.37 - 46.14 mm. In one specimen, no fracture to the zygoma occurred at a force of 2108 N. Results of the quasistatic studies of the steering wheels indicated that the mechanical response was nonlinear and sigmoidal. The EA wheel was softer (linear stiffness = 110.25 N/mm) compared to the standard steering wheel (linear stiffness = 139.06 N/mm). However, the energy absorbing capacity of the EA wheel was about 50% of the standard steering wheel at a deformation of 25.4.

In dynamic tests, fracture forces ranged from 1770-4604 N for EA wheels impacts and from 1499-1591 N for standard wheel impacts. Higher impact velocities produced, in general, higher forces, accelerations, contact areas, and steering wheel deformations. The head injury criteria (HIC) for fractured specimens varied from 65-383 for the EA wheel. One specimen on the EA wheel at an impact velocity of 2.68 m/s and another specimen on the standard wheel at an impact velocity of 2.01 m/s did not produce clinically significant fractures to the facial structure. Fractures were classified into solitary strut fractures, complex strut fractures and transfacial fractures. These included: orbital floor, zygomatic arch, maxillary sinus and nasal arches for the solitary strut classification, nasofrontal, nasoethmoidal and tripod for the complex strut classification, and the three classic LeForts for the transfacial fracture classification. These were done using the 1.5 mm CT images. A good correlation was found between the defleshed preparations and CT images. While it was difficult to detect hairline and some mild non-displaced fractures in 3-D CT images, clinically significant fractures (described above) were routinely identified.

Fractures correlated well with peak transformed forces at the impact site; however, a weak correlation was found with maximum resultant accelerations, HIC, steering wheel deformations, and contact areas. These findings suggest that a criterion to evaluate the probability of facial fracture should be based upon the impact force.

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BIOMECHANICS OF LUMBAR FACETECTOMY AND DISCECTOMY

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INTRODUCTION

Apophyseal joints, intervertebral discs and vertebral bodies are the medium through which a primary transfer of external force (generalized) takes place along the spinal column. Muscles and ligaments add to the stability of the spine. The instability of the facet joint and degeneration of the intervertebral disc have been implicated in the etiology of low back pain. Lumbar facetectomy and lumbar discectomy, unilateral or bilateral, partial or complete, which are frequently done during many surgical procedures, alter the mechanism of load transfer and segmental stability of the low back. Although some preliminary studies have been conducted to study the mechanics of facetectomy or discectomy, a quantification of the biomechanics of facetectomy and discectomy is not available. Therefore, we conducted *in vitro* experiments using human cadaver spines to study the strength and stability due to these two common operative procedures of the lumbar spine.

METHODS

Six intact, functional units (FU) from three fresh human cadaver spines were used in the study. Three FU were taken from the L2-3 level and three were taken from the L4-5 level. This approach was done to take into account the variability in the anatomical characteristics of the lumbar spine. All subjects had died from cardiac disease, and the specimens were x-rayed and examined to rule out metastasis, pre-existing trauma or severe degenerative disease. All muscles and fat were removed from the spines without causing injury to the ligaments. The lumbar spines were stored at -20° C until the day of experimentation and thawed to room temperature. L2-3 and L4-5 FU were obtained from the spines by transecting the appropriate intervertebral discs at mid-height. These FU were fixed both superiorly and inferiorly in polymethyl-methacrylate; the plane at the disc mid height was kept parallel to the surfaces of the fixed specimen. The intact FU were tested in compression-flexion mode of loading by placing the specimens below the piston of a MTS testing device. A low magnitude load 200-400 N was applied at a constant deformation rate of 2.5 mm/s and the specimen was unloaded. The compression/flexion load was applied at an eccentricity of 5 cm anterior to the center of the disc. The load-deflection curve was checked to insure that the specimen was still within elastic limits. Later, the intact specimen was tested under injured conditions: facetectomy; interspinous and supraspinous ligament transection; and posterior anulus disruption with partial discectomy. Kinematic information was derived from the motion analysis of retro-reflective targets placed on the vertebral bodies, articular facets, and spinous processes at the superior and inferior levels. The load and deflection data of the piston was obtained by recording these variables as a function of time on a digital oscilloscope.

RESULTS AND DISCUSSION

The mechanical compressive load-deflection response was non-linear with continuously changing stiffness (a measure of the ability of the structure to withstand the externally

applied force vector). Due to this non-linear behavior, the mechanical response was evaluated at discrete deformation intervals to determine the loss in strength due to injury: facetectomy and discectomy. Facetectomy altered the capacity of functional units to withstand increasing compression-flexion loads resulting in a decrease in strength of approximately 28-50% compared to the intact specimen. Posterior anulus resection with partial discectomy resulted in a shift of the mechanical response towards the abscissa implying an increase in the flexibility (decrease in stiffness) of the specimen. Loss of strength up to 77% compared to intact was observed. Motion significantly increased due to facet disruption and disc removal. Kinematic analysis indicated that the changes in the linear translation in the postero-anterior (PA) direction following the facetectomy and discectomy were of lesser magnitude than found in the intact specimen (PA movement of 2.2 mm with intact, 1.5 mm with facetectomy and 0.8 mm with discectomy) suggesting a movement of the instantaneous center of rotation anteriorly with facet joint disruption and disc removal. The increase in the facet distractions in these injured preparations compared to intact (Figure 1) was higher in discectomized and facetectomized specimens without interspinous and supraspinous ligaments (2.73 ± 0.83 mm) followed by facetectomized preparations without ligaments (1.27 ± 0.08 mm), and facetectomized preparations with intact interspinous and supraspinous ligaments (1.03 ± 0.11 mm). These increases in the flexibility and localized deformations in the components of the spine under a single-cycle load application may offer insight into the possible onset of instability with certain common lumbar operative procedures. This may render the spine additionally unstable under complex physiologic cyclic loading.

ACKNOWLEDGMENT

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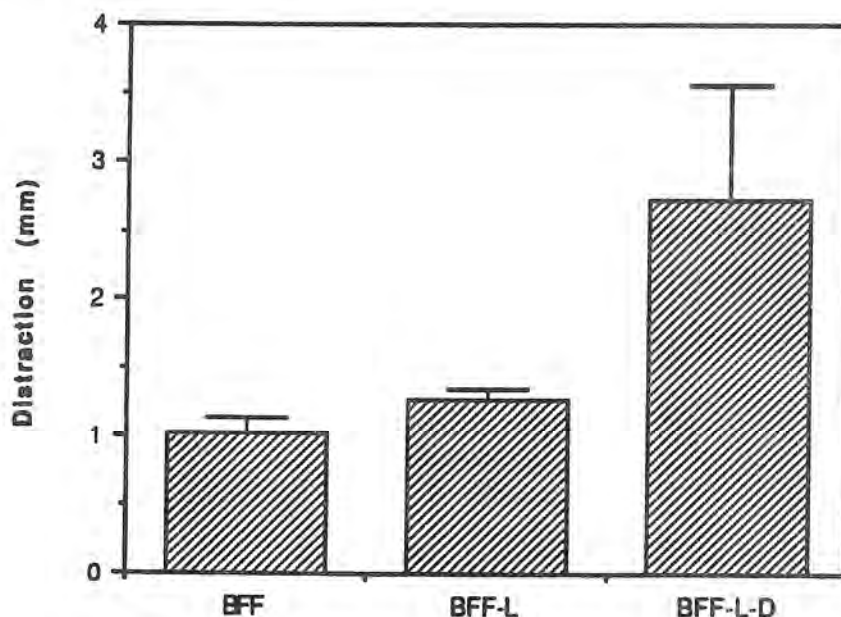


Figure 1. Graph representing increase in distractions of the facets (compared to intact) after bilateral facetectomy (BFF), bilateral facetectomy with interspinous and supraspinous ligament transection (BFF-L), and posterior anulus resection with partial discectomy (BFF-L-D).

CORRECTION OF POINT OF APPLICATION READINGS OF PIEZO-ELECTRIC FORCE PLATES

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Introduction

In the study of locomotion, force plates are used to measure the vertical (F_z), fore-aft (F_y) and medio-lateral (F_x) components of the ground reaction force vector, as well as its point of application. A commonly used force plate, equipped with piezoelectric transducers, yields very accurate force measurements, but the accuracy in the calculated point of application is not so good. Typically, errors ranging between -30 and 30 mm may occur, depending on the position where force is applied. Errors of this magnitude may be unacceptable for certain applications, for example in the calculation of inversion/eversion moments about the subtalar joint in human walking or running. The purpose of the present study is to evaluate the accuracy of the point of application determined with a commonly used piezoelectric force plate (Kistler type 9287), and to develop a correction algorithm that reduces the magnitude of the observed errors.

Materials and methods

Test were performed with three force plates, each of which was installed on a mounting frame grouted into a concrete foundation. The force plates were connected to electronic amplifier units (KISTLER type 9861A) operating at ranges of 2000 N in F_z , and 500 N in F_x and F_y . The 8 output signals of each unit were sampled, and F_z and the coordinates of the point of application (x and y) were calculated according to the manufacturer's instruction (Kistler, 1984). Static vertical forces between 0 and 2000 N were applied at 117 positions, which were defined by a grid of perpendicular lines, 9 parallel to the y -axis, 13 parallel to the x -axis, with a spacing of 70 mm. The forces were applied on a sturdy wooden board, equipped with a ball shaped stylus, fitting in small holes drilled at each intersection on the grid. The same set up was used for point of application accuracy determinations during dynamic loading, where a subject ran over the board. The accuracy of the point of application determination was evaluated using:

$$X = \sum_{p=1}^{117} (\hat{x}_p - x_p)^2 \quad \text{and} \quad Y = \sum_{p=1}^{117} (\hat{y}_p - y_p)^2$$

where x_p and y_p are the actual coordinates of the point p where F_z was applied, and $\hat{x}_p$ and $\hat{y}_p$ are the calculated coordinates. Based upon the observed error distribution, the following correction algorithm was selected:

$$\hat{x}_c = \hat{x} - (P\hat{y}^2 + Q) [\hat{x}(\hat{x}^2 - A^2)] \quad \text{and} \quad \hat{y}_c = \hat{y} - (R\hat{x}^2 + S) [\hat{y}(\hat{y}^2 - B^2)]$$

where $\hat{x}_c$ and $\hat{y}_c$ are the corrected coordinates, A is 210 mm (half the distance between the transducers along the x -axis), and B is 350 mm. P , Q , R and S are constants whose values were determined by quadratic regression analysis of the errors measured in one plate. Subsequently, this algorithm was applied to correct errors in $\hat{x}$ and $\hat{y}$ of the other two plates.

Results

If F_z was increased from 0 to 2000 N, $\hat{x}$ and $\hat{y}$ were found to plateau between 1000 and 2000 N. The errors $\hat{x}_p - x_p$ at 1500 N were found to range from -7 to +6 mm within the transducer rectangle, and from -15 to +16 mm at the edges. Errors $\hat{y}_p - y_p$ at 1500 N varied between -15 and +11 mm within the transducer rectangle and between -19 and +20 mm at the edges. Error directions and magnitudes were symmetrically distributed with respect to the center of the plate. Table 1 shows that applying the correction algorithm to the three plates increases the accuracy considerably. In dynamic loading, the observed errors were essentially the same as in the static experiments.

Table 1. This table shows for three force plates the improvement in accuracy of the calculated point of application. See text for definition of X and Y (values are expressed in mm²).

	PLATE I		PLATE II		PLATE III	
	X	Y	X	Y	X	Y
before correction	3125	6221	3134	7773	3134	7645
after correction	534	541	452	1146	351	507

Discussion

In this study, an algorithm was developed to improve the accuracy of point of application determinations for one type of piezo-electric force plate. The algorithm represents a function which is symmetrical with respect to the middle of the plate. Application of the algorithm increased the accuracy considerably in three plates of the same type (Table 1). Such an increase in accuracy is particularly useful when the position of the foot on the force plate varies randomly from trial to trial, such as in research on running.

Considering the high accuracy of the force plate in measuring forces, the question arises how the inaccuracies in the calculated point of application are caused. Since the error distribution is the same in more than one plate and is symmetrical with respect to the center of the transducer rectangle, it is likely that the errors are inherent to the method used to determine the point of application. One possible explanation is that the output of the individual transducers does not increase linearly with the force input. This is very unlikely: the deviation required to explain the errors in the point of application by far surpasses the linearity indicated by the manufacturer, which was confirmed by own measurements. The second explanation is that the resultant force on each individual transducer is not acting at the transducer's center, as assumed by the manufacturer. Such a situation could be caused by a minimal bending of the top plate. Although modeling of the influence of bending might be possible, it is unlikely that it will lead to a more effective algorithm than the one developed in this study.

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3-D CAMERA CALIBRATION WITHOUT A CALIBRATION OBJECT

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Camera calibration in lower limb or whole-body movement studies is usually thought to require a large calibration object which should span the full field of view. In biomechanics, the Direct Linear Transformation (DLT) of Marzan & Karara (1975) is rather popular, despite its need for a voluminous and accurate, three-dimensional calibration object. Since such an object may be costly and difficult to construct, and rather unwieldy in non-laboratory environments, Woltring (1980) described a procedure (SMAC, or Simultaneous Multiframe Analytical Calibration) requiring a number of planar calibration grids to be placed at uncalibrated positions and attitudes throughout the field of view. One of these positions and attitudes defines the object-space coordinate system, and the remaining positions and attitudes are estimated simultaneously with the required camera parameters including those describing image distortion. Dapena et al. (1982) described a procedure in which the "internal" and "external" camera parameters were estimated in distinct phases; however, this may result in estimates for the internal parameters which are not appropriate for the actual field situation. Recently, Wood & Marshall (1986) and Hatze (1988) investigated the DLT in terms of its capacity to allow extrapolation outside the object's volume and in terms of its numerical properties.

The results of Dapena et al. (1982) suggested that the use of a small calibration object (in his case, a one-dimensional stick with a number of known points on it) can be sufficient for calibrating a 3-D camera configuration. However, the procedure of Woltring (1980) requires at least a two-dimensional calibration object. Motivated by a practical case of missing calibration information on a knee joint study, the possibility was considered that the 3-D landmark clusters implanted in the subjects' femur, patella, and tibia might provide sufficient photogrammetric control to allow full calibration, where the cluster at the femur in mid-stance was (arbitrarily) taken as the reference cluster. The original initialization procedure of Woltring (1980) and its subsequent, unpublished modifications could not accommodate this situation. However, the work of Hatze (1988) suggested that a constrained DLT could be used, where less than the minimally required, six spatially distributed landmarks might be sufficient if prior knowledge of some internal camera parameters is imposed on the DLT. This was feasible since the principal point and distances and the image shearing factor ALPHA (cf. Woltring, 1980) were approximately known. These prior estimates are merely used during initialization and are left free during the iterative phase of the calibration process. However, if the calibration distribution is very shallow (e.g., during straight locomotion), it is possible to strengthen the calibration process by slightly constraining the principal points and distances.

For simulated data commensurate with high-precision film equipment, the modified SMAC-procedure was found to accommodate reliable recalibration of a configuration with three cameras at 2 m from the subject, spaced at 30 degrees. The landmark clusters contained four points arranged in pyramids with side lengths of approximately 6 cm. Nine clusters were observed (three at heel strike, three at mid stance, and three at toe off). Thus, it appears possible to "self" calibrate a camera configuration on the moving subject, without the need for a cumbersome, separate calibration device. It should be noted, however, that this approach allows reliable assessment of relative movement only; the stability of the object-coordinate

system defined in terms of the designated reference cluster is low, in view of the small size of the cluster. Typically, one might combine this approach with a set of markers affixed to an object of larger size in the field of view, such as a force plate, and designate this object to be the reference object.

Interestingly, there was no need to constrain the parameter ALPHA during the iterative process if its true value is equal to zero. For realistic configurations and noise levels, ALPHA had very small values (typically, on the order of 0.01), and constraining it to zero had only a slightly beneficial effect. The same was found for the constrained DLT procedure designed for this study. A possible explanation might be in the way that Marzan & Karara (1975) transformed the DLT parameters into conventional camera parameters where they assumed $ALPHA = 0$ which is true for orthogonal axes only. This might not be the case for data from video scanning or distorted cinematographic film.

For the real data in this study collected with a SELSPOT-II system, some care was required to properly initialize the procedure. The clusters in a selected reference position (mid-stance) were mimicked as well as possible, and their 3-D landmark coordinates were reconstructed for a mimicked camera configuration, following a normal calibration procedure. In order to reduce the influence of measurement noise, data were sampled and averaged over 2 seconds; however, it was found that the standard deviation in the reconstructed coordinates was well below 0.5 mm. Subsequently, the calculated 3-D coordinates were used as local reference data and this procedure resulted in a residual image standard deviation of only 3 image digitization units.

While SMAC was also found to converge under its non-linear distortion mode option (radial distortion only), no improvement in accuracy was obtained. This was to be expected since the SELSPOT-II optics have been designed for minimal distortion, unlike the lenses recommended for SELSPOT-I in the past. Further research is in progress to determine the utility of the modified procedure for routinely "self-calibrating" cameras from actual movement data.

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A COMPUTERIZED METHOD FOR REGISTRATION OF SAGITTAL BACK SHAPE AND MOBILITY

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The reliability of a new computerized method for registration of thoracal and lumbar back shape and mobility was tested by one examiner on six healthy volunteers on five consecutive days. A flexi-curve was applied on the back of the subjects from the spinal process of the first thoracic vertebra to the sacral bone. The shape of the flexicurve was then fed into a personal computer by moving a mouse along the curve. Registrations were done with the subjects standing and sitting with straight back, and with maximally flexed and extended back. The program was written in pascal. Many different parameters were registered such as the distance between THL and S2, the inclination angle of the back, where the mobility takes place in the back, level and degree of thoracal kyphosis and lumbar lordosis. In addition a picture of the shape of the back was shown.

The results showed a varying reliability for the different parameters. The method should be useful when recording the clinical course of a disease and for evaluation of different treatments.

Two-page abstract not received in time for publication.

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MULTI-ELECTRODE DATA ACQUISITION AND PROCESSING OF ELECTROMYOGRAPHIC (EMG) RECORDINGS.

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Introduction and Theory: The concept of noise in EMG processing is not universally agreed upon. A portion of the noise is believed to be caused by the spatial relationship between the recording electrodes and the muscle. Lindstrom, *et al.* [1] showed that the distance between the electrodes and an active myoelectric site affects both the gain and phase of the recorded signal. As the distance increases, the tissues cause a greater attenuation of the signal. EMG recordings from a single pair of surface electrodes, therefore, cannot be representative of the electrical activity in the entire muscle over all time [2,3,4]. Seigler *et al.* [4] found that more information about the myoelectric signal was needed to improve the estimate of the active state of a muscle and hypothesized that the use of multiple electrodes may be the answer. The purpose of this paper is to present and demonstrate an EMG acquisition and processing approach that can be used to better estimate the electrical activity of the muscle. This approach uses multiple pairs of electrodes to record the myoelectric activity to reduce spatial noise.

Model: The processing model consisted of two phases: 1) acquisition and preprocessing of the EMG signals, and 2) spatial compensation. Four bipolar pairs of Ag-AgCl electrodes (8 mm diam.) were used over each muscle. The four pairs of electrodes were placed at the same level of the muscle with the distance between the bipolar sites set at 1.5 cm as suggested by Basmajian and DeLuca [5]. Each EMG channel was put through a root mean square (RMS) circuit and then recorded. The recorded RMS EMG channels (REMGp) were first normalized with respect to each other at constant contraction levels to compensate for differences in skin impedance between electrode pairs. Once normalized, the channels for each muscle were averaged with equal weighting as,

$$REMG_d(n,m) = \frac{1}{4} \sum_{a=1}^4 REMG_p(n,m,a)$$

where n is the point index, m is the muscle group, and a is the electrode pair for that muscle group. These new signals (REMGd) are then the "derived" electrical activity of the muscle groups and represent an average REMG activity.

In order to quantitatively evaluate improvement of the proposed EMG modeling procedure, a repeatability measure based on Hershler and Milner [6] was used. The repeatability criterion was a ratio of the variance across repeated trials at each point in time to the total variance for all the trials. The formal definition is given by:

$$VR = \frac{\sum_{i=1}^k \sum_{j=1}^n \frac{(E_{ij} - \bar{E}_i)^2}{k(n-1)}}{\sum_{i=1}^k \sum_{j=1}^n \frac{(E_{ij} - \bar{E})^2}{(k n - 1)}}$$

where k = number of time points, n = number of trials, E_{ij} is the actual value of the j th EMG signal at time i and $\bar{E}_i$ is the average of the EMG values at time i averaged over j trials. $\bar{E}$ is the grand mean over all trials. The variance ratio (VR) ranges between 0 and 1. As VR approaches 1 the EMG signals are totally irreproducible. As VR approaches 0, the EMGs are completely reproducible.

Experimental Methods: Experiments were conducted to verify the improvement in repeatability of the multi-electrode model for the tibialis anterior and soleus muscles. These experiments consisted of repeated trials of isometric exertions. The laboratory equipment included EMG and torque transducing instrumentation connected to an HP-1000 minicomputer. The subjects were seated with the foot firmly attached to a fixed ankle torque transducer that measured the resultant isometric torque about the ankle in both plantar flexion and dorsiflexion. An oscilloscope was positioned in view of the subject with two signals displayed. The top

signal was produced by a signal generator while the lower signal was an output from the torque transducer in the foot fixation apparatus. Thus, the lower signal was controlled by the amount of torque the subject created about the ankle. During the trials, the subjects were asked to follow the signal produced by the signal generator by applying appropriate ankle torque to the fixture. Three different types of trials using different generated waveforms were used: a slow sine wave (5 Hz.), a faster sine wave (1.0 Hz.) and a square wave (1.0 Hz.). Fifteen trials were run for each condition, lasting 8 s per trial.

Results: VRs were estimated for 1) all single pairs of electrodes, 2) all combinations of two pairs, 3) all combinations of three pairs, and finally for 4) all four pairs together. This was done for both of the muscle groups. Figure 1 shows the VR values for the two muscles as a function of the number of electrodes averaged together. The mean VR values decreased as a function of the number of pairs of electrodes included. This decrease in VR appeared to be an exponential decay as the number of pairs was increased. This relationship was seen in both muscles. The standard deviations of the VR estimates also decreased as more electrode pairs were included. A Newman Keuls statistical test found a significant decrease in the VR between using one pair and two pairs of electrodes. In two of the three subjects there was also a significant decrease between two and three pairs of electrodes. In a separate analysis the difference between including three or four pairs was not found to be significant.

Discussion: The general pattern of electrical activity seen in the four physical channels were consistent with expected rises in activity with increased torque generation. There were, however, differences in the secondary bursts of activity recorded. These bursts were not believed to be simply processing noise caused by the RMS estimator but representative of actual EMG activity from specific areas of the muscle. These bursts represent significant variations of the input to the muscle and eventual output of force. By including the multiple estimates of this activity these regional effects can be incorporated. The repeatability study showed that multiple electrodes improve the reproducibility of the REMG. As the number of electrodes was increased, the repeatability of the

signals increased, however, the rate of improvement was minimal after inclusion of three electrodes. It is suggested from these results that a minimum of three electrode pairs be used in multi-electrode EMG data acquisition and that no more than four are necessary. The multi-electrode work of Hogan and Mann [3] also found that the improvement in EMG processing increased number of electrodes on the biceps brachii. Even though their procedures differ significantly from those presented here, their results also showed that the rate of improvement decreased as the number of electrodes increased. A future expansion of multi-electrode EMG analysis is to use many more electrodes over more than one muscle. The processing of these arrays would be complex, however, the benefits of observing multiple muscle actions non-invasively would be great.

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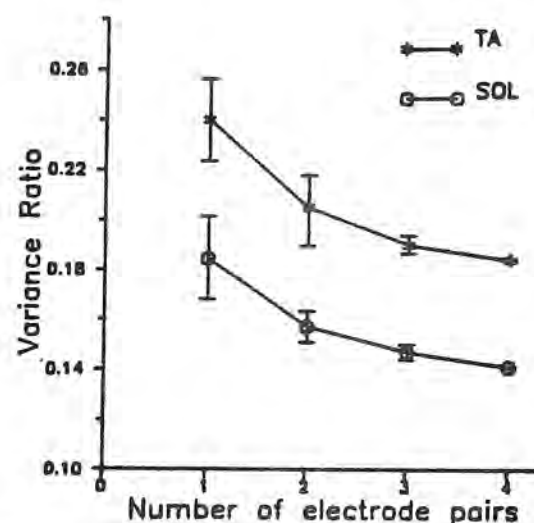


Figure 1: Variance Ratios for the tibialis anterior and soleus.

GROUND REACTION FORCES IN ELITE HIGH JUMPING

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Introduction : Fosbury jumpers often have to deal with pain and injuries at the take-off foot. According to several investigators (Krahl et al., 1978; Muraki, 1983) the main cause of these foot problems is due to the fact that the take-off foot is planted primarily in an abducted position and not in the direction tangential to the curved run-up. This positioning of the foot combined with high ground reaction forces at the moment of take-off results into abnormal foot deformations which are most frequently found medially at the ankle joint.

Methods : In this experiment, ground reaction forces (GRF) were recorded during an indoor competition in which elite high jumpers participated. Since there were athletes running up from the left-hand side and others from the right-hand side of the track, two Kistler force plates (type Z 4852, 90 cm x 60 cm) were built in. Because the take-off place varied significantly from one athlete to the other and sometimes even for one athlete from one performance to the other, GRF could only be measured for a limited number of trials. The GRF of two athletes who hit the force plate at each of their trials (P. Sjøberg, world record holder at the time of the experiments with 2.42 m. and S. Matei, best performance: 2.32 m.) will be discussed. The GRF acting on an athlete can be divided into high- and low-frequency forces. The first can be considered as "impact" forces, the latter as "active" forces which are dominantly influenced by the movement an athlete is going to perform after the impact. Simultaneously with the recording of the GRF, high speed video images of the take-off foot were obtained for qualitative purposes (cf. figure 1).

Results and discussion : In table 1 the impact and active forces of the two athletes are shown in relation to the performance level. The vertical impact forces vary between 6100 N (8.4 BW) and 7150 N (8.9 BW). These values exceed the figures reported by Baumann and Stucke (1980). They found values ranging from 3000 to 6000 N. The horizontal impact forces, ranging from 4100 N (5.6 BW) to 4750 N (6.5 BW), are also higher than the values of Baumann and Stucke which varied between 2500 N and 4000 N. The ratio between horizontal and vertical impact forces amounts to 2/3 or more, a value which is rarely found in other track and field disciplines with the exception of javelin throwing (Deporte and Van Gheluwe, 1987). The fact that a high jumper changes the direction of his centre of gravity by about 70° within 0.15 s explains the large GRF during the take-off phase of this particular movement. The vertical

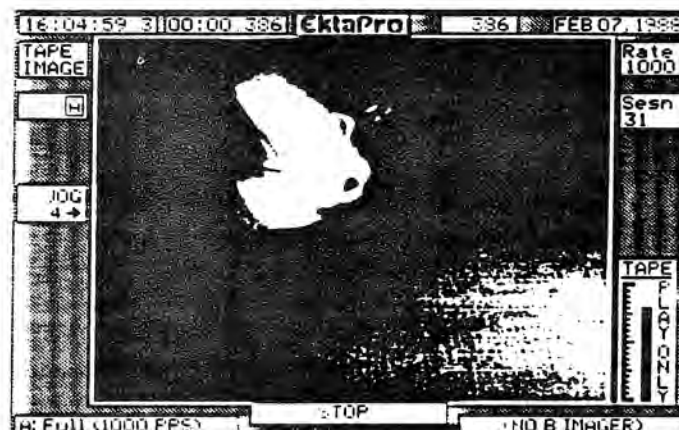


Figure 1 : High speed video images (Kodak Ektapro 1000 Motion analyser - 1000 frames/s) of the take-off foot of an athlete recorded under actual competition conditions.

active forces also reach high values up to 5300 N (7.3 BW). Interesting to note is that the world recordholder developed significantly less vertical active (propulsive) force than the other athlete (values in BW). A better bar clearance technique and/or a more advantageous position of the centre of gravity (P.S. : 199.0 cm, 83.0 kg and S.M. : 183.2 cm, 74.3 kg) are possible explanations.

		athlete	unit	2.20	2.25	2.30	2.34	2.38	2.42*	2.42* (m)
Impact										
Vertical GRF	P.S.	N	---	7150	6850	6850	7050	6950	7100	
		BW	---	8.8	8.4	8.4	8.6	8.5	8.7	
	S.M.	N	6100	6300	6150	6500*	---	---	---	
		BW	8.4	8.6	8.4	8.9	---	---	---	
Horizontal GRF	P.S.	N	---	4500	4250	4300	4700	4700	4550	
		BW	---	5.6	5.2	5.3	5.8	5.8	5.6	
	S.M.	N	4100	4600	4500	4750*	---	---	---	
		BW	5.6	6.3	6.2	6.5	---	---	---	
Active										
Vertical GRF	P.S.	N	---	4800	4700	4800	4700	4400	5400	
		BW	---	5.9	5.8	5.9	5.8	5.4	6.6	
	S.M.	N	5050	5150	5200	5300*	---	---	---	
		BW	6.9	7.1	7.2	7.3	---	---	---	
Horizontal GRF	P.S.	N	---	2500	2600	2550	2550	2450	2700	
		BW	---	3.0	3.2	3.1	3.1	3.0	3.3	
	S.M.	N	2500	2800	2850	2800*	---	---	---	
		BW	3.4	3.8	3.9	3.8	---	---	---	

* These heights were not cleared, but the trials can be considered as valid in terms of GRF

Table 1: Impact and passive forces ($\pm 3\%$) in relation to the performance level for two elite high jumpers

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Video Analysis of Take-off Actions
in Ski Jumping

----- Finding parameters for predicting flight length -----

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The performance in ski jumping is determined by the approach run, the take-off, the flight and landing. The most important parts are the take-off and flight. Sagesser (1981) has used a new force measuring system which was installed on the jumping platform. He predicted the length of the jump by determining at what point in the jump the peak force was exerted. This system of measurement was concerned with the vertical force exerted during the jump on force plates which were placed parallel to the ground below the jumping surface. We thought it necessary to determine not only the strength of this force but also its direction and timing. This, then, would be useful in clarifying the relationship between our parameters and flight length. Accordingly, the force directions can be predicted from examining the acceleration rate, which is determined by analyzing the take-off actions in the videos. The purpose of this paper is to find the optimum parameters for predicting the flight length in ski jumping using video movie analyses.

Method

It is Generally well known that knowledge of the velocity and acceleration are obtained by video analyses. When the World Cup Ski Jump Competition was held in Sapporo in 1987, we tried to analyze the take-off actions of several jumpers by video movie, including the winner (Nikkanen M.). Its frame speed was sixty frames per second. During the competition, we could not reserve a suitable position to take the pictures. However, we took pictures from a position slightly to the side of and facing the jump platform. Then the raw data were modified through a projective transform equation. And a video analysis system was made using a personal computer (NEC PC-9801).

Results

We thought it possible to find features of take-off motion by observing jumpers' forms in the take-off phase. Figure 1 shows take-off actions of two jumpers who were the top two ranking in the World Cup Series in 1987. It was understood as shown in the stick pictures, that the bodies above the knee were moving largely on their knee joints. And this motion was similar in six jumpers. However, we found differences in the motion of the upper half of the body. In the three jumpers who gave the best performances in this competition, we found that the upper half of the body moved upward very little on the take-off

platform. However, in the case of three Japanese jumpers, their upper bodies were moving upward earlier than the rest of their bodies.

We predicted influence on the jump performance because such motion makes an increased area of the body hitting the air as the jumper moves through the air. It makes an increase in the air resistance. It is the concern of each jumper to be in the position which creates the least drag force. And during take-off, the jumper's goal is to improve his ability to control the body and its actions. The acceleration vector shows the force direction of each jumper. Both the variation of the force and the timing of the maximum magnitude when the jumper pushes off from the platform, are shown by acceleration vectors. The vectors indicate a different style for each jumper. Then we examined the relationship between acceleration vectors and flight length. The acceleration vectors are divided into two components on the right angle coordinates (X-Y axis).

We searched for a way to predict flight length (Y) using the multiple regression. The parameters were obtained from acceleration data on the horizontal axis. One is the peak magnitude of acceleration (X1), and other is the point (X2) at which peak magnitude is exerted by the jumper on the platform. The equation of regression is as follows : $Y=97.65+1.45X_1+192.86X_2$. Figure 2 shows the relationship between the predicted and the actual flight lengths. The multiple correlation was near 1.0 ($R=0.9808$). So we thought their parameters were suitable for predicting the flight length.

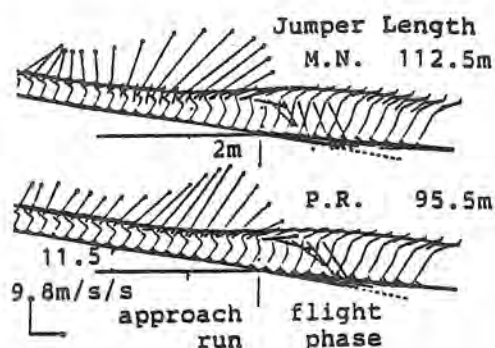


Fig.1 : The take-off action is shown by stick pictures. And the magnitude and direction of acceleration is shown by the vector line in the center of mass of the jumper's body.

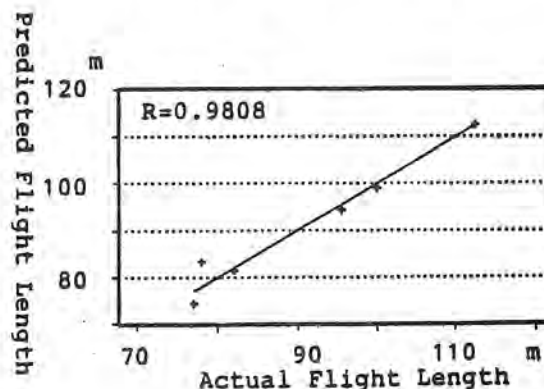


Fig. 2 : The result of multiple regression analysis. Y-axis shows predicted flight length. X-axis shows actual flight length.

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THE DEVELOPMENT OF THE ISOKINETIC MUSCLE TORQUE IN THE DORSIFLEXORS OF THE ANKLE IN CHILDREN 6-15 YEARS OF AGE.

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Muscle force in children is correlated with age, height, bodyweight and sex. The aim of this study was to investigate the development of torque in growing children. The muscle group tested was the dorsiflexors of the ankle.

Material.

137 children, 71 girls and 66 boys, participated in the study. The children were randomly chosen from the register of the population. None of the children had any neuromuscular disease. Informed consent was obtained from the children as well as from their parents.

The children were grouped into four age groups with mean ages of 72 ± 2 , 109 ± 2 , 144 ± 2 and 181 ± 2 months (mean $\pm$ SD). Anthropometric measurements included height, weight, circumference of the calf and length of the foot and the lower leg on both sides.

Method.

The isokinetic strength of the dorsiflexors of the ankle was measured with the Cybex II dynamometer (Lumex Inc., Bayshore, New York). The children were sitting in a standardized position with the knees flexed 90 degrees, fastened and stabilized with padded straps. The axis of the dynamometer was aligned with the talocrural axis. The method used is presented elsewhere (1). The active range of motion was controlled by a level indicator. Strength was tested at six different speeds, the order of which was randomized; i.e. 60, 15, 120, 30, 180 and 240 /s and isometrically. The peak torque, peak torque angle, range of motion and work, i.e. area under the torque curve, at each velocity on the left and right side were measured.

Statistics.

Student's t-test for paired and unpaired observations was used. The influence of the anthropometric variables upon the developed force was assessed by regression analysis.

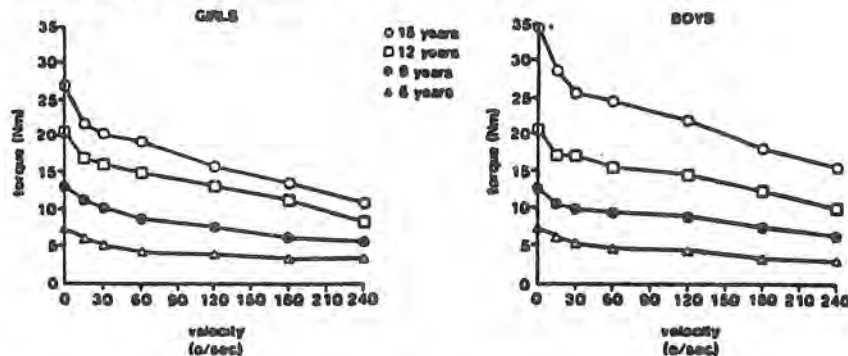
Results.

Peak torque and work.

As expected, the maximal developed peak torques increased with age. The developed torque was significantly higher in boys than in girls at 9 years of age at the velocities 180 /s and 240 /s on the dominant side. In the 15 year old group, the boys developed significantly higher torques than the girls at all velocities on the non dominant side and at all but 60 /s at the dominant. In the other age groups there were no significant sex differences. The developed torque, generally, increased with increasing weight and height. However, the developed torque/kg bodyweight was not significantly different in the 15-year-old girls compared to the 12-year-old girls. The torque-velocity relation in different ages is shown in Fig 1.

Regression analysis showed that the correlation between anthropometric data and developed muscle torque was almost equally influenced by age, weight and height. The other measured anthropometric measures did not improve the correlation. The inclination of the regression line got more negative with age in both boys and girls.

Fig 1. Torque-velocity relation in different ages.



The work developed, i.e. the area under the torque curve, showed that children 6- and 9-years-old had a significantly greater work area at the highest velocities on the dominant side. The work differed between the sexes only in the 15-year-old group, the boys being stronger than the girls. In the 15-year-olds, the work area was significantly greater at the lowest velocities on the dominant side. The same tendency was found in the developed dorsiflexor torque.

Range of motion and peak torque angle.

The youngest children had a range of motion of 101 ± 13 degrees while the oldest had a range of motion of 87 ± 14 degrees. The difference was significant. The values of the 9-year-olds and 12-year-olds were in between. The mean peak torque angle for the different velocities decreased significantly from 42 ± 14 degrees plantar-flexion at 6 years to 25 ± 8 degrees plantar-flexion (2) at 15 years of age. No sex differences or side differences were found in children of the same age.

Reproducibility.

Ten 6-year-old children, 5 girls and 5 boys, were tested twice with a test interval of less than two weeks. The coefficient of variation was ca 17 %.

DISCUSSION

The force-velocity curve shows that the boys have their most intensive force development between 12 and 15 years while this occurs earlier in girls. The muscle force around the ankle joint is different on the dominant and the non-dominant side as shown in adults (1). In this study the younger children showed higher torque values and developed work on the dominant side at higher velocities while in the older children the non-dominant side was stronger at slower velocities. It seems that the boys have their most intensive force development between 12 and 15 years while this occurs earlier in girls. The range of motion was decreasing with age indicating stiffer soft tissue around the joint. This is in line with the findings of Bosco and Komi (3) who compared jump height with different prestretch conditions and found that children cannot use the influence of the prestretch cycle as well as adults. It seems that the strength is greater in the dominant side in the younger children and in the non-dominant side in the older children.

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3D MEASUREMENT OF REARFOOT MOTION DURING RUNNING

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It has been claimed that an erroneous motion pattern during running may lead to injury to the lower extremity. The motion pattern has so far almost exclusively been studied with two-dimensional models.

The purpose with this study has been to develop a method to measure motion of the lower leg and rearfoot in three dimensions, during running on a treadmill.

With the computerbased optical measurement system SELSPOT it is possible to detect the three-dimensional position of infrared LED markers. With three markers on each segment studied (lower leg and rearfoot) the positions and orientations of these segments may be determined unambiguously. The test setup is shown below.

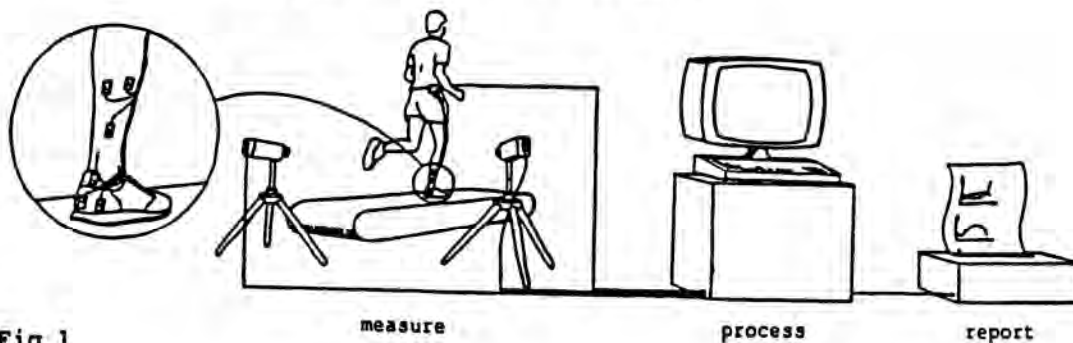


Fig 1

One coordinate system is associated with the markers on the rearfoot and one with those on the lower leg. A calibration routine will make sure that these coordinate systems will get a predetermined orientation in respective segment. The sampling frequency may under these circumstances (two cameras and six markers) be 830 Hz.

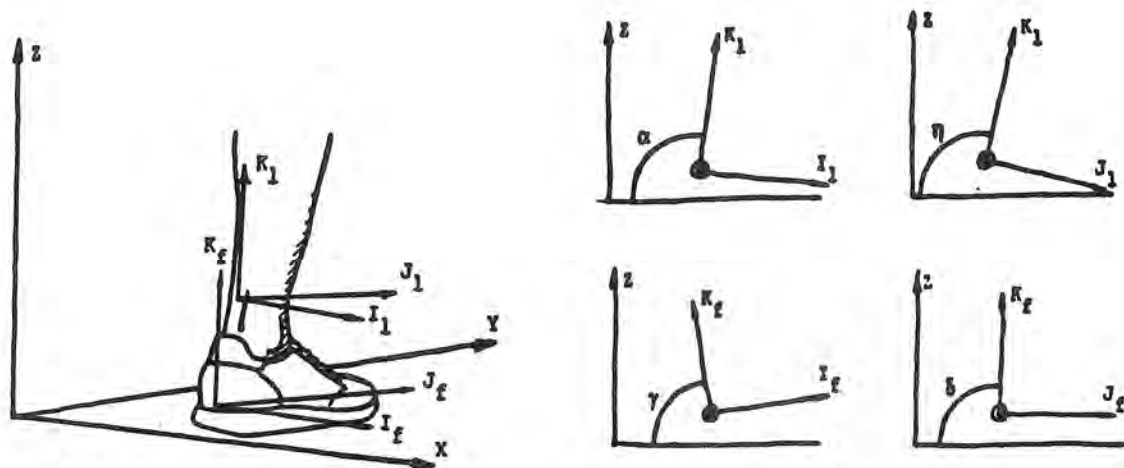


Fig 2

Our first approach has been to determine the rotations between these coordinate systems and between respective coordinate system and the room fixed system, instead of the rotations about the real rotation axes of the foot. Some of these angles have a traditional definition used by

clinicians but some have never got an accepted three-dimensional definition before. These angles can be defined in several ways but which definition we choose is mostly a matter of convenience. Each measurement may be long enough to contain several steps. The steps are separated in the software and the results printed in form of a single step plot as in figure 3, or as a printed report containing max and min values of interest for single steps as well as means and standard deviations.

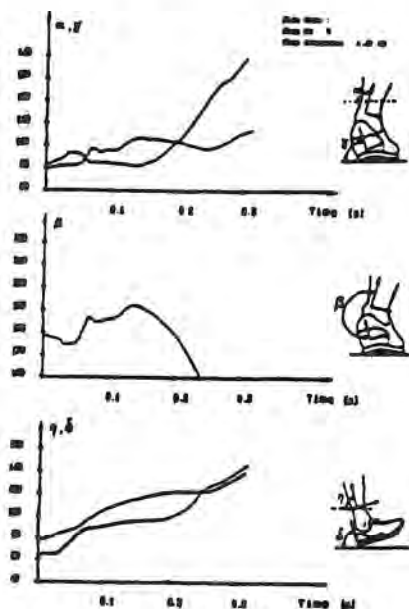


Fig 3

The three-dimensional method to measure motion in the lower extremity have some advantages compared to the two dimensional as the rotations are measured around body segment axes instead of around room fixed axes. The natural step to take in the future is to develop a method to calculate the rotations about the true motion axes of the foot.

ELECTROMYOGRAPHIC SIGNS OF FATIGUE DURING DYNAMIC AND STATIC KNEE EXTENSIONS.

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INTRODUCTION: Several studies of repeated isokinetic knee extensions have shown an initial steep decrease in mechanical output and mean power frequency of the EMG, followed by stable levels. It has been suggested that such decreases are due to fatigue of the fast twitch motor units. The present preliminary study aimed to analyse the relationship between mean power frequency of the EMG and mechanical output in a highly fatiguing situation when most of the fast twitch motor units already were fatigued.

SUBJECTS AND METHODS: Four clinically healthy (untrained) subjects (25-35 years) volunteered for the study. The subjects performed 70 repeated maximum isokinetic knee extensions (range of motion: 90 degrees). Immediately thereafter (without any rest) they performed a maximum static knee extension (45 degrees of flexion) until exhaustion. EMG signals were registered using surface electrodes, from the three superficial bulges of the quadriceps femoris. A programme on a computer determined the mean power frequency (Hz) and the root mean square (RMS) value of the EMG signals and peak torque for each dynamic contraction. The same variables were also determined for the 5 initial seconds and the 5 terminal seconds of the static contraction.

RESULTS: Peak torque decreased 34% during the 70 dynamic contractions. The decrease was steepest during the initial 30 contractions followed by a less prominent decrease (figure 1). Also the mean power frequency of the 3 EMG signals decreased during the initial 30 contractions. The decrease was most prominent in the rectus femoris muscle (37%), (figure 2).

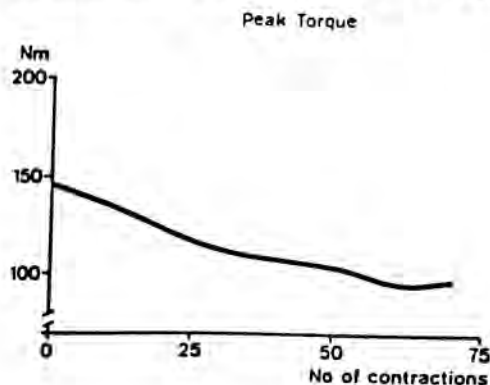


Figure 1: Peak torque.

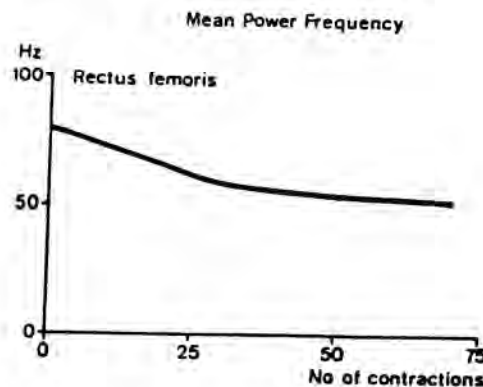


Figure 2: Mean power frequency of the rectus femoris.

The mean power frequency of the rectus femoris had a higher degree of correlation with the mechanical output ($r=0.83-0.87$) than the two other muscles; corresponding values for vastus medialis and vastus lateralis were $r=0.15-0.63$ and $r=0.50-0.76$, respectively.

Torque decreased 20% during the static contraction (figure 3). In contrast no distinct decrease in mean power frequency of the three muscles occurred as exemplified for the rectus femoris muscle in figure 4.

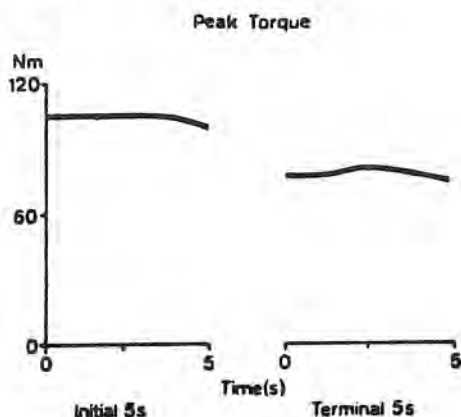


Figure 3: Torque.

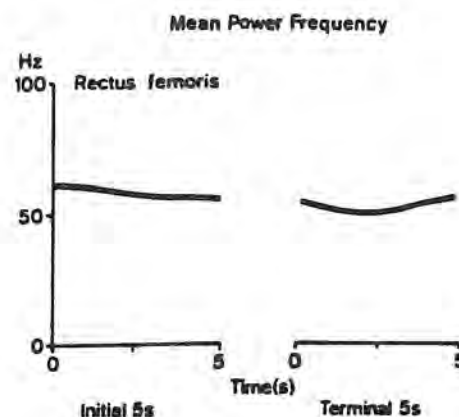


Figure 4: Mean power frequency of the rectus femoris.

CONCLUSION: The decrease in mechanical output during the dynamic contractions appeared to be composed of 2 parts: an initial part with a steep decrease in mean power frequency followed by a part with less or none decrease in mean power frequency. Similar results have been obtained recently in studies of knee extensors of orienteers and in plantar flexors of sedentary subjects (Gerdle et al., 1987, 1988).

An important finding of the present preliminary study was that the decrease in mechanical output during the static contraction was not accompanied by distinct decreases in mean power frequency. We have earlier suggested that the steep initial decrease during dynamic repeated knee extensions is due to fatigue of fast twitch motor units. The question that arises from the present study is if the mean power frequency variable really is an indicator of fatigue of slow twitch motor units.

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FORCE-MEASUREMENT IN COMPETITIVE CYCLING WITH A MOBILE EQUIPMENT
(G.Bergmaier, F.Hediger and M.Märki; Zürich / Switzerland)

Introduction

Until today there has been a deficit of biomechanical research in competitive cycling and the control of the technomotoric component has been banished into laboratories - a fact which is too restrictive for optimizing, e.g. pedaling technic, style, pedaling efficiency, choice of new materials or position for competitive cyclists and also for our practice-orientated biomechanical further investigations outdoors on track and road.

Measuring Device

So we have developed a mobile, small, robust, light and inexpensive measuring device that picks up torques, makes "force" directly visible on a handlebar-LCD-display to the cyclist (Bio-feedback) and sends these data directly to the nearby measurement stations (up to 200 m) for further processing.

The measuring device on the bicycle can be remounted relatively easily and is not so sensitive as regards temperature, humidity and vibration. Strain gauges on both crankarms (torque), inductive approach-switches (64 impulses per revolution for the angle), Reed-contacts for resets and the telemetric transmission (Multiplex, 40 MHz FM, width-modulated impulses, processed by digital logic-circuits) seem to be the most appropriate possibility. Our choice of signal-recording and signal-transmission gives a good basis for a many-sided storing and analysing not attached to the bicycle.

At the moment we prefer a handlebar-LCD-display (and a special logic circuit) for the cyclist with the integrated forces and the minimum and maximum-peaks of the effective propulsion force on both cranks.

Tests and practical use

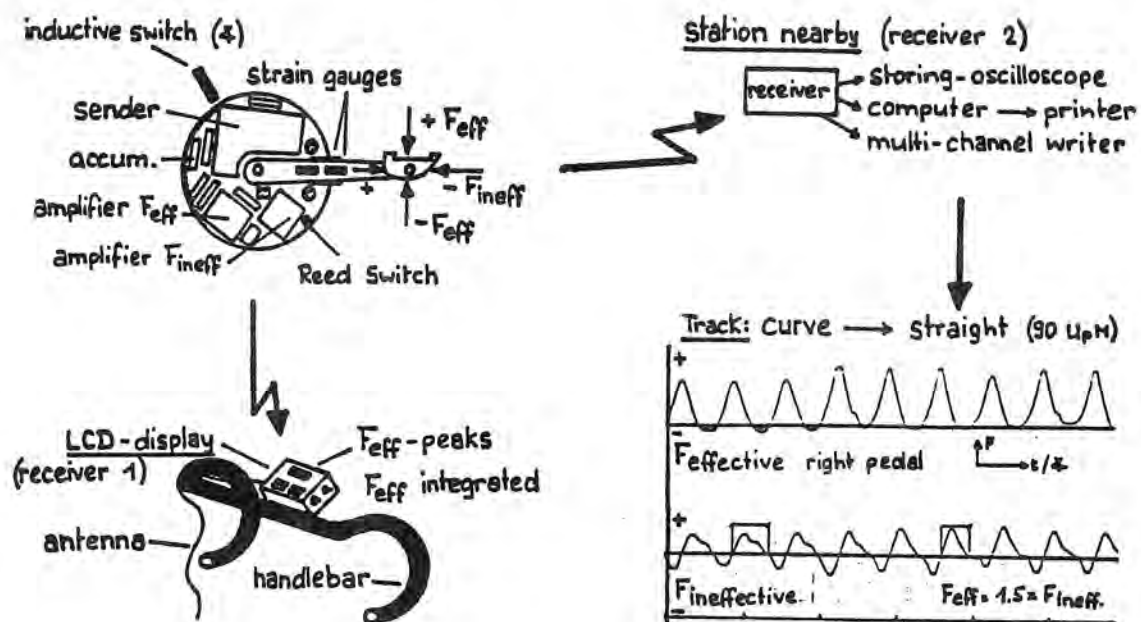
The gauge-measurements in the laboratory confirmed an exact linear connection between the applied force and the sender input signal. The same is the case in the LCD-indicators after the processed signals. The measurement after 13 hours of continual research deviated only 3 mV from the data at the beginning, and at a temperature difference of 40 degrees C only 10 mV.

Tests and trainings of Swiss elite and pro racers, on track and road, have confirmed the efficiency of our equipment and (as a first aim) its help for increasing personal performance.

Irregular signals appeared only after 3 hours when the accumulators had been discharged. Even the strongest vibrations had caused no damage. The room needed for the measuring equipment was considered to be little with its size of a chainring and did not disturb the test-persons at all.

The recording of values and functional curves of the effective and ineffective components of the used force throughout the pedal stroke with multi-channel-printer and computer proved to be very effective. Especially a storing-oscilloscope and computer designed curves proved to be fast and useful in normal training, although the accuracy is limited for absolute values.

Diagrammes such as the one below (and the trainers instructions based on them) helped the respective racing cyclist to achieve a better technique after some training, which of course has to be practised and consolidated for a long time.



Discussion and future

The next handlebar display (under construction) will be reduced in size with the help of a microprocessor which operates a LCD-screen. In its present form it is of course useful, but not yet the best for the cyclist. An acoustic indicator with headphones for both crankarms will soon be developed so that in a training on the road the cyclist's attention does not get distracted from the traffic too much.

Besides the bicycle, also an electronic circuit is being developed for the integration of the operating forces over a certain period of time and distance to make the planned biomechanical experiments and individual material tests in connection with optimal length of the crankarms, power-cranks, position, oval chainrings, best gears and pedaling frequency of the discipline.

MODELING AND SIMULATION OF HUMAN GAIT IN THREE DIMENSIONS WITH MULTIBOND GRAPHS.

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Introduction

The human being can be considered as a neuromusculoskeletal system. For the modeling of such large systems with spatial (3D) motions in which different physical domains are involved, multibond graphs are useful. In this paper multibond graphs are applied to the skeletal part of the model in behalf of the simulation of human gait. The systematic method of modeling with bond graphs [1] starts with a description of the physical system in terms of ideal elements (e.g. inertias, springs, transformers, etc.) and their connection structure. This so-called Ideal Physical Model is then depicted into a graph: a (multi)bond graph. The basic element of the connection or 'junction structure' in the (multi)bond graph is the bond, depicted by a half arrow, which represents the energetic relation (power) between the elements connected by the bond. Multibond graphs are a generalisation of the single bond graph notation, which is comparable to the generalisation from scalars to vectors and used to describe multidimensional systems. After computational causality is assigned to the relations in the graph, a set of equations is derived, which describes the dynamics of the system. The model can be simulated after addition of an adequate integration algorithm to the equations and when the input data, parameter values and initial conditions are available.

Modeling

The skeleton is approximated by a system of rigid bodies interconnected by joints. For a simple model of human gait the skeleton is divided into seven bodies: two feet, two shanks, two thighs and one upperbody. Each joint is a single connection point between two bodies and has only rotational degrees of freedom. Floor contact is approximated as just another spherical joint to a body (earth), which has preserved motion i.e. zero velocity. However the floor contact is a discontinuous connection (dependent on the stance or swing phase of the respective leg) and during contact the connection point shifts along the foot. The connection point coincides with the center of support, which is the point where the resultant of the ground reaction forces applies. In the joints compliance and damping properties can be assigned. The resultant of torques, caused by active (muscle) forces, are applied to the bodies via the joints. Body fixed coordinate frames are chosen in the center of mass in each body along the principal axes of inertia. The diagonal inertia matrix described in a body-fixed frame, is constant. Due to the rotational velocity of the body with respect to the inertial frame, gyroscopic torques arise and are taken into account. Euler parameters are used in the calculation of coordinate transformation matrices. In this way singularity problems and trigonometric function evaluations associated with the use of Bryant or Euler angles, are circumvented and the equations can be solved computationally more efficient [2]. Due to the kinematic constraints between the bodies, the inertias are not causally independent. Different methods can be used to derive a set of explicit differential equations, which can be solved by explicit integration routines [3]. Instead, a set of implicit differential and algebraic equations is derived from the multibond graph, which can be solved with a technique used to solve ordinary differential equations [4]. This technique consists of an implicit integration routine (e.g. GEAR) combined with iteration routines (e.g. Newton-Raphson) and has better numerical

stability than explicit integration routines. A variable step and order algorithm is formulated to adjust automatically the time step to changes in the dynamic behaviour of the physical problem during simulation. Dependent on whether compliance and damping properties are assumed in the connection to the floor or not, a Lagrange Multiplier is needed to solve the closed kinematical loop between the two floor contact points during the double support phase. In either case additional implicit equations are derived. All derived equations can be solved with the same numerical technique. In order to deal with the discontinuity caused by the floor contact, different models will be formulated for the different phases in gait. The switch from one model to another will be based on the height of the feet above the floor.

Simulation

As input data to the models, resultant torques and data on the shift of the center of support for both feet are necessary. At this moment, these data as well as parameter values and initial positions and velocities are not available. The presented modeling and numerical techniques have been evaluated on testproblems (for more details, contact authors). Therefore, the following is an outline of the research in the near future, which will not only present the necessary data but will yield a validation possibility of the model as well. The parameter values concerned with inertias and positions of connection points or centers of mass can be derived from anthropometric measurements and data from literature [5]. Parameter values concerned with compliance and damping values of the joints can be estimated or derived from literature [e.g. 6]. From position, velocity and acceleration data of minimal three points on each body, the coordinate transformation matrices between bodies and angular velocities and (angular) accelerations of the bodies can be calculated [8]. From these data the resultant torques can be calculated with the use of the already formulated models only with different causality. At this stage it is possible to adjust parameter values by comparison of measured and calculated ground reaction forces. Simulation of the models with the calculated resultant torques and the measured data of the center of support will result in a prediction of the human gait pattern as was measured. After this validation process, simulation runs can be made to investigate the influence of changes in for example resultant torques or parameter values on the gait pattern.

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TELEMETRIC IN-VIVO MEASUREMENT OF THE PRESSURE CONDITIONS IN THE INTERFACE OF AN IMPLANTED ACETABULUM PROSTHESIS

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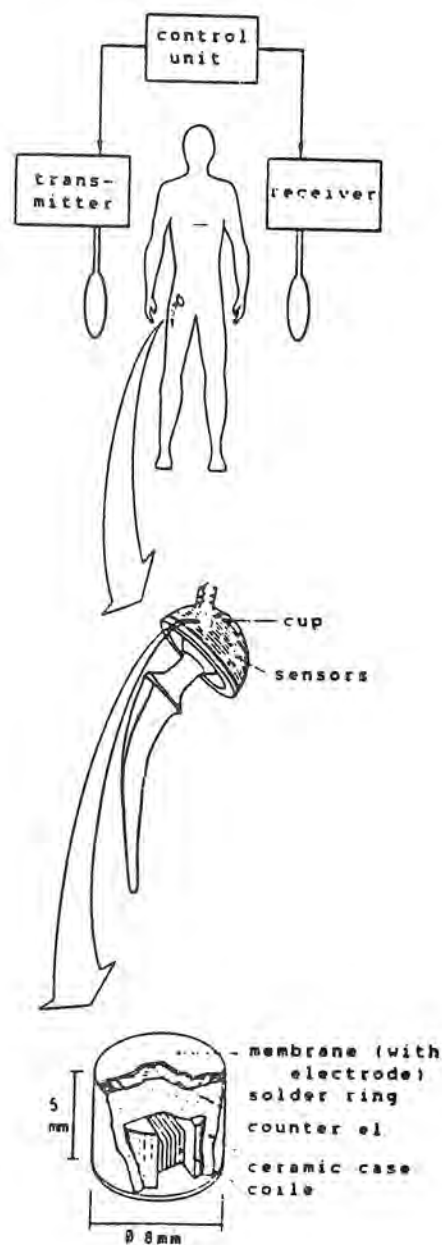
The postoperative conditions in the interface of an acetabulum prosthesis and the pelvis bone are important determining the success of the implantation. In order to allow for a comparison of several different fixation methods, a telemetric procedure has been developed at the Institute of Physics in Basle to measure pressures directly active on the cup surface.

Our probes (5mm thick, 8mm diameter) consist of a simple oscillating circuit made up of a small coil with a ferrit core, a parallel chip capacitor for frequency adjustment, and the actual sensor.

The sensor functions according to the same principle as a plate capacitor. One of its plates curves in upon pressure which reduces the space between it and the counter electrode. This increases its capacity, leading to a change in the resonance frequency of the entire circuit. Thus, the resonance frequency becomes a direct function of the pressure active on the probe.

The sensor is constructed by soldering two Al_2O_3 plates together after applying a thin conducting layer to each one. The solder consists of an amorphous CuTi foil and a CO_2 laser provides the energy source. Since the coil and the parallel capacitor must be protected against the enormous pressures produced, they are placed in a ceramic case that is put on the sensor.

In order to activate the circuit an external electromagnetic field of variable frequency is applied by means of a conductor loop fastened to the patient. A second conductor loop receives the signals that the probe emits when the activating field reaches the momentary resonance frequency of the probe. As soon as the resonance signal of a



probe is detected, the electronic control unit ensures the permanent regulation of the transmission frequency in such a way the pressure dependent resonance is followed. In order to filter out this comparatively weak signal from the fairly loud background noises, electronic controlling devices are employed that will not be dwelt on in this context. Our apparatus is able to execute about 130 measurements per second. The data are directly processed by a personal computer.

Now that the probes and measuring devices have been developed and all tests have produced positive results, work is being done on an implantable cup with 5 probes that will probably come into use this winter.

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IN VIVO STRESS RELAXATION OF THE LOADED FOOT

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In order to understand the loading of the human foot better a test method has been developed to measure the stress relaxation as seen between the lateral femoral condyle and a force platform. This is a preliminary evaluation of this method through screening tests on four subjects under various conditions. Since it is understood that the twenty five bone load frame distributes forces among bones, soft tissue, ligaments and fascia it is further desired to eventually isolate the contributions of each. Although force relaxation is being measured, it is an indication of the energy absorption and dynamic properties of the tissues.

Unpublished industrial research has shown that material stress relaxation can be modeled by taking the time rate of change of force, after a step strain, to be directly proportional to the peak force and inversely to the time. Integration of this first order differential equation yields:

$$\text{Force} = \text{Peak Force} \times (1 - \text{Decay Constant} \times \text{Ln}(\text{Time in seconds}))$$

In 1986 it was found that this equation also fits stress relaxation in the foot tissues to a very high degree of correlation. This paper concentrates on the method of accurately determining the decay constant.

METHODS - An adjustable height crosshead mounted over a force platform allows a vertical step strain to be placed on the knee through a manual lever driven plunger. In these tests strains from 5 to 10 mm produced forces in the range of 200 to 800 N with an average of about 450 N.

The protocols needed for repeatable data are subtle but easily mastered. The bare condyle can stand very little force without pain so the subject first was sat with knee under plunger and the leg adjusted for a vertical tibia. The exact position of the foot on the platform was marked with lines for heel and lateral border. Trial forces of the plunger onto a circular knee molded plate were made until it was certain that alignments were correct. Then a small amount of flexible dental material was placed right on the condyle, the plate replaced and a small load applied until the material had set. All parts were marked and referenced to leg. Crosshead heights and lever stops were measured and maintained for each subject. A firm smooth strain monitored by an LVDT was then applied in less than a second and recorded at one hertz for 70 seconds. If the force is applied in a jolt there is always an overshoot of the peak from the curve fit. These initial screening tests were applied in separate sessions of multiple relaxations about three minutes apart. Just the general levels, trends and problems with the method were sought.

RESULTS - Thirty nine separate relaxation tests were run. The average decay constant was 0.08 and the average force 450 N. The total data shows a trend of +0.013 (or 16%) / 100 N but its meaning is not clear at this time. The test sequence for flat relaxed feet is shown in the table.

<u>Subject</u>	<u>Day</u>	<u>Force Range (N)</u>	<u>Decay Constant Trials (x 1000)</u>				
One	One	322-396	64	74	77	57	47
	"	277-307	83	57	62		
	"	259-456	97	69	63	61	56
	Two	698-797	98	101	95	92	
	Four	482-687	78	84	72	75	
Two	One	647-1037	102	112	98	85	77
	Two	560-639	105	95	101	103	
Three	Two	351-441	66	56	66	59	
Four	Four	199-236	73	69	92		

Several sequences show a decline in decay constant with the repetition but others don't. Although it is expected that the fascia will be stretched and aligned much more data is needed to see if this decline is real.

The slow data acquisition allowed only the average stiffness to be calculated. They ranged from about 200 to 1000 N/cm the same range as known heel pad stiffness. Since the force displacement curve increases exponentially this is only a tangent stiffness. Some tests were run at 10X the acquisition speed to determine this. Only one has been reviewed. It showed about 400 N/cm in the first 4 mm and 800 N/cm in the next 4 mm.

It is noteworthy that in a separate sequence with subject three where the 1st and 2nd toes were extended about 40 degrees with a plug set under them the stiffness of the foot doubled over the flat foot. The decay constants showed no effect, however, and were in the same range as in the table.

DISCUSSION - It is to be expected that as the method is used, streamlined and analyzed with the introduction of in vitro tests that the meaning of results will be enhanced. Especially important is the partitioning of the various elements in the process. Since the data is more constant at higher loads, around body weight, this will be the range used provided the subject can stand the knee loading.

It is hoped that the long time relaxation decay can be extrapolated to the short times of foot strike to aid in the determination of tissue overuse injury, restoration and training effects. Obviously the test method, once made routine and more reliable, can be run before and after intensive activity of various types and even in a longitudinal sense during a training period.

A DIREKT MEASUREMENT OF THE DEFORMATION OF A RUNNING SHOE MIDSOLE AND OF THE HEEL PAD AT HEEL STRIKE DURING A RUNNING STEP.

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The knowledge of the deformation, which occurs at different levels in the ground-shoe-heel system, is essential for a better understanding of the shock-absorbing mechanism at heel strike during running.

Material tests in which a weight is dropped on the sole, can result in a precise measure of the deformation of this part of the shoe. But such tests separate the shoe from the foot and therefore don't give an exact reproduction of what really happens during heel strike. So preference is given to tests with subjects, although the deformation of the sole or the heel pad during heel strike is difficult to determine. The compression of the total heel region is assessed from the downward movement of a marker on the lateral malleolus. Two studies report on the deformation characteristics of the heel pad by an impact method.

The purpose of our study is to establish a method to determine *in vivo* and separately for midsole and heel pad the deformation characteristics at heel strike during running. Therefore, a high-speed (150 frames/s) 35 mm X-ray film-method is developed. On the obtained pictures the thickness of the midsole and of the heel pad can exactly be measured every 6.66 msec of the heel strike.

For safety the use of a low local radiation is a *conditio sine qua non*. In our experiment the total radiation for each test person has been much less than the safety factor for public persons (: factor 10).

The volunteer test persons are two well trained runners (heelstrikers). In synchronization with the X-ray film groundreaction forces are measured with a Kistler force plate (1020s-1) and the total movement is laterally filmed at a speed of 200 frames/s with a Photosonics high-speed camera.

The limited space around the X-ray film-equipment has forced us to use a jumped running step. The heel strike characteristics of this movement have to be similar to those of the actual running motion. Therefore the kinematic and kinetic data obtained during X-ray tests are compared with data of natural running. These data are acquired while the same test persons have been running on an outside track with a foot contact on a built-in force plate. The shoe conditions have been kept constant and the running speed has been 3.6 ± 0.2 m/s and 5.0 ± 0.2 m/s.

The kinematic data of the jumped running step (8 trials) just before ground contact are : Knee angle $159.5 \pm 6.5^\circ$, Angle of the shoe sole $23.6 \pm 4.0^\circ$, Vertical velocity of heel $V_{z0} = -1.3 \pm 0.1$ m/s. The kinetic data of the jumped running steps, taken with z_0 the force plate, are as follows : Vertical impact force peak $F_{z1} = 2.3 \pm 0.3$ BW, Time of impact peak $t_{i1} = 21.5 \pm 2.5$ ms. All these data reveal a good resemblance with the heel strike pattern during actual running.

With the X-ray source on the medial side of the landing foot, a cross section view in the sagittal plane is obtained on the lateral side of the rear part of this foot (i.e. 33% of the shoe length; Fig. 1).

Analysis is done by means of an interactive computer program, using a Housron Instruments digitizer coupled to an IBM AT. After scaling one of the frames of a series constituting a single strike, contour lines of the calcaneus and sole can be digitized continuously. The program automatically selects the zone of interest and performs an equally spacing procedure (the 33% zone mentioned above becomes divided in 50 equal segments; see fig. 1). Thus, the deformation at specified locations can easily be followed throughout the strike.

For each frame, the program gives, besides the graphical output (see fig. 2), a list of the thickness and absolute and procentual deformation of each segment, and integrates the deformation for the total, or a (user delimited) part of the digitalization of the sole and heel pad.

Our force-deformation curves of midsole compression are of a surprising shape compared with the force-deformation curves obtained with impact testing machines. The important role of the increasing contact area ground-sole due to pronation and plantar flexion of the foot can be gathered from our results. The force-deformation curves of heel pad compression show a hysteresis loop which illustrates the shock-absorbing qualities of the subcalcaneal fat pad.

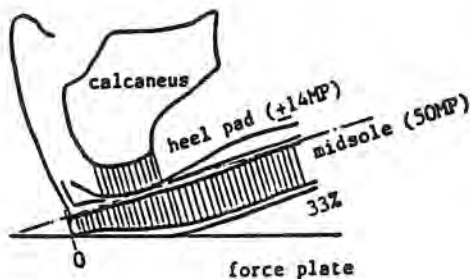
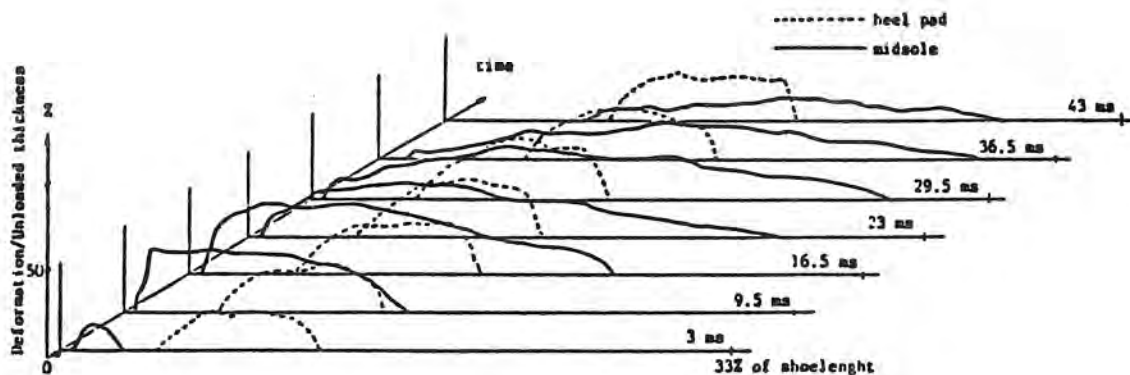


Fig. 1 : X-ray picture of the landing foot and measuring indications.

Fig. 2 : shows the procentual deformation of midsole and heel pad during heel strike.



CONTRIBUTION TO THE STABILITY ANALYSIS OF INTACT AND INJURED LUMBOSACRAL SPINAL SEGMENTS

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Introduction. Despite the general recognition of the significant role of the muscles in preventing and treating low back disorders, little is known about the mechanisms by which muscles provide segmental stability to the human spinal column. The biomechanical function of counter-balancing external physiological loads and achieving spinal posture and movement can be assigned to the large superficial muscles. Due to their morphology, i. e. relatively small cross-sections and locations closer to the spinal centers of rotation, intersegmental muscles cannot contribute much to the tasks requiring large bending moments. It may be reasoned that they play an important role in maintaining segmental stability. Besides these general observations, the stabilizing capabilities of these muscle groups have only been analysed recently by means of an in vitro study [1]. In the present work, a detailed qualitative and quantitative parameter study of the intersegmental muscle stabilisation is carried out by the use of a 2-D nonlinear finite element model.

Methods. The analysis utilizes a two-dimensional sagittal-symmetric finite element model developed recently for mono- and multi-segmental calculations. It is based on 15 springs representing an intervertebral joint and accounts for large lumbosacral deformations and nonlinear soft tissue material characteristics. Detailed morphological and biomechanical studies have been done to provide this mathematical model with an appropriate set of spinal data. The loadings consisted of axial compression acting alone or in combination with flexion/extension moments and anterior/posterior shear. The loads were applied incrementally to reach their maximum values. Additional load consisted of an intersegmental muscle force vector, which varied with magnitudes of 20 (M20), 40 (M40) AND 60 (M60) newtons, applied to the center of the spinous process. The orientation of the muscle force vector was varied from 360 degrees in the sagittal plane. The computed results were: load-displacement curves, ranges of motion and neutral zone. The neutral zone was defined as the spinal motion elicited by a load equal to 10% of the maximum moment. Besides the intact (INJ.0), three injuries were studied: INJ.1. Annular injury and nucleus removal; INJ.2. Transection of supra- and interspinous ligaments; and INJ.3. Transection of supraspinous, interspinous, and capsular ligaments and ligament flavum. The resulting data were further analysed to observe relationships between the biomechanical parameters and the stabilizing mechanisms of intersegmental muscles.

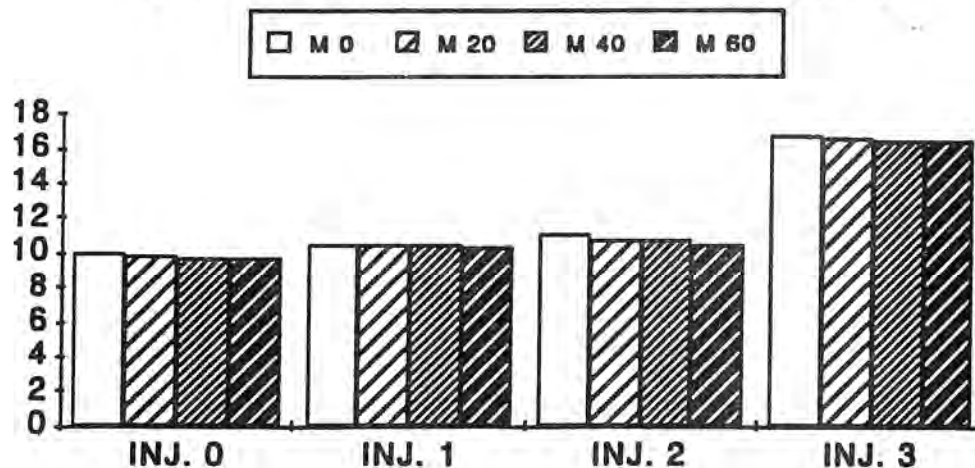
Results. In general we found increased neutral zone and range of motion with increasing injury. Applying the intersegmental muscles forces did not significantly affect the range of motion (Figure 1) but decreased the neutral zone (Figure 2). Results in the two figures are due to the muscle force vectors oriented at 135 degrees to the mid-disc plane. The optimal values of the muscle force vector (magnitude and orientation), considering the intersegmental stability, were found to be in good agreement with the physiological and morphological properties of the actual muscles. In particular, increase of muscle activity beyond a certain limit did not improve the intersegmental stability.

Conclusions. Based upon the results of our mathematical model and the earlier experimental analysis [1], the following conclusions may be reached. In contrast to the neutral zone, there was no significant effect of muscle forces on the ranges of motion. Therefore, we believe that the neutral zone should be preferred as an indicator of spinal stability/instability. It is hoped that further research may provide a methodology to measure the neutral zone in vivo. The results suggest that intersegmental muscles play an important role in fine tuning the stability of the intact spinal segments and have the potential for stabilizing the spine after injury.

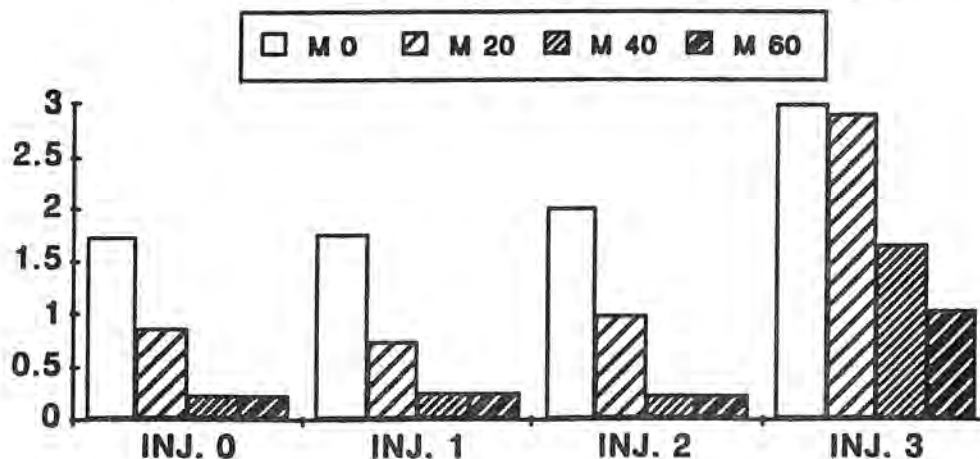
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Acknowledgements: Supported by DFG grant NO 164/2-1 and NIH grants AR 30361 and AR 34699.

RANGES OF MOTION EFFECT OF INJURIES AND MUSCLE FORCES



NEUTRAL ZONES EFFECT OF INJURIES AND MUSCLE FORCES



THREE-DIMENSIONAL MOVEMENTS OF THE WHOLE LUMBAR SPINE

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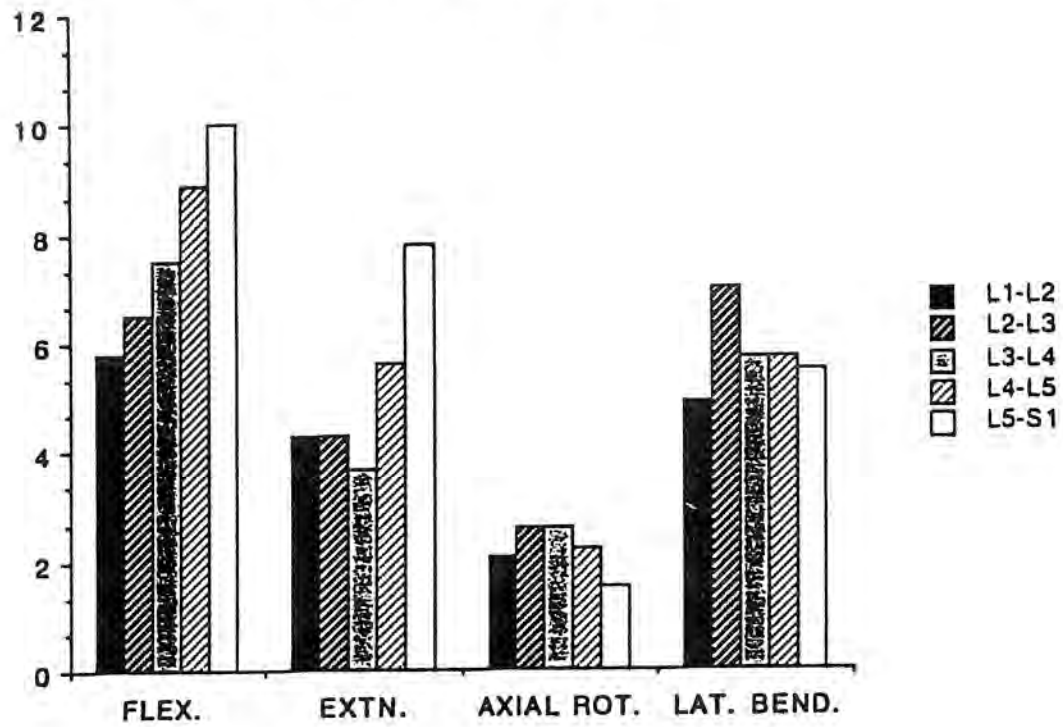
INTRODUCTION. The lumbar spine is an important region of the spine, often involved in trauma, degeneration and disease. Knowledge of the normal movements and biomechanical characteristics of this region are necessary for evaluating clinically pathological conditions that may produce spinal instability. Although physical characteristics of the lumbar functional spinal units have been obtained, the biomechanics of whole lumbar spine is less well known. The purpose of this invitro study was to quantitatively determine three dimensional movements of the whole human lumbar spine.

METHODS. Ten fresh cadaveric whole lumbar spine specimens six from L1 to sacrum and four from L1 to ilium were used. The sacrum or ilium was embedded in an epoxy block to provide fixation to the test table and to align the centers of L2 and L5 in a vertical direction. The upper vertebra (L1) was provided with another epoxy block which included a loading jig. To load the specimen at the upper vertebra we utilized six pure moments : flexion (+MX), extension (-MX), bilateral axial rotation (+MY) and bilateral lateral bending (+MZ). The maximum moment was 10 newton meters (N.m), reached in five equal steps. Each moment was applied individually and in a three load/unload cycles to minimize the viscoelastic effects of the specimens. On the third load cycle readings were taken of the vertebral motion at each load-step, after allowing 30 seconds of creep to occur. The relative motions of each vertebra were measured by stereo-photogrammetry, providing six degree-of-freedom measurements. Three motion parameters were defined on the third load cycle. Neutral zone (NZ) was the displacement at the zero-moment point measured from the neutral position ; elastic zone (EZ) was the displacement from the zero-moment point to the maximum-moment point ; and the range of motion (ROM) was the sum of the NZ and EZ. This technique allowed free and natural movement of the spine.

RESULTS. Although all the six degrees-of-freedom were measured, only the main motions (rotations in the directions of the applied moments) in degrees are given. Since the motions were about equal for the left and right lateral bending ; and for the left and right axial rotation, for these motions average results are given.

CONCLUSIONS. In flexion (+MX) and extension (-MX), more intervertebral motion takes place at L4/5, L5/S1 than at the upper levels. In axial rotation (MY), least motion takes place at L5/S1. In lateral bending (MZ), least motion takes place at L1/2 and biggest motion takes place at L2/3, while similar magnitudes of motion are seen at L3/4, L4/5 and L5/S1.

RANGES OF MOTION HUMAN LUMBAR SPINE



EFFECT OF KNOWLEDGE OF THE LOAD ON LIFTING TECHNIQUE.

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Low back pain secondary to lifting continues to be a major problem in industry, and in management of patients with back problems. The effects of many factors on lifting technique have been studied, but the effects of balance have seldom been considered.

Previous work in our lab indicated that in many lifts the center of mass of the combined lifter and load was anterior to the toes at the time the load cleared the floor. For a subject not to fall forward under these circumstances, the laws of dynamic equilibrium require that the center of mass must have an upward and a posterior component of velocity.

As time goes on, the upward velocity will be slowed and reversed by the action of gravity. The posterior velocity must be great enough that the center of mass will arrive over the feet before gravity overcomes the upward velocity.

The hypothesis is that lifting techniques are different when subjects do and do not know the weights they are lifting. The differences arise because the subject does not know how much velocity to build up if he does not know the weight. The purpose of this study is to test the hypothesis.

Ten healthy male subjects, ages 20-40, with no history of back pain or trauma and normal physical findings participated in the study. Each subject lifted a 20 N weight box from the floor to a shelf set at knuckle height directly in front of him. The box had two fixed handles 27 cm above the bottom and contained 300 N, 250 N, 150 N, or no additional weight. The subject could not see into the box. The foot and load positions were constant throughout the study, with the load always positioned 37 cm in front of the subject's feet. The lift was performed as sagittally symmetrical as possible, but otherwise the subject was given no instructions as to the technique of lifting.

Each weight was lifted twice; one time the subject was told the weight before lifting, the other time he was not told the weight. The order of the weight lifts was random.

Motion was recorded using an optoelectronic system (Motion Analysis), and ground reaction force was recorded using a Kistler force plate. The data were analyzed using a rigid 5 link model to calculate moments, velocities, and accelerations.

Paired t tests were used for statistical evaluation.

At the 5% level, statistically significant differences at 0 kg. load were found in the following parameters: peak moments at lumbosacral articulation and at the knee, and peak velocity of the acromion and the hand. At 15 kg. only the peak acromion velocity showed a statistically significant difference, smaller than at 0 kg. For each parameter, the mean value was greater when the load was not known.

At 25 kg. no statistically significant differences were found. At 30 kg. only the difference in peak acromion acceleration was statistically significant. However, in this case, the value was greater when the load was known.

PARAMETER	KNOWN WEIGHT	UNKNOWN	P
At 0 KG. Load			
Peak L5-S1 extension moment	7.3 NM	9.5 NM	.029
Peak knee extension moment	2.5 NM	3.7 NM	.031
Peak acromion velocity	.95 M/S	1.3 M/S	.005
Peak hand velocity	1.01 M/S	1.3 M/S	.072
At 15 KG. Load			
Peak acromion velocity	.87 M/S	.97 M/S	.018
At 30 KG. Load			
Peak acromion acceleration	4.6 M/SxS	4.1 M/SxS	.048

If we assume that the mean estimate is 20-25 kg. then the hypothesis is confirmed. We would expect the greatest differences at 0 kg. The results show many parameters with large differences. At 15 kg. only one parameter has a statistically significant difference, and this difference is less than at 0 kg. For each parameter, the data indicate that more force was applied when the subject did not know the load he was lifting.

At 30 kg. a statistically significant difference was present in the data for one parameter. It indicated that more force was used when the subject knew the load.

MEASUREMENT OF UPPER LIMB MOTION USING FLEXIBLE ELECTROGONIOMETERS

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INTRODUCTION

Attempts to measure the joint angles of the upper limb have resulted in the design of a variety of electrogoniometers (Askew et. al., 1980), the majority of which have been based on the use of potentiometers. In general, instruments of this nature have been of bulky design and have been individually adapted to the joint under study. This paper describes the use of multi-axis, flexible electrogoniometers based on strain-gauge technology. These devices are lightweight and extremely flexible, with negligible interference to the subject.

The human arm may be considered as a linkage mechanism which is used to position the hand in space. The nature of this seven degree-of-freedom system results in a single joint angle being insufficient to describe the orientation of the arm. Combinations of joint angles such as shoulder flexion with elbow flexion describe functional movements more adequately. Studying the coordination of the range of motion available at the wrist, elbow and shoulder complex becomes of great importance when pathological or surgical conditions restrict functional activity (e.g. rheumatoid arthritis or total joint replacement). This leads to the question of compensation at the unaffected joints, which must occur if the required activity is to be successfully performed. The current measurement system enables simultaneous measurement of multi-axis movement at several joints, incorporated with data collection and subsequent analysis.

METHOD

Commercially available flexible electrogoniometers were attached to the joints of the upper limb. Twin-axis electrogoniometers (Penny & Giles TYPE M180) monitored elbow flexion/extension and the glenohumeral movements of abduction/adduction and flexion/extension. Elbow flexion was measured using one channel of the electrogoniometer. The normal subjects performed a variety of activities representative of daily living, including hand to mouth (eating), hand to head (grooming), hand to left and right shoulder (dressing) and hand to buttock (toileting). Each activity was practised and subsequently performed five times, with rest periods of five seconds between each arm movement. The joint motions were recorded at a frequency of 50 Hz and the signals were stored on a 4-channel digital data recorder (Penny & Giles TYPE DL1001). Associated software enabled calibrated angle-time and angle-angle graphs to be produced using an Amstrad PC1640 microcomputer.

RESULTS

Figure 1 shows a typical set of results for the motion of shoulder and elbow joints during the toileting activity. As the hand was moved behind the body the shoulder was extended 40° and initially abducted 14° with the elbow being flexed 46°. The return of the arm to the neutral position by the side of the trunk required a second abduction movement of 12° to enable the forearm segment to move laterally prior to forward flexion of the shoulder. The coordination between shoulder and elbow joints is of prime importance for smooth efficient motion of the arm segments and figure 2 shows the combined plots of shoulder abduction with elbow flexion (fig. 2a) and shoulder flexion with elbow flexion (fig. 2b). The similarity in motion patterns for the forward and recovery stages of the activity highlight the advanced level of inter-relation between the joints of the upper limb.

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PEAK POWER OUTPUT IN WHEELCHAIR SPRINTING.

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Few data have been published on the maximal (anaerobic) work capacity in handrim wheelchair propulsion. Coutts and Stogryn (1987) published values for maximal power output calculated over 5 sec. periods during an anaerobic Wingate test with six wheelchair athletes using their own wheelchair. For the Wingate test as developed for arm- and leg exercise, an optimum load exists at which the highest maximal power output levels can be reached (Dotan & Bar-Or, 1983). Similar data for wheelchair exercise are however not available.

In kinematic analyses of wheelchair-racing the phenomenological distinction in push- and recovery phase is usually made. The combination of stroke parameters with force data has not been published and it is usually assumed that during the full contact phase force is applied to accelerate the rims, although the existence of a braking phase during the first part of the push was suggested (Sanderson & Sommer, 1985).

The objective of this (pilot)study was to evaluate the power output that could be produced during wheelchair sprinting on a stationary ergometer and to determine in what range an optimum load can be expected at which the highest power output is found. In addition, the relation between push-time as determined from cine and torque was studied to evaluate the existence of a braking torque during the first part of the push.

Six male able-bodied subjects (age 27 ± 4.05 yr, weight 75.45 ± 13.1 kg) performed nine 20-second sprint-tests on a prototype stationary wheelchair ergometer. The friction load varied from 0 to 22.4 Nm per wheel (0-8 kg, 57 cm handrim). Inertia was simulated using adjustable flywheels. Torque applied to the rim and velocity were recorded at 100 Hz, separately for both wheels. This enabled a maximum sampling period of 20 seconds. After filtering at 20 Hz power output was calculated. Five friction levels (0, 2, 4, 6 and 8 kg) were filmed at 50 Hz. Statistical analysis comprised a one-way ANOVA with repeated measurements.

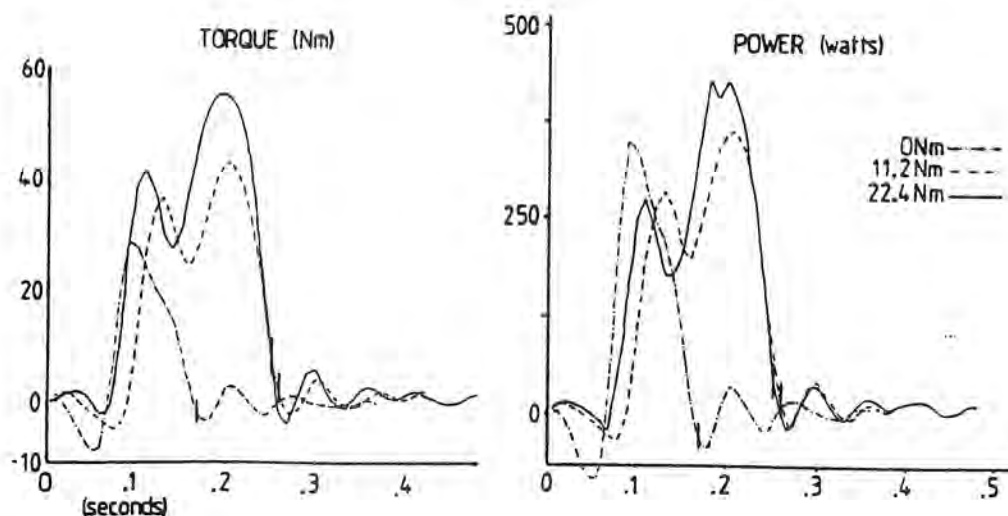


Figure 1. Torque and Power for one subject against friction loads of 0, 11.2 and 22.4 Nm. The vertical dashes indicate the end of the push-phase.

Of the three indices of performance of the Wingate test as mentioned by Bar-Or (1987), the maximum power averaged over 5 seconds (P5) and the rate of fatigue (RF) over 15 seconds were calculated (Table 1). Also given are the maximal Power output (Pmax) measured within one cycle and the highest mean Power outputs calculated over the complete cycle (Pcycle). Pmax remained steady over the nine tests, while other indices increased. Examples of the individual torque and power curves are given in Figure 1. A clear braking phase occurred during the beginning of a push phase which was highest at the lower loads. Both determinations of push phases correlated ($r=.72$, $p<.001$) despite difficulties in determining the exact starting positions of a push both by the inspection of torque and film.

Contrary to other indices, Pmax was not dependent on friction load, but stayed at around 375 Watts per arm. Pcycle, P5 and RF did more or less reach an optimum at 14.0 Nm. On basis of these indices an optimum load might be found around a resistance level of 14 Nm per wheel.

The average value of P5 for this study (75.9 W for one arm over 9 friction loads) was slightly higher than the results of Coutts & Stogryn, who reported values of 137 to 148 Watts for three paraplegic athletes, but did not state resistance levels. Probably due to the possibility of continuous force application P5 for armcranking has been reported from 350 (Dotan & Bar-Or, 1983) to 530 Watts (Patton & Duggan, 1988).

As already suggested by Sanderson & Sommer (1983), a braking force occurs when the hands come in contact with the rim. This braking force occurred at higher speeds and might have been the result of a lack of technique in our subjects. These losses may be prevented by applying a propulsion technique currently in use in wheelchair racing in which athletes beat the rims.

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Condition Load (Nm)	S1 7 (sd)	S2 2.8 (sd)	S3 5.6 (sd)	S4 8.4 (sd)	S5 11.2 (sd)	S6 14.0 (sd)	S7 16.8 (sd)	S8 19.6 (sd)	S9 22.4 (sd)	F-value
Pmax (W)	362.1 (91.5)	363.4 (79.2)	376.9 (75.1)	369.0 (85.5)	362.2 (40.3)	385.6 (120)	408.8 (114)	364.4 (77.2)	382.8 (102.2)	.8
Pmax (Nm)	29.2 (7.7)	32.2 (6.7)	35.4 (6.1)	37.4 (5.1)	39.0 (4.0)	44.6 (4.6)	49.3 (7.5)	47.9 (7.4)	49.9 (7.8)	15.9**
Vmax (m/s)	3.5 (0.2)	3.2 (0.2)	3.0 (0.4)	2.7 (0.3)	2.7 (0.1)	2.5 (0.2)	2.4 (0.3)	2.2 (0.2)	2.2 (0.2)	50.1**
P5 (W)	24.0 (17.3)	30.5 (15)	65.6 (8.4)	79.4 (17.5)	81.6 (19.4)	91.5 (19.7)	89.9 (20.4)	99.6 (17.7)	101.0 (19.6)	34.0**
RF (%)	13.0 (7.8)	17.7 (5.8)	28.6 (20.4)	28.7 (8.4)	24.3 (7.9)	35.0 (18.4)	35.0 (6.9)	33.1 (9.3)	35.4 (10.1)	3.5*
Pcycle (W)	31.8 (15.1)	69.4 (20.2)	83.0 (12.7)	88.7 (13)	94.5 (10.2)	111.4 (16.3)	111.0 (25.2)	111.4 (15.7)	114.1 (11.9)	32.9**
Pcycle (Nm)	6.1 (1.1)	6.2 (1.6)	8.0 (1.2)	9.4 (1.1)	10.3 (1.2)	13.5 (2.0)	14.0 (1.9)	15.7 (1.3)	16.3 (1.3)	130.1**

TABLE 1.

Average values (N=6) for Torque and Power indices listed over nine friction loads together with their F-values. Velocity is expressed as tangential velocity of the hand rims.

*: $p<0.05$ **: $p<0.01$

BIOMECHANICAL DETERMINANTS OF ABILITY TO RISE FROM A CHAIR

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Introduction

Loss of ability to get up from a chair without assistance can have substantial consequences, particularly in elderly persons when it contributes to their institutionalization. The design of effective therapies to regain or retain the ability to rise from a chair calls for a better understanding of its biomechanical requirements than presently exists.

Methods

This study employed a computer-simulation model to quantify the biomechanics of rising from a chair. Other, ongoing, studies suggest that dynamic effects are not necessarily major. Consequently, the model considered only the requirements for quasi-static equilibrium during the rise. The model incorporated ten rigid links to represent planar configurations of the feet, lower and upper legs, pelvis, lower and upper trunk, head/neck, lower and upper arms, and the hands. Mean values for male adults were selected from the anthropometry literature and assigned to the masses, lengths, and mass center locations of the segments. The segments were placed into prescribed configurations. For each configuration the location of the total body mass center and the net torque on each joint were computed.

Some of the configurations examined were those that had been observed earlier in 7 young and 10 elderly male adults performing a sagittally-symmetric chair rise without use of hands, and under well-controlled conditions. These subjects were observed to first move their upper body segments forward while remaining seated, then to rise to a standing position. The body configuration at the switchover point was one of major interest.

Results and Discussion

The calculations suggested that there are two important determinants of ability to rise from a chair slowly without use of hands. One is the need to bring the total body mass center forward of the heels. The second is the need to diminish the joint torque strengths required in the ankles and knees, and sometimes in the hips, for quasi-equilibrium.

In the fully upright sitting position the total body mass center lies (all numbers quoted are approximate values) 33 cm posterior of the ankle joint. In order to rise, the mass

Full motion of the shoulder, including humeral rotation can now be measured with the development of triaxial electrogoniometers (Nicol, 1987) and these measurements will be implemented in parallel with the evaluation of selected rheumatoid patients.

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ACKNOWLEDGEMENTS

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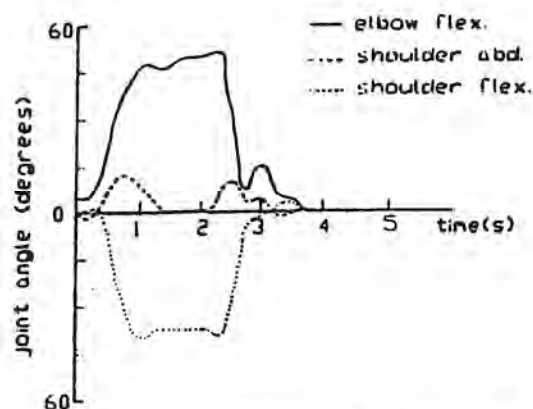


Figure 1 : Typical angle-time plot for normal toileting activity - hand to buttock.

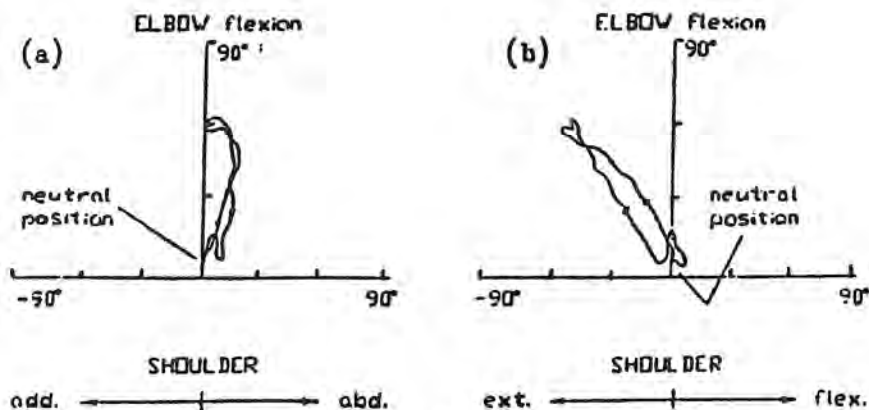


Figure 2 : Typical angle-angle plots for normal toileting activity - hand to buttock.

STRETCH REFLEX RESPONSES OF THE HUMAN M. TRICEPS SURAE FOLLOWING MECHANICAL STIMULATION

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In stretch shortening cycle contractions the stretch reflex activation can be interpreted as the servo mechanism to control stiffness behavior of the leg extensor muscles (Dietz et al. 1981; Gollhofer et al. 1984). In complex movement conditions it is difficult to separate the electromyographic responses into reflexly generated and centrally generated neuronal activity. The purpose of the study was to examine the variants and invariants of the stretch reflex responses in well controlled conditions. In analogy to the studies carried out by Gottlieb and Agarwal 1979 and Berardelli et al. 1982 the experiments were performed using mechanical stimulation around the ankle joint by applying various dorsiflexion stimuli.

16 male students of physical education participated in the study. They were informed about the various experimental conditions and gave their consent.

The mechanical stimulation was applied by a motor torque device (Geisinger) controlled by a digital feedback-system. In all experimental conditions the subjects sat in a chair. The thigh of the right leg was fixed and the foot was mounted on the rotation platform so, that the rotation axes of ankle joint and motor drive coincided.

The subjects were instructed not to resist or intervene the mechanical perturbation ("let go"-condition). They were blindfolded and via headphone from audible feedback disconnected. The various stimuli were applied randomly.

Torque and angular position were stored on a micro computer system (Victor-Sirius I) together with the signals of the full-wave rectified surface emgs from M. Soleus and M. Gastrocnemius. In order to calculate the average movement pattern from each condition eight repetitions were applied. Single dorsiflexion stimuli were analyzed with respect to the amplitude (3,6;5,4;7,2;10,8;14,4 deg; velocity: 87,33 deg*s⁻¹) and with respect to the velocity (43,66;87,33;131,00;174,70 deg*s⁻¹; amplitude: 7,2 deg). Additionally double dorsiflexion conditions were conducted (first and second stimulus 87,33 or 174,70 deg*s⁻¹; time delay between first and second stretch 100 or 200 ms).

In all experimental conditions mainly monosynaptic reflex responses could be detected. In single stretch condition the integral as well as the amplitude of the monosynaptic stretch response was not related to the amplitude of the dorsiflexion but rather linearly related to the stretch velocity. The latencies of the answers were within a range of 3-5 ms only sensitive to the velocity of the dorsiflexion. In double stretch condition all combinations of the first and second stimuli were applied. The responses to the first stimulus were ordinary and directly comparable to the data of the single stimulus experiments. The responses to the second stimulus however were either totally suppressed or drastically diminished (Table 1).

1 st stimulus		2 nd stimulus 47.11 deg*s ⁻¹			2 nd stimulus 174.70 deg*s ⁻¹		
deg*s ⁻¹	Response	100 ms	delay	200 ms	100 ms	delay	200 ms
87.33	60.90% (22.27)	4.58% (10.08)		20.65% (22.68)	16.58% (18.56)		22.01% (14.01)
174.70	100.00%	1.40% (3.16)		2.64% (5.69)	11.66% (17.66)		28.59% (25.98)

Tab. 1: Mean values (stddev.) of the integrated reflex responses (M. Soleus) in double stimulus experiment. The values are expressed in % (reflex response in 174,70 deg*s⁻¹ corresponds to 100%).

In agreement to the studies of Berardelli et al. 1982 it is suggested that the monosynaptic stretch reflex response is a function of the acceleration of the stimulus applied. This result is supported by the linear relationship between stretch velocity and reflex response together with nearly unchanged latencies. Furthermore, the results indicate that in "let-go"-conditions the monosynaptic reflex arc is subjected to a refractivity which is dependent (1) on the strength of the first stimulus, (2) on the time interval between both stimuli and (3) on the strength of the subsequent stimulus.

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BIOMECHANICAL MEASUREMENT OF THE SWALLOWING REFLEX

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INTRODUCTION

Disorder of the swallowing reflex is called dysphagia and presents a major problem in the comprehensive rehabilitation of stroke and head injured patients. Current clinical practice involves diagnosis based on qualitative assessment. Noninvasive quantitative measurements are necessary to help the clinicians to objectively assess the dysphagic patient, and to objectively identify the patient at risk of aspiration. We are developing such techniques for quantifying the oral and pharyngeal phases using several biomechanical parameters.

METHODOLOGY

For the measurement of the oral phase, we have identified and developed techniques to measure several biomechanical parameters which characterize the oral musculature. These include: (1) lip closure pressure (contact compressive stress), (2) lip pulling force (contact shear force), (3) tongue thrust (in the lateral directions and in the forward direction), and (4) swallow suction pressure.

For the assessment of the pharyngeal phase, we measured acceleration at throat with two ultraminiature accelerometers placed on the outside of the throat, at the level of the larynx, above and below the cricoid cartilage. We simultaneously measured the swallow suction pressure with a catheter placed at the base of the tongue.

RESULTS

Swallowing in normal subjects gave rise to a typical acceleration characteristic of swallowing (Fig. 1). There was no time lag between the initiation of pressure wave and the acceleration wave. In contrast, the characteristic acceleration was either absent or significantly delayed (average 4.2 seconds) in dysphagic patients (Fig. 2). Moreover, normal subjects made only one attempt at swallow while dysphagic patients required a number of attempts (average 4.3) before they could trigger a swallow. The swallow pressure represents the oral phase activity and the acceleration represents the pharyngeal phase activity. The time lag between the pressure and acceleration waves gives a measure of the coordination between the two phases. The chance of aspiration increases with the time lag. We are using these measurements to objectively assess the dysphagic patients

initially (Fig. 2) and after periods of about three weeks.

CONCLUSION

The accelrometry technique, together with oral phase bio-mechanical measurements, can aid the clinician to objectively assess the dysphagic patient and to identify the patient at risk of aspiration.

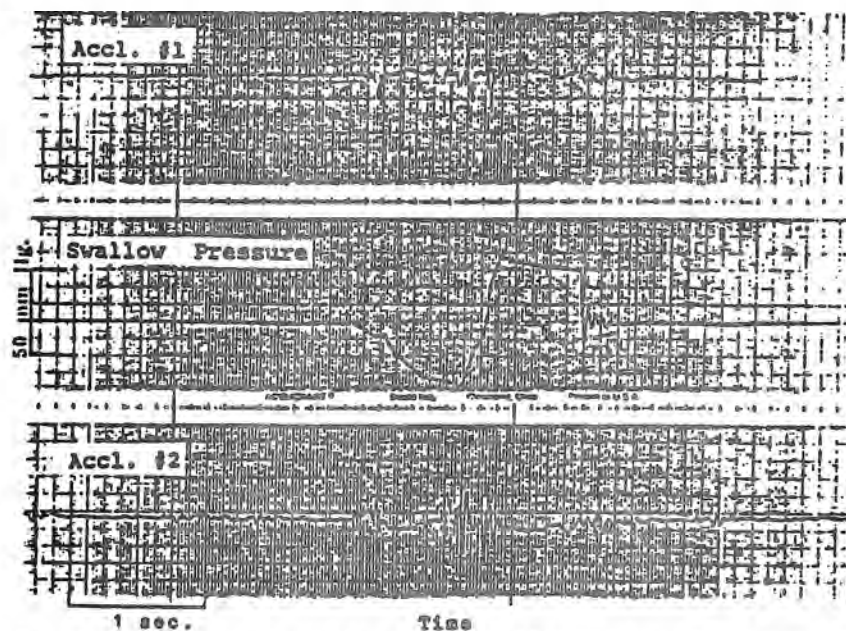


Figure 1: Typical acceleration and swallow pressure recording in a normal subject. Note there is no time lag between pressure and acceleration waves.

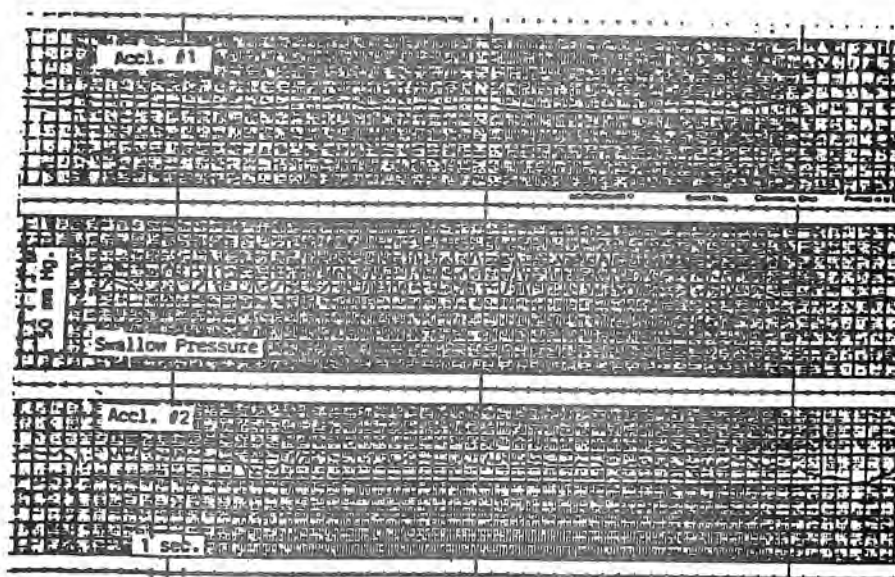


Figure 2: Recording in a dysphagic patient. Note the number of attempts as seen from swallow pressure. Also note the absence of the typical acceleration pattern.

MECHANISMS UNDERLYING THE AGE-RELATED DECREASE IN SPEED OF CONTRACTION. An experimental study of fast- and slow-twitch muscles, single motor units and skinned muscle fibers in the rat.

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Introduction

A decreased speed of contraction during aging, measured under both isometric and isotonic conditions, has been observed in various mammals, including man (e.g. Gutmann et al. 1971; Caccia et al. 1979; Larsson et al. 1979; Bassey & Harries 1987; Larsson & Edström 1986; Edström & Larsson 1987). It has been speculated that the reduction in speed of contraction in old age, which is seen before the senile muscle wasting becomes manifest (Larsson & Edström, 1986), may be due to an age-related loss of fast-twitch muscle fibers or to changes in the properties of the contractile material irrespective of the muscle fiber type, or both (Campbell, McComas & Petito, 1973). Physiological, morphological and enzyme-histochemical methods were used to study fast- and slow-twitch muscles, motor units and skinned muscle fibers in young and old rats with the aim of improving the understanding of the mechanisms underlying the decreased speed of contraction in old age.

Methods

Young (3-6 months) and old (20-24 months) male albino rats of the same strain (Wistar) were compared. Contractile, morphometrical and enzyme-histochemical properties were studied in fast- and slow-twitch muscles (tibial anterior, soleus) and single motor units. Single motor units were functionally isolated by microdissection of the ventral root filaments and the muscle fibers in the unit were identified via glycogen depletion. The sarcoplasmic reticulum (SR) volume and Ca^{2+} uptake activity was measured in chemically skinned single fast- and slow-twitch muscle fibers (for detailed information of methods see Edström & Kugelberg 1968; Edström & Larsson 1987; Larsson & Edström 1986; Kugelberg & Lindegren 1979; Salviati et al. 1982).

Results

Twenty-three and 44 % longer contraction times of the isometric twitch ($p < 0.01-0.001$) were observed in the fast-twitch tibial anterior and the slow-twitch soleus muscle in old age, respectively. In the soleus muscle, the absolute and relative number of type II fibers was lower in the old animals. In the tibial anterior muscle, on the other hand, there was no age-related difference in either total fibre number or fiber type proportions. In enzyme-histochemically identified single motor units, the contraction time increased ($p < 0.05$) 20 % in fast- and

24 % in slow-twitch units in old age. A strong linear correlation ($r=0.83$; $p<0.001$) between the maximum rate of increase in tetanus force and the contraction time of the isometric twitch was observed in motor units from young but not from old animals. In fast-twitch skinned fibers, the volume of the SR decreased 18 % ($p<0.05$) and the specific activity of the Ca^{2+} pump decreased 32 % ($p<0.001$). In slow-twitch fibers from the soleus, SR volume and intrinsic function was not affected by age.

Conclusions

It is concluded from the present experiments that: there is a differentiated age-related change in the speed of contraction between fast- and slow-twitch muscles, motor units and muscle fibers; the reduced speed of contraction in old age is primarily related to altered mechanical properties of both fast- and slow-twitch muscles; in fast-twitch fibers, an impairment of intrinsic function and volume of the SR system appears to be the dominant underlying mechanism; in slow-twitch fibers, the underlying mechanism(s) remains obscure. Other elements, than SR properties, in the excitation-contraction coupling which influence the speed of contraction, such as regulatory and/or contractile proteins, may have been altered during ageing. An increased muscle stiffness due to altered parallel elastic components is another conceivable underlying mechanism. Indirect evidence of an altered resting tension at L_0 as well as an increased proportion of fat and connective tissue in old soleus lend some support to the latter alternative.

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A COMPUTER MODEL ANALYSIS DETERMINING THE LUMBAR COMPRESSIVE AND SHEAR FORCES DURING VARIOUS TRUNK CURL-UP EXERCISES

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Concern has developed recently that certain styles of sit-up exercises practiced today may result in increased loads being placed upon the spine and thus, may lead to the development of low back pain(1,3). In the literature to date, no non-invasive models have been developed to determine the lumbar forces during such exercises. Therefore, a need existed for the development of an non-invasive model which determines lumbar forces during the execution of curl-up exercises.

BIOMECH6, a 2-dimensional static computer model in the sagittal plane, determines the lumbar compressive and shear forces during curl-up exercises. Several anthropometric parameters required for the model to operate were previously obtained. The purposes of this study was to determine the two remaining parameters (iliopsoas tension and lumbar spine curvature) and the lumbar compressive and shear forces during various curl-up exercises.

Methods:

The curl-up styles analyzed by BIOMECH6 were the longlaying (0° hip and knee flexion), hooklaying (hips flexed 45°, knees flexed 90°) and the bench (hips, knees flexed 90°). During each curl-up 2 positions were analyzed: the initial start position and the position when the trunk was 45° to the supporting surface. Using a system of external spine markers and Queen's University software, a photograph was taken in each curl-up position and digitized to determine the lumbar spine curvature; 9 males and 10 female subjects were used. To determine the iliopsoas tension in each position, 14 males and 16 females were used as subjects. Five-second maximal voluntary isometric contractions on the right leg were performed with the Kinetic Communicator Exercise System at hip flexion angles of 0°, 45° and 90°. Peak torque in each position was analyzed as a percentage of each person's maximum torque. These data were entered into the computer model and the lumbar forces were determined for each curl-up position.

Results and Discussion:

The maximum compressive and shear forces were found to be at the L3/L4 level and L2/L3 level, respectively, during the initial position of the longlaying (LL) curl-up. The maximum compressive force ranged from 743-1306 N (male) and 634-1262 N (female), depending on body weight. These results are slightly lower than those suggested by Nachemson (3). The maximum shear forces ranging from 148-246 N (male) and 119-238 N (female) were comparable to literature values (2).

Since there is limited comparable literature, the relationship in the force reduction from one position to another is discussed. The lumbar forces were maximized during the initial position of the LL curl-up because in this position the lumbar lordosis (Fig.1) and iliopsoas tension were maximized. During the initial position of the hooklaying curl-up, the compressive forces were only reduced by 5% (male) and 4% (female) while the shear forces were reduced by 46% (male) and 39% (female). In this position, the lordotic curve reduced slightly and the iliopsoas still generated 80% (male) and 70% (female) of the maximal tension produced during the LL curl-up. The bench curl-up (initial position) minimized the lumbar forces; the compressive forces

were reduced by 17% (male) and 18% (female) while the shear forces were reduced by 87% (male) and 97% (female). In this position, the lumbar spine flattened out and the contractile tension of the iliopsoas was only 50% (male) and 42% (female) of its maximum. The lumbar forces reduced for the 45° curl-up position in all 3 variations in comparison to the initial position, except for the bench curl-up where the shear forces increased slightly.

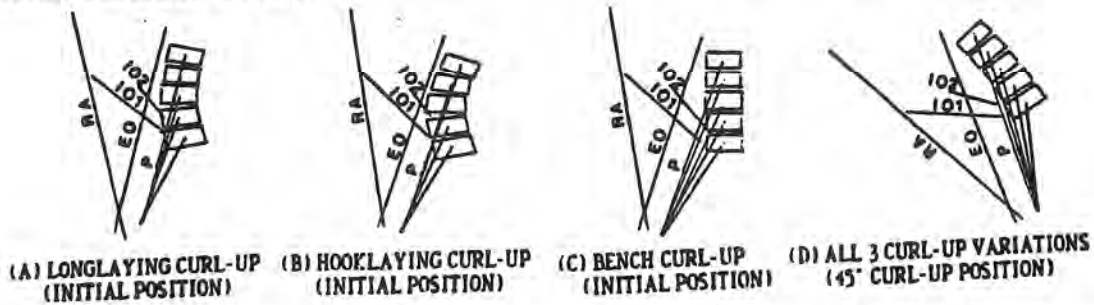


Fig. 1: Lumbar Spine Curvature of the 4 Female BIOMECH6 Models (RA=rectus abdominis; EO=external oblique; IO1=lower and IO2=upper internal oblique; P=psoas).

Conclusions:

The results of the present study indicate that performance of the bench curl-up minimizes the lumbar compressive and shear forces. Additionally, the magnitude of the forces produced by BIOMECH6 seem to be reasonable and fit well with in the range of those expected during such exercises.

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PREDICTION OF THE PINCH AND GRASP FORCE EXERTED BY THE HAND USING SURFACE AND FINE WIRE ELECTROMYOGRAPHY

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INTRODUCTION

Armstrong (1979) presented a straight forward method for estimating the hand forces using surface EMG from the forearm. Many factors affect the relationship between applied hand force and EMG. For instance, the hand forces are heavily influenced by the posture of the wrist (Miller and Wells, 1988). It is also known that the surface electrodes may only collect the signals from the activity of the superficial muscles. The ability to predict hand grip forces is important in investigating occupational hand and wrist injuries. This study investigates the errors in this process.

METHODS AND PROCEDURES

Bipolar fine-wire electrodes, 40 μ m in diameter, were implanted in the flexor digitorum profundus muscle (FDP) and the digitorum superficialis muscle (FDS) of the index/long finger. The interpolar distance was about 4 mm. Bipolar silver/silver chloride surface electrodes, 0.7 cm in diameter, were placed on the anterior side of the forearm (SFM) and the posterior side of the forearm (SEM). The interelectrode distance of each pair of bipolar electrodes was 3 cm. A common ground electrode was located on the lateral epicondyle. The total forces exerted by a power grip and a tip pinch were measured by two Linear Variable Differential Transducers (LVDT).

The subjects participating in the test were three, male, ranging from 25 to 30 years of age who had no history of hand injury. Five wrist positions (neutral, flexed, extended, ulnar deviation, radial deviation) with two contractional speeds (slow, fast) and mixed wrist position of the power grasp and two-finger pinch were investigated. In each trial, the subject performed several grasps or pinches using various forces at a fixed wrist posture and speed during 20 seconds. The calibration trials consisting of a ramp MVC and a 50% MVC contraction in neutral position were performed before and after all trials.

The myoelectric signal was amplified through a bioamplifier (bandwidth 1 Hz to 1 KHz). The signal was then

full wave rectified, low-pass filtered at 6 Hz through a EMG processor, and was collected, with the force signal using an A/D converter system at 50 Hz/channel and stored.

The EMG data were further filtered using a digital filter algorithm (Butterworth low pass filter) with a cut-off frequency at 1.5 Hz. The calibration constants, between EMG and force, were produced for each EMG channel from the calibration trial using the least squares method.

RESULTS AND DISCUSSION

The correlation coefficients (R), averaged over the three subjects, between the different muscle EMGs and the hand forces under various wrist postures and speed are shown in Table 1. The correlation coefficients (R) of flexor digitorum profundus (FDP) EMG with the hand forces are the highest of any of the four EMGs measured, ranging from 0.81 to 0.96. For the surface EMG of flexors, the Rs are high at neutral and extended postures and lower at flexed and mixed postures. The correlation coefficients of the slow contractions are higher than those of the fast contractions. Bigland and Lipold (1954) indicated that the muscle tension, velocity and the EMG are interdependent which means that the correlation between the force and the EMG is changing as the contraction speed changes.

The results of the average relative errors of the predicted forces using the different EMGs were summarized in Table 2. The errors are heavily influenced by the postures of wrist. During flexion the predicted forces were too high. This phenomenon can be accounted for by the fact that the relationship of the EMG with the muscle force is modulated by the muscle length. This is shortening with the flexing of the wrist. The results of the predicted force during the slow grasp in extended position and slow pinch in radial deviation could be improved from 38% to 27%, 36% to 29% (FDP) respectively, if the first contraction of each trial was employed to calculate the calibration constants for the trial itself. Figure 1 illustrates a typical result of the predicted force from FDP with 33% average relative error and Figure 2 illustrates a typical result from sur-

face flexor EMG with a typical average relative error of 48%.

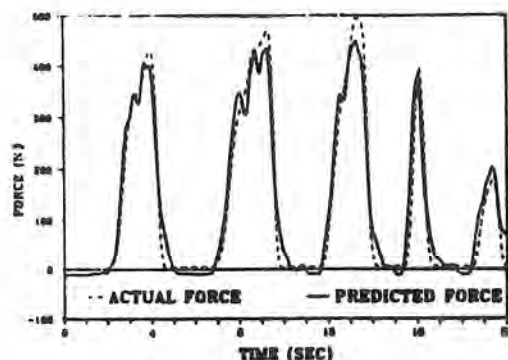


Figure 1. The predicted force using FDP EMG

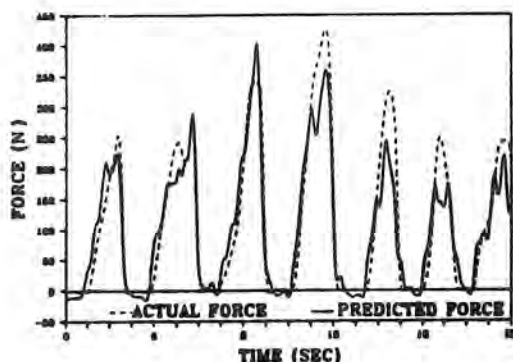


Figure 2. The predicted force using surface EMG

The results from the FDP are the best among the predicted forces at each wrist posture. At neutral and extended posture, the results of the surface EMG flexors are very close to those of FDPs, but at other wrist posture, especially in mixed postures, the results are poorer.

From Table 2, we can see that the results of the predicted forces are also influenced by the speed of the contraction. The results of the slow grasps or pinches are better than those of the fast. Also found that the predicted force using the calibration curve at 50% MVC are better than when using the 100% MVC.

CONCLUSION

The process of predicting force from EMG provides the good estimate of the pattern of force application, however the errors in the process are not insignificant. Other methods of measuring this force, such as instrumented tools, prediction of hand force from object

weight or from subject recall, may be even more inaccurate or may not be feasible. Surface EMG therefore provides one possible approach to determine hand forces if the errors found here are taken into consideration.

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Table 1. The Correlation Coefficients of EMGs with the Hand Force

	Grasp						Pinch					
	N	Fl	Ex	Ul	Rd	Mx	N	Fl	Ex	Ul	Rd	Mx
FDP	.95	.93	.93	.93	.89	.86	.81	.84	.93	.86	.86	.83
FDS	.93	.78	.89	.95	.96	.93	.85	.75	.94	.86	.87	.87
RFM	.96	.84	.91	.89	.84	.87	.70	.80	.90	.81	.76	.76
SEM	.93	.94	.70	.93	.83	.74	.88	.66	.90	.75	.63	.63
	Grasp						Pinch					
	N	Fl	Ex	Ul	Rd	Mx	N	Fl	Ex	Ul	Rd	Mx
FDP	.96	.94	.96	.90	.88	.94	.93	.93	.94	.91	.83	.83
FDS	.94	.71	.93	.88	.88	.93	.77	.87	.90	.89	.70	.70
RFM	.94	.88	.98	.94	.88	.91	.84	.93	.88	.91	.74	.74
SEM	.91	.91	.85	.84	.74	.90	.91	.69	.88	.71	.70	.70

Table 2. The Relative Errors of the Predicted Forces (%)

	Grasp						Pinch					
	N	Fl	Ex	Ul	Rd	Mx	N	Fl	Ex	Ul	Rd	Mx
FDP	25	35	36	33	35	31	44	39	43	44	45	45
FDS	34	53	71	60	67	48	71	38	61	70	66	66
RFM	33	64	40	44	48	39	58	42	53	54	64	64
SEM	72	61	169	84	100	82	54	76	69	111	131	131
	Grasp						Pinch					
	N	Fl	Ex	Ul	Rd	Mx	N	Fl	Ex	Ul	Rd	Mx
FDP	29	43	41	38	36	31	47	40	51	48	56	56
FDS	37	63	63	43	44	38	50	60	77	67	78	78
RFM	38	67	44	61	50	38	71	49	63	65	71	71
SEM	72	53	78	134	111	77	56	199	90	133	146	146

* N: neutral position, Fl: flexed, Ex: extended, Ul: ulnar deviation, Rd: radial deviation, Mx: mixed

MEASUREMENTS OF MUSCLE FIBER CONDUCTION VELOCITY AT THE M. BICEPS AND M. TRICEPS BRACHII UNDER ISOMETRIC LOAD

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INTRODUCTION

Muscle fiber conduction velocity (MFCV) is related to physiological characteristics which are dependent on the contractile properties of different fiber types. Animal research showed that the soleus muscle with 24% FT fibers had a significantly lower conduction velocity than the M. extensor digitorum longus with 98% FT fibers (2). So far, MFCV measurements in humans have been restricted to the M. biceps brachii because its long fibers allowed the use of conventional electrodes. In other muscles the fibers are considerably shorter. To be able to place electrodes between motor point and the end of the fiber in small muscles, very short electrode assemblies are necessary. A unit with two pairs of interlaced electrodes with an inter-electrode distance of only 5 mm was built, to allow the measurement of the MFVC in short fibered muscles. Using this electrode assembly, in the present study the MFCV was measured in two muscles of different fiber composition (1). The intraindividual day-to-day reproducibility and the variability between subjects was determined for the MFVC.

METHODS

Eight students of the physical education department were tested on three different days. MFVC measurements were taken from the biceps and triceps brachii at 20, 40, 60, 80, and 100% of the maximum voluntary contraction (MVC) as well as in isometric dynamic explosive contractions. Three trials were collected in each condition. The angles between trunk and arm, arm and forearm, wrist and force transducer system were kept at 90 degrees. The freedom of movement in the shoulder and elbow joint was mechanically restricted. The wrist was fixed in an U-shaped device and connected to a piezoelectric force transducer. The subjects were able to monitor their force application as a voltage trace on an oscilloscope.

The electrode assembly (with integrated preamplifiers) had a size of 40 x 16 x 4 mm. It was positioned at a site between motor point and tendon insertion. The electrodes were repositioned until a maximum cross correlation between the two EMG signals could be found (mean $r = .92$, $n = 864$). The skin was abraded until the impedance was below 20 KOhm. The recordings were taken from the long head of the triceps and the common head of the biceps between the motor point and fiber insertion. The electrode sites were marked to allow for day-to-day recordings from the same site.

The signals were recorded at a frequency of 25 KHz and stored after AD-conversion on an IBM-AT-PC. The MFCV was determined using a cross-correlation technique (3).

RESULTS

1. Intraindividual comparisons revealed for the isometric static and dynamic contractions in seven from eight subjects increased MFCV values in the triceps ($p = .05$) as compared to the biceps brachii. For all subjects the following mean values were found:

	Biceps	Triceps
Mean MFCV dyn.	4.0 m/s	5.2 m/s
Mean MFCV stat.	4.5 m/s	5.7 m/s

2. Although the relation between force and MFCV is non-linear in single subjects the mean MFCV values for all subjects increase with the level of force ($r = .98/p = .01$).
3. The MFCV of both muscles were significantly lower for isometric dynamic as compared to isometric static conditions ($p = .01$).
4. Table 1 shows the inter- and intra-individual coefficients of variability within an experimental session, between days and between subjects.

CV (%)	isometric static		isometric dynamic	
	biceps	triceps	biceps	triceps
CV intra	4.6	7.6	5.2	5.8
day/day	9.6	15.3	6.1	16.6
CV inter	14.2	24.5	10.4	23.4

Table 1. Coefficient of Variability of MFCV

DISCUSSION

Table 1 shows that the difference of the MFCV between subjects is higher than the inter-day variability. Therefore, the MFCV seems to be appropriate for individual analyses. The variabilities within a testing session and between days appear to be low. It should be noted that the triceps MFCV was significantly higher than the biceps MFCV. Because of the higher FT fiber percentage in the triceps this result may reveal a relationship between fiber type and MFCV. The significantly lower MFCV in isometric dynamic contractions could be caused by different recruitment strategies in static and dynamic force production. The MFCV appears to be a reproducible and predictive parameter in the description of physiological properties of muscles of different fiber composition.

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TIBIAL BONE ACCELERATION AND GROUND REACTION FORCE PARAMETERS DURING RUNNING

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INTRODUCTION & REVIEW OF LITERATURE

Animal experiments (Dekel et al., 1978) suggest that repetitive impulse loading of joints is a major factor for the development of degenerative osteoarthritis. From accelerometer experiments at different sites of the human body Voloshin and coworkers emphasize in a series of publications the importance of the shock absorbing properties of the human locomotor system (Voloshin & Wosk, 1982). A major limitation of acceleration experiments, however, has been the restriction to measurements at the skin surface rather than the bone itself (Light et al., 1980; Ziegert & Lewis, 1979; Lafortune, 1987). Comparing a bone mounted and a skin mounted tri-axial accelerometer during running, Hennig & Lafortune (1988) found large differences in shape and magnitude of the acceleration signals. The amplitudes from the skin mounted transducer were more than twofold higher than those registered at the bone. Previous reports on bone acceleration measurements were based on only single subjects. Using data from several subjects, it is the purpose of this contribution, to compare Ground Reaction Force and tibial acceleration parameters.

METHODS

Six male volunteers participated in this study. Two "Entran" tri-axial strain gage accelerometers with a mass of 6 g were fastened to two sites of the tibia. One transducer was attached to a 4.7 mm diameter Steinmann traction pin, that was inserted approximately 5-7 cm into the bone at the lateral condyle of the tibia. The second accelerometer was fastened to the skin above the bone at approximately half the distance between the medial malleolus and the medial tibial condyle. In each condition the subjects performed five trials of running (4.5 m/sec) across a "Kistler" force platform in a soft running shoe. The three force components of the force plate and three acceleration signals from each of the two accelerometers were sampled simultaneously by a computer based data acquisition system (12 bit; 1kHz/channel). A low pass digital filter was applied to the force (150 Hz) and acceleration data (50 Hz). Two LOCAM cameras filmed the contact phase of the subjects with the force platform at a frequency of 200 Hz. The film data were digitally low pass filtered at 30 Hz.

RESULTS

The axial acceleration as it is experienced by the bone is determined by gravitation, the impact and the centrifugal forces due to the rotatory movement of the shank. Although the centrifugal forces have a considerable influence onto the accelerations (Lafortune & Hennig, 1989), the present data reflect the actual accelerations at the tibia. For the description and comparison of the force and acceleration data following parameters were chosen: first and second peak of the vertical force in BW (PVF1, PVF2); first peak horizontal braking force in BW (PHF1), time to peak vertical and horizontal force in ms (TTPVF1, TTPHF1); peak positive

axial and A-P acceleration in g (PPA, PPAP), peak M-L acceleration in g (PML),

PVF1	PPA	TTPVF1	TTPPA	PVF2	PHF	PPAP	PML
2.15	5.32	30.5	25.3	2.75	1.09	6.31	4.62

Correlation Matrix for Variables: H₁ ... H₆

	PUF1	PUF2	PHF1	TTPUF1	PPA	PPAP
PUF1	1					
PUF2	-.54	1				
PHF1	.7	-.21	1			
TTPUF1	-.9	.35	-.66	1		
PPA	.76	-.41	.28	-.89	1	
PPAP	.68	-.11	.94	-.75	.37	1

DISCUSSION

As it was already shown by Lafortune (1987) for a single subject the AP acceleration peaks (PPAP) during running proved to be higher than the axial accelerations. This is partly due to the effect of the centrifugal forces. Eliminating the centrifugal force through film data calculations the modified peak axial accelerations (MPPA=9.4 g) become larger than the modified A-P accelerations (MPPAP=8.3g). The time to peak axial acceleration (TTPPA) differs significantly from the time to peak force (TTPVF1), and PVF1 shows only a moderate correlation with PPA ($r=0.76$). However, a high negative correlation ($r=-0.89$) is obvious for the comparison of PPA with the time to peak vertical force 1 (TTPVF1). These findings indicate a substantially different behavior of tibial acceleration and vertical ground reaction force and therefore impose a serious challenge to the concept of "effective mass" (Nigg et al., 1981). Similar findings are present when comparing the axial accelerations with the resultant force parameters. A high correlation of $r=0.94$ was found between the peak braking force and the AP-acceleration peak. This finding may be explained by a strong muscular stabilization of the body during impact.

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MUSCLE FUNCTION IN PRIMARY FIBROMYALGIA - A PRELIMINARY STUDY.

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INTRODUCTION: This study aimed to compare strength and endurance between healthy controls and patients with primary fibromyalgia. Within this aim we also wanted to compare the development of fatigue according to EMG variables between the two groups.

SUBJECTS: 10 healthy physiotherapists (the controls) and 8 patients with primary fibromyalgia (the patients). The patients were selected according to the criteria presented by Yunus et al., 1981. All subjects were between 40 and 50 years.

METHODS: The subjects were seated in a specially constructed chair and performed 100 repeated maximum isokinetic shoulder forward flexions (1.05 rad/s). Surface EMG were registered from the right trapezius and the anterior part of the right deltoid. The EMG signals and the torque signal from the dynamometer were analysed on a computer. The programme determined the mean power frequency (Hz) and the root mean square (RMS) value of the EMG signals and work (Joule) for each contraction. Each flexion started with the subjects resting their hand against the thigh. The shoulder flexions were performed up to 90 degrees flexion. Thereafter the subjects were instructed to relax their muscles and passively follow (using gravitation torque) the lever arm down and when their hand was felt against the thigh again immediately perform a new maximum shoulder flexion. Before and during the experiment (every 5th contraction) the subjects rated their perception of fatigue in the shoulder muscles using a 10 graded scale (Borg, 1982). For details concerning methods see: Gerdle et al., 1988.

RESULTS: The patients rated higher perception of fatigue than the controls and reached an extremely high feeling of fatigue within few flexions (figure 1).

No significant difference in work existed between the two groups. Moreover, both groups exhibited steep decreases during the initial 40-50 contractions followed by a stable work level (figure 2).

Mean power frequency of the two EMG signals also showed a pattern with two phases. No significant differences existed, initially or during the stable level, for the two EMG signals between the two groups.

The ability to relax the muscles during the passive extension

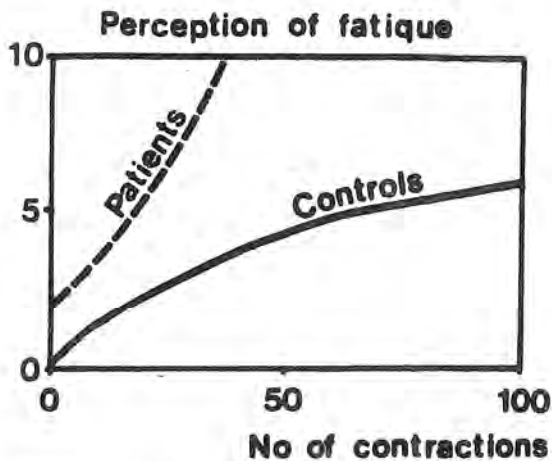


Figure 1: Perception of fatigue.

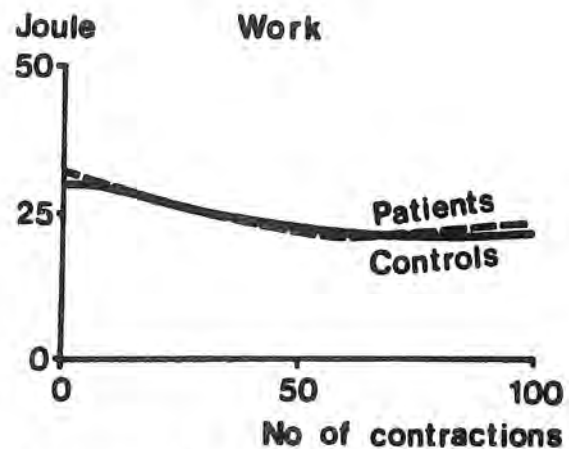


Figure 2: Work (Joule).

differed between the two groups. This EMG activity (RMS) was expressed in per cent (%) of the RMS value during the active flexion for each contraction (the signal amplitude ratio). Starting with only a small difference between the two groups the signal amplitude ratios of the controls decreased below 10 %.

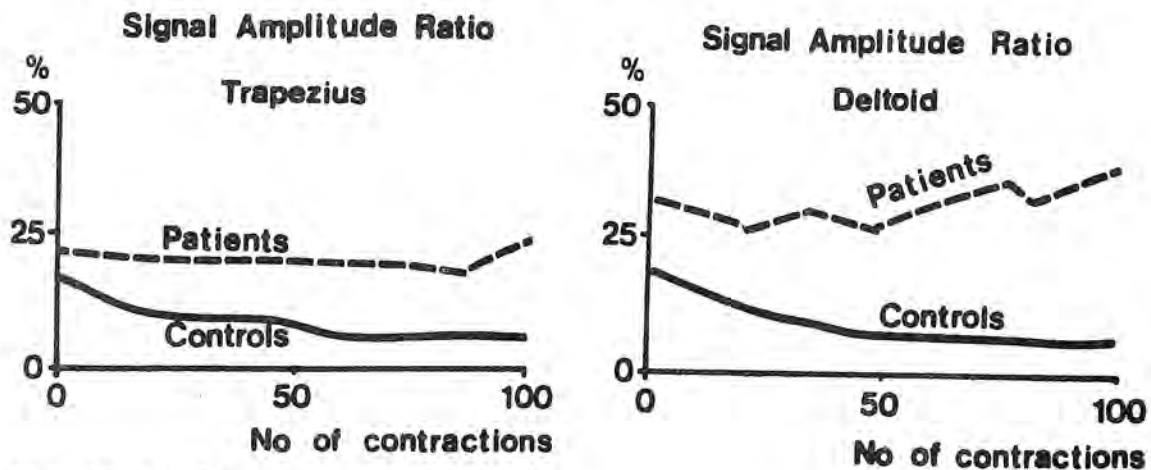


Figure 3: Signal Amplitude Ratios.

No distinct decrease occurred for the patients. During the last 50 contractions significantly higher signal amplitude ratios existed for the patients when compared to the controls.

CONCLUSION: The results indicated that tension and poor coordination played a part in developing or maintaining muscle pain in patients with primary fibromyalgia. From the present results there appears to be no need to train with the purpose to achieve better strength or endurance. It appears to be more adequate to train the ability to relax the muscles during supposed pauses.

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A NEW COMPUTER-ASSISTED MODEL TO ASSESS THE RANDOMNESS OF THE MUSCLE FIBER ARRANGEMENT IN SINGLE MOTOR UNITS.

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Introduction

Clustering of muscle fibers of the same enzyme-histochemical type is a common histopathological finding, which indicates a rearrangement of the muscle fibers in the motor unit. This is secondary to reinnervation of denervated individual muscle fibers due to collateral sprouting from nearby axons (Kugelberg et al. 1970). However, early in the denervation-reinnervation process there is a rearrangement of the muscle fibers within the motor unit territory resulting in a less random distribution of the motor unit muscle fibers. This rearrangement takes place before fiber type grouping becomes manifest. We have previously designed a planimetric model to assess the randomness of the muscle fiber arrangement in single motor units (Edström & Larsson 1987): 1) The motor unit territory was divided into eight sectors. 2) The number of muscle fibers per sector was counted. 3) The number of muscle fibers in each sector could be predicted on the basis of the size of the sector in relation to the size of the unit territory and the total number of muscle fibers in the unit assuming a random arrangement of the fibers within the motor unit territory. 4) The average difference between the measured and predicted number of muscle fibers was calculated in the different sectors and corrected for variations in total fiber number between different animals. The method was used to detect signs of denervation-reinnervation in ageing skeletal muscles (Edström & Larsson 1987). The aim of this study was to develop a more sensitive method to detect rearrangements of the muscle fiber distribution within a motor unit.

Methods

Single motor units were functionally isolated by microdissection of ventral root filaments. The muscle fibers in the unit were identified via glycogen depletion in muscle cross-sections incubated with an uridine diphosphate glucosyl transferase medium (Sasse 1966) prior to the PAS staining to enhance the contrast between glycogen depleted and non-depleted muscle fibers (Pool, Donselaar & Griep 1978). The coordinates of each glycogen depleted muscle fiber in the motor unit were identified with the aid of a digitizer connected to a microcomputer (MOP-Videoplan). The distances from each motor unit fiber to the rest of the fibers in the unit were measured and plotted. A random arrangement of the motor unit muscle fibers should give a gaussian distribution of the individual distances. The standard deviation of the average corrected distances has been used as an index for the randomness of the muscle fiber arrangement.

Results

Single motor unit muscle fibers from the fast-twitch tibial anterior and the slow-twitch soleus muscle of young (3-6 months) and old (20-24 months) rats have been compared. As can be seen from the figure there is a less random distribution of the muscle fibers in an old (B) slow-twitch soleus motor unit compared with the unit from a young (A) animal.

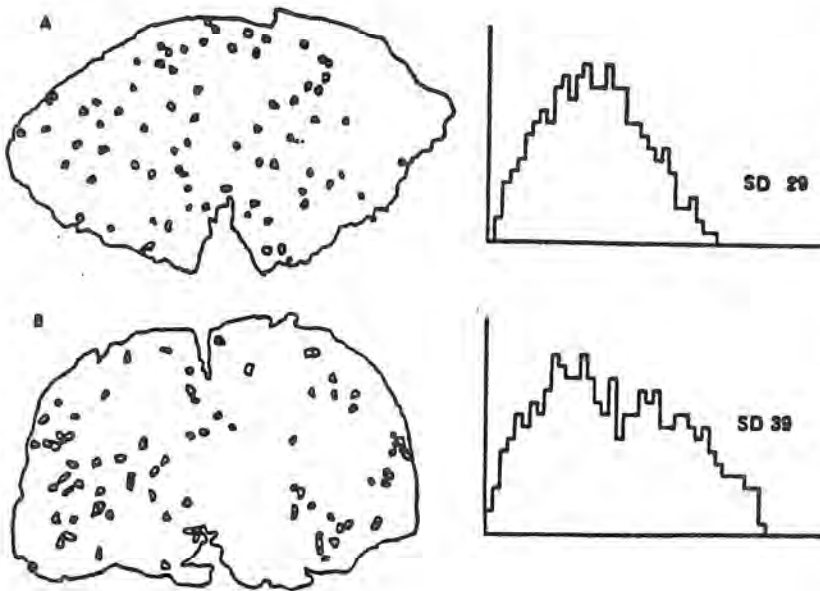


Figure. Glycogen depleted fibers in two slow-twitch motor units in the soleus muscle from a young (A) and an old (B) animal. A histogram of the individual distances between the fibres are presented together with the standard deviation (SD). A total of 5776 and 7569 distances are plotted for the young and the old animal, respectively.

A total of 60 glycogen depleted and enzyme-histochemically identified single fast- and slow-twitch motor units from young and old rats are presently being analyzed with this technique.

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THE EFFECTS OF AGE ON CONTRACTILE, MORPHOMETRICAL AND ENZYME-HISTOCHEMICAL PROPERTIES OF THE RAT SOLEUS MUSCLE.

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INTRODUCTION

The mechanisms behind the profound differences reported in contractile and morphological characteristics of skeletal muscle between young and old mammals remain obscure and various hypotheses have been proposed. The present study was undertaken with the aim of elucidating the mechanisms behind the age-related decrease in maximum force, the decreased speed of contraction and the decrease in type II (fast-twitch) muscle fibre proportion in old rat soleus muscle.

MATERIALS & METHODS

Male Wistar rats, divided into 8 different age groups ranging from 2 to 25 months, were used. Animals that were sick or showed gross pathological organ changes were excluded from the study. The sciatic nerve was supramaximally stimulated and the contractile properties of the soleus muscle were recorded. The muscle was then carefully dissected free, clamped at its optimal length, as determined from the maximum twitch force, and frozen in freon chilled with liquid nitrogen. The muscle was cut at the motorpoint in serial 10 μ m thin cross-sections and stained for myofibrillar ATP-ase after acid preincubation (Brooke and Kaiser 1969) or formaldehyde fixation (Hayashi and Freiman 1966). Total muscle fibre number, fibre type proportions and fibre sizes were measured.

Means and standard deviations were calculated from individual values by standard procedures. One-way analysis of variance (Oneway) was performed in order to evaluate the significance in age-related trends. Different age-groups were compared using Tukey's honestly significant difference. A p-value <0.05 was accepted as significant.

RESULTS

The total fibre number was not significantly changed between 2 and 12 months, but showed a decrease in subsequent ages. The absolute number of type I (slow-twitch) fibres increased up to the age of 7-8 months in parallel with a decrease in the type II fibre number. As a result, the type I fibre proportion increased from 71 $\pm$ 7% in the 2-month-old rats to 97-99 $\pm$ 1-4% in animals 7-8 months or older. Fibres of all types increased in size between 2 and 7-8 months. In older animals the type II fibres became successively smaller and more angulated. This was more marked among type IIC fibres compared to type IIA. The type I fibre size tended to decrease in animals 16 months and older, but the decrease was significant only at the age of 23 months or older.

Contraction time was significantly increased only in animals 20 months or older, i.e. at a time when there was no further change in fibre type proportions. The half-relaxation time showed a tendency towards a prolongation throughout the whole age-span, although it was significant only between the 3-4- and 20-22-month-old groups. The force of a completely fused tetanic contraction declined with age. However, when related to the cross-sectional fibre area (i.e. the specific tension), no

significant change was seen.

DISCUSSION

The age-related decrease in type II fibre number could be due to either a selective denervation of these fibres or to a transformation of these fibres into type I, as has previously been reported for rat soleus muscle during development (Karpati and Engel 1967, Kugelberg 1976). If the former alternative was true, then the denervated type II fibres must have been reinnervated by slow-twitch motor units since no signs of fibre loss were seen at these ages. This is unlikely, since the innervation ratio has been shown to be unaltered in the soleus muscle between 5 and 34 weeks (Kugelberg 1976), thus excluding reinnervation at these ages. Alternatively, a degeneration of type II fibres might have been compensated for by regeneration of muscle fibres or fibre splitting. However, signs of such changes were rare among young animals, which is in contrast to old age where fibre splitting was a prominent phenomenon. In the present study, the decrease in type II fibre number occurred prior to any detectable fibre loss, but was paralleled by an increased number of type I fibres, thus supporting the hypothesis that the age-related loss of type II fibres constitutes a continuation of the transformation process occurring during development.

The increase in contraction time was observed at ages where there was no further change in fibre type proportions and is thus related to a qualitative change in the contractile apparatus rather than to a loss of fast-twitch fibres.

The age-related decline in tetanus force together with an unaltered specific tension indicate that the force decline is mainly due to a quantitative loss of contractile material and that the capacity of the remaining contractile machinery to generate maximum tension is kept intact.

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A COMPARISON OF TECHNIQUES USED BY ELITE MALE AND FEMALE GYMNASTS PERFORMING A HANDSPRING VAULT

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Introduction

The handspring vault (Figure 1), one of the most difficult vaults in the past, is seldom used in today's major competitions. Yet, it is regarded as one of the most important fundamental skills in vaulting because it contains important elements which provide the foundation for learning complex vaults more safely and efficiently. This is one of the reasons that the handspring vault was chosen as a compulsory vault for women in the 1964 Olympics. It was also the basis for the cartwheel on - $\frac{1}{4}$ turn off compulsory vault for the 1972 Olympics, which was the foundation for the $\frac{1}{2}$ turn on - $\frac{1}{2}$ turn off compulsory vault for the 1980 Olympics, which was in turn the basis for the Tsukahara vault for the 1988 Olympics. Due to heavy emphasis on difficulty and originality in today's gymnastics, gymnasts tend to pursue more complex vaults rather than spend time mastering the basic handspring vault. This incorrect approach, in addition to the lack of basics, have been prime causes for vaulting injuries, which are among the most serious in gymnastics. Such injuries could be minimized by a better understanding of the important mechanical factors governing successful performance of this fundamental vault and then applying this knowledge to the development of proper training progressions. The purpose of the study was to compare techniques used by men and women performing handspring vault at the 1987 Pan American Games. Techniques used by male subjects have been described in the previous article by Takei (in press) and were compared with those used by women.

Methods

The subjects were 24 female gymnasts participating in the 1987 Pan American Games. A high-speed 16 mm motion-picture camera was used to record the vaulters' performances during the actual competition. A projection head and a digitizer linked on-line to a microcomputer were used to analyze the films. The x- and y-coordinates of 21 points, defining a 14-segment model of the body, were recorded for each frame analyzed. The data were then smoothed using a second-order polynomial least-square approximation to five data points, as described by Wood (1982). A deterministic model was developed to identify the mechanical factors for the subsequent analysis. The average forces exerted during the support phases were determined using the impulse-momentum relationships. The angular momentum about a transverse axis through the center of gravity (CG) of the gymnast during the flight phase was calculated, as described by Hay et al. (1977) and the segmental moment of inertia data of Whitsett (1963).

Results and Discussion

For women, means for the average horizontal and vertical forces exerted on the gymnast were 1.7 and 4.6 times the body weight on the board and 0.8 and 1.1 times on the horse, respectively. Means of the normalized angular momentum of pre- and postflights were 0.53/s and 0.33/s, respectively. Significant correlations indicated (a) large horizontal velocity (V_H) and distance of the hurdle; (b) large vertical velocity (V_V) at touchdown (TD) on the board, a large change of V_V during board support, and a shorter time of board contact; (c) large V_H and V_V at takeoff (TO) from the board; (d) shorter time of horse contact; (e) large V_H and V_V at TO from the horse; (f) high body CG at TO from the horse; and (g) large distance and height and a longer time of postflight were important determinants for successful results. Comparison of findings in the present study with those of men indicated despite a significantly greater V_H at TO from the floor for men, length of hurdle for men and women did not differ significantly when it was adjusted for standing height of the subjects. Women achieved significantly smaller gain of V_V on board and, consequently, significantly smaller V_V at TO from the board and TD on the horse than men. Although the angular momentum (L) of preflight was significantly greater for men, there was no significant difference between them when it was adjusted for height and mass. Women (a) contacted the horse with significantly smaller V_H and V_V ; (b) had significantly longer time of horse contact; (c) had significantly smaller average vertical and horizontal forces and impulses exerted during horse contact; (d) had significantly smaller gain of V_V and loss of L on horse; and (e) departed from the horse with significantly smaller V_H and V_V than men (Table 1). The significantly smaller V_H and V_V at TO from horse for women resulted in significantly less height, distance, and time of postflight, even after adjustment for the difference of horse height.

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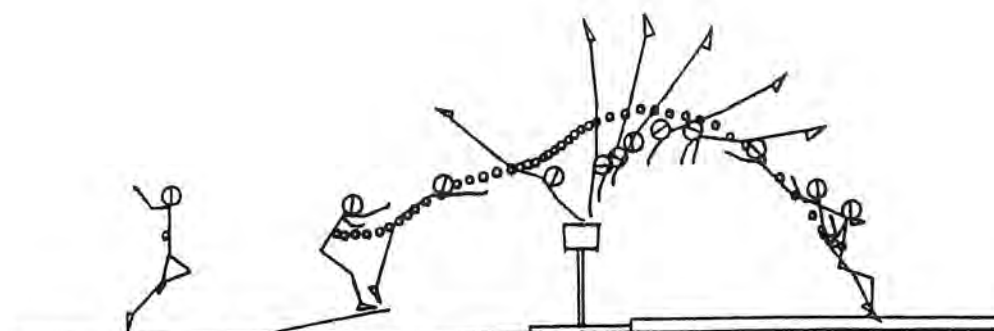


Figure 1. Selected positions and the trajectory of the center of gravity of a subject performing handspring vault

Table 1

A Comparison of Score, Mass, and Height of Subjects and Selected Mechanical Factors in Performing Handspring Vault

	Men (N = 40)	Women (N = 24)	Mean Diff (M - W)	t	p
Score, pts.	9.40 (0.27)	9.37 (0.22)	0.021	0.34	.733
Mass, Kg	61.93 (6.50)	47.96 (7.22)	13.97	7.77	<.001
Height, m	1.66 (0.07)	1.55 (0.07)	0.11	6.55	<.001
Time, s					
Board Support	0.13 (0.01)	0.14 (0.01)	-0.01	-1.15	.257
Preflight	0.21 (0.05)	0.24 (0.04)	-0.03	-1.93	.059
Horse Support	0.22 (0.03)	0.25 (0.03)	-0.03	-2.96	.005
Postflight (TPST)	0.78 (0.09)	0.64 (0.07)	0.14	7.15	<.001
TPST (ADJ)*	0.78 (0.09)	0.67 (0.07)	0.11	5.55	<.001
Displacement of CM, m					
Preflight Distance (DPRE)	1.09 (0.21)	1.07 (0.19)	0.02	0.39	.698
DPRE/(Ht)**	0.66 (0.13)	0.69 (0.13)	-0.03	-1.13	.265
Postflight Distance (DPST)	2.78 (0.42)	1.99 (0.31)	0.79	8.44	<.001
DPST (ADJ)/Ht	1.67 (0.27)	1.35 (0.21)	0.32	5.27	<.001
Postflight Height (HPST)	0.29 (0.13)	0.12 (0.07)	0.17	6.56	<.001
HPST/(Ht)	0.17 (0.08)	0.08 (0.05)	0.09	6.03	<.001
Horizontal Velocity, m/s					
Board Takeoff	5.19 (0.42)	4.61 (0.36)	0.58	5.82	<.001
Change on Horse	-1.62 (0.42)	-1.50 (0.37)	-0.12	-1.25	.216
Horse Takeoff	3.56 (0.37)	3.11 (0.37)	0.45	4.77	<.001
Vertical Velocity, m/s					
Board Takeoff	3.68 (0.23)	3.34 (0.24)	0.34	5.54	<.001
Horse Touchdown	1.79 (0.51)	1.26 (0.44)	0.53	4.40	<.001
Change on Horse	0.50 (0.35)	0.20 (0.42)	0.30	3.04	.004
Horse Takeoff	2.29 (0.60)	1.46 (0.49)	0.83	6.10	<.001
Angular Distance in Flight, °					
Preflight	67.5 (12.0)	70.6 (12.0)	-3.1	-0.98	.334
Postflight	159.4 (17.5)	143.5 (13.4)	15.9	4.08	<.001
Angular Momentum, Kg ² /s					
Preflight	94.8 (17.5)	61.9 (14.1)	32.9	8.25	<.001
Change on Horse	-50.1 (15.4)	-24.4 (10.0)	-25.7	-8.10	<.001
Postflight	44.7 (8.1)	37.5 (6.8)	7.2	3.81	<.001
Average Horizontal Forces, N					
On-Board	-1158 (238)	-819 (201)	-339	-6.07	<.001
On-Horse	-480 (143)	-313 (104)	-177	-5.36	<.001
Average Vertical Forces, N					
On-Board	2970 (335)	2175 (333)	795	9.23	<.001
On-Horse	761 (148)	521 (138)	240	6.56	<.001

Note: Standard Deviations are included in parenthesis.

*(ADJ) = Adjusted for the difference of horse height. **(Ht) = Standing height of the subject

KINEMATIC ANALYSIS OF WHEELCHAIR PROPULSION FOR THREE SPEEDS OF PROPULSION

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INTRODUCTION

Recently, a great deal of interest has been expressed concerning the factors affecting wheelchair propulsion. Many investigations have examined wheelchair propulsion techniques using both a lever arm (Brubaker et al., 1983; Cerquiglioni et al., 1981) and a pushrim (Higgs, 1984). The lever arm propulsion method has been shown to be more efficient than the pushrim method (Engel and Hildebrandt, 1974; Brubaker et al., 1983); however, lever arm propulsion is not permitted in wheelchair racing events. Consequently, the studies on pushrim propulsion techniques remain a major biomechanical research interest. Sanderson and Sommer (1985) examined the pattern of propulsion by paraplegics using the pushrim method and found considerable differences in the style of propulsion. The physiological and biomechanical parameters measured did not show any superiority between the styles of propulsion.

The purpose of this investigation was to examine the kinematics of the pushrim technique of wheelchair propulsion for two paraplegics at three speeds.

METHODOLOGY

Two paraplegics male long distance racers volunteered for this investigation. The level of neurological lesion was different for subjects 1 and 2 (T12-L1 and T8 respectively). To simulate wheelchair propulsion in the laboratory, the wheelchair was mounted on rollers. After a "warm-up" period the three speeds of propulsion were determined for each subject. The subjects were asked to propel their wheelchair until they felt that maximal velocity was reached. Three trials were recorded in order to obtain their maximal speed. The three speeds used in the investigation were 60, 80 and 100% of the subjects maximal speed of propulsion.

For each speed of propulsion, the subjects were filmed at 50 Hz with a high speed camera (Locam) placed perpendicular to the sagittal plane of the subject. Three trials were recorded for each speed. The segmental linear and angular displacements, velocities, accelerations, and momentums were obtained for a complete wheelchair cycle by processing the digitized cinefilm coordinates (BIOMECH, University of Waterloo). All data was time normalized and then ensembled averaged over the three trials.

RESULTS AND DISCUSSION

For both subjects, a decrease in the amount of time spent in the contact phase and an increase in pushing frequency was found as the speed was increased over the three trials (60, 80 and 100%). This result is in agreement with Van Der Woude et al. (1988) and Walsh et al. (1985) and suggests that the time spent in the contact phase was not a critical factor in the production of propulsive forces.

Considerable differences were observed in the pushing style of each subject. An analysis of hand movement revealed that subject 1 used a circular pushing style while subject 2 used a pump arm action. Steadward (1979) advised against the pump arm action for long distance racers but he did not provide any biomechanical evidence to support his view. The peak angular velocities for flexion and extension of the shoulder, elbow, and wrist were found to be higher for the circular pushing style (subject 1) at each speed of propulsion. These results contradicted Sanderson and Sommer (1985) who found that the peak angular velocities for flexion and extension of the shoulder and elbow joints were higher for the pump arm action. The peak angular velocities for the upper limb joints did not provide the information on the changes in the angular velocities for both phases of propulsion. Sanderson and Sommer (1985) claimed that a pump arm action would necessitate a more abrupt change in the hand direction; therefore, the shoulder and elbow joints would be subjected to more deceleration at the end of the contact phase and acceleration at the start of the recovery phase. This was observed in our investigation. The results showed that the changes in the angular velocities for the joints of the upper limbs were higher for subject 2 than they were for subject 1.

The motion of the trunk was found to vary across speeds and between subjects. Subject 2 exhibited a trunk motion opposite to that of subject 1. In the first part of the contact phase subject 2's trunk was brought forward while in the last part of the contact phase the trunk was brought backward as the elbows were extended. This suggests that since subject 2 had a neurological lesion at T8 he needed to use his upper limbs to bring the trunk backward to compensate for a weaker abdominal musculature. The pattern of the trunk angular momentum was different for both subjects due to the weakness of abdominal muscles.

CONCLUSIONS

This investigation showed that as the speed increased across trials (60, 80, and 100%), the contact phase decreased and the pushing frequency increased. The subjects were found to have considerable differences in the pushing style. The variation in forward and backward motion of the trunk was suggested to be related to the level of neurological lesion.

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VALIDATION OF A MUSCULOSKELETAL MODEL TO PREDICT FORCE IN HUMAN QUADRICEPS MUSCLES

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INTRODUCTION

One of the most perplexing problems facing the biomechanics researcher is the redundant nature of the musculoskeletal system. The human knee joint is complex and is crossed by a large number of muscles so that the problem of solving for individual muscle, ligament, and bone contact forces is indeterminate. As a result of the indeterminacy problem of the knee joint, there have emerged various mathematical models. These models, however, have lacked the criterion measures needed for direct validation. In the studies that involved an in vivo muscle force comparison criterion (i.e. Komi et al., 1987) the muscle forces were obtained via a buckle force transducer that was clipped onto the tendon. This method may produce errors in the muscle force measure since the tendon orientation and the tendon-muscle length are altered by the buckle, thereby possibly influencing the normal production of force.

This purpose of this study was to validate a musculoskeletal knee model for the prediction of quadriceps muscle forces with in vivo force measurements as the comparison criteria.

METHODS

Muscle Model

In order to predict the quadriceps forces during knee extension and flexion a musculoskeletal model was developed to determine the rectus femoris, vastus lateralis, vastus medialis, and vastus intermedialis force contributions. This model was adapted from the work of Dowling and Norman (1988) in which the muscle forces at the elbow joint were predicted from EMG and kinematic data.

The force of each of the four muscles was predicted from the EMG using submaximum isometric contractions for calibration and modified by two unitless ratios representing the force-velocity and length-tension relations. Muscle length was determined by integrating the moment arm data as a function of joint angle from the data of Herzog (1985). The contractile component length was altered by the stretch of the series elastic component. Since the force in the patellar tendon is not equal to the force in the quadriceps tendon, due to the intervention of the patella, the four predicted muscle forces were summed and modified by a fifth order polynomial fitted to the data of Van Eijden, et al. (1986) to predict the patellar tendon force.

Validation

The musculoskeletal force model was validated by comparison with the in vivo human patellar tendon (PT) forces reported by Lamontagne et al. (1988). As well as supplying information on PT force, knee joint moments, knee joint angles, and also EMG from rectus femoris, vastus lateralis, and vastus medialis were also provided.

The PT forces were obtained by bonding a mercury strain gauge to the proximal midline of the patellar tendon and recording the strain gauge deformation throughout the range of motion. The knee moment and knee joint angle data, over the full range of motion, were obtained from

a CYBEX isokinetic dynamometer at 60 degrees per second. The raw EMGs from the quadriceps muscles were rectified and filtered to provide the linear envelopes needed for the model.

The resultant quadriceps force (sum of the four muscle forces from the model) was compared to the PT force for each of three experimental trials on the basis of the root mean square error (RMSE), the percent RMSE, the coefficient of determination (r^2), and the maximum difference.

RESULTS AND DISCUSSION

The root mean square errors between the model predictions and the measured patellar tendon forces ranged from 41% to 52% with r^2 values ranging from 0.56 to 0.68. These predictions were not as good as other investigations (i.e. Dowling and Norman, 1988) but, on closer examination of the data, it was discovered that there were two bursts of force during an extension/flexion movement. The first burst of force was expected during the extension phase and was well predicted by the model. The second burst unexpectedly occurred during the flexion phase and was not well predicted by the model due to the lack of EMG activity. It was hypothesised by the authors that this second burst of force was due to passive sources not included in the model. This was supported by the smaller percent RMSE (19% to 24%) and larger r^2 values (0.71 to 0.91) when only the extension phase was analysed. The true nature of the effects of the passive forces on patellar tendon forces predictions require further investigation.

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BIOMECHANICAL FACTORS AFFECTING ACCURACY IN PISTOL SHOOTING

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The research, sponsored by the Australian Sports Commission, was designed to investigate which of selected biomechanical factors most influenced accuracy in elite calibre pistol shooting. The major purpose for the research was to identify which parameters should be incorporated into a biofeedback system which was to be designed to assist elite shooters in training. The 16 subjects participating in the study were elite shooters and members of the Australian Pistol shooting squad. The group was comprised of seven senior males, seven senior females and two juniors. The biomechanical parameters examined included: Body Sway, Pistol Movement, Variation in Pistol Alignment, Mean Grip Pressure, Variation in Grip Pressure and Sighting Time. These parameters were established as important contributors to accurate shooting from a literature review and discussions with leading Australian shooters and coaches. A Walther CO₂ air pistol was chosen to be used in the study as representing the various competition pistols.

With the exception of Sighting Time, the parameters investigated were examined in the time period one second prior to the shot being fired. The one second interval was chosen by the researchers as this period was considered to be crucial to accurate shooting. The criterion variable, the measure of shooting accuracy, was defined to be the displacement of the Fall of Shot from the centre of the target. The research also examined whether movement of the aim point of the sights on the target (Aim Point Excursion) was related to the criterion measure. Many coaches believe that shooting accuracy is closely related to maintaining a small aiming pattern on the target prior to the shot being fired. Where possible the parameters were also examined with respect to their components, displacement in the vertical and horizontal directions. Movement or variation in a parameter was computed as the excursion, or total distance travelled, over the one second time period. Body Sway was recorded as the movement of the shooters' centre of pressure using a Kistler force platform. Pistol Movement and Variation in Pistol Alignment were computed from the movement of two points on the pistol barrel, digitised from high speed film. Mean Grip Pressure and Variation in Grip Pressure were measured from strain gauges mounted on cantilevers in the pistol grip. Aim Point Excursion and Sighting Time were obtained from the Noptel Ky Aim-point detection system. The Noptel Ky system consisted of a laser gun attached to the pistol barrel, an optical target located below the actual shooter's target and a microprocessor for collecting and processing the information. The time of pistol discharge was obtained from a microphone attached to the pistol. The relevant parameters were sampled as electrical signals at 100 hertz by computer using a 12 bit analogue to digital board.

The project specifically aimed at identifying the relationship between the various biomechanical parameters and shooting accuracy. All 16 subjects were tested during a data collection session. Each subject fired up to 50 shots of which 20 were monitored by the research equipment. The data on at least 10 of these shots per subject was used in the analysis. A single score was computed for each parameter per

subject, by collapsing the data over each subject, before analysing the relationship between the various biomechanical parameters and shooting accuracy. The analysis was also performed on each subject's data to identify if the previously established relationships, obtained from the group data, held true for individual subjects. Five of the subjects were retested on two further occasions to identify if the relationships previously established for those shooters held true over a period of several months.

Correlation and stepwise regression statistical analyses were utilised to identify which of the parameters most influenced accuracy. Table 1 provides the mean value, standard deviation and the correlation coefficient between the particular variable and Fall of Shot for the group of shooters and its significant level. A stepwise regression analysis technique was used to identify which biomechanical parameters most contributed to accurate shooting. The analysis revealed that Body Sway followed by Pistol Movement most influenced accuracy (Mult.R=0.73 $p=0.01$ $n=16$). The correlation coefficient for Body Sway versus Pistol Movement ($r=0.2$ $p=0.1$ $n=16$) was not significant, indicating that the Pistol Movement was not related to the Body Sway. A correlation analysis of the vertical and horizontal components of shooting accuracy revealed that Body Sway influenced the vertical Fall of Shot ($r=0.63$ $p=.01$ $n=15$) whereas Pistol Movement influenced the horizontal Fall of Shot ($r=0.61$ $p=0.01$ $n=16$). This analysis also disclosed that Variation in Pistol Alignment was also highly related to the horizontal Fall of Shot ($r=0.61$ $p=0.01$ $n=15$). Sighting time was negatively correlated with the criterion variable Fall of Shot ($r=-0.56$ $p=0.01$ $n=16$). This indicated that for the group of shooters tested, the greater time that the shooter spent aligning the pistol sights on the bull's eye, the less was the displacement of the shot from the target's centre. Although there was a significant relationship between the Fall of Shot and Aim Point Excursion, this relationship was not strong ($r=0.20$ $p=0.01$ $n=271$). Although the relationship did exist, the results of the analysis implied that other factors were more important in achieving accuracy in shooting. No relationship was found to exist between Aim Point Excursion and the Fall of the Shot for the time interval 0.5 seconds after the shot, generally regarded as the period of follow through. The analysis of the individual results revealed that the factors affecting shooting accuracy vary considerably between shooters. Contrary to the group results, one shooter was affected by Mean Grip Pressure and Variation in Grip Pressure. The repeated tests indicated that while different shooters varied in the importance of the specific biomechanical parameters to affect shooting accuracy, there was also a variation for the same shooter at different testing times.

Table 1

<u>VARIABLE</u>	<u>VALUE</u>	<u>SD</u>	<u>CORR. COEF</u>	<u>SIGNIFICANCE</u>
Fall of Shot	12.5 mm	5.3 mm	1.0	0.000
Aim-Point Excursion	156 mm	50 mm	0.20	0.012
Body Sway	5.0 mm	1.1 mm	0.55	0.017
Pistol Movement	7.9 mm	1.7 mm	0.36	0.095
Var. in Pistol Align	2.0 deg	0.4 deg	0.29	0.149
Mean Grip Pressure	25.4 % of max	10.3 %	0.14	0.314
Var. in Grip Pres.	7.3 N	4.0 N	-0.21	0.222
Sighting Time	5.2 sec	2.1 sec	-0.56	0.012

MECHANICAL VERIFICATION OF THE EFFECTIVENESS OF THE FIRST STEP FOR QUICK CHANGE OF THE FORWARD RUNNING DIRECTION. T.Ohtsuki and M.Yanase.
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The authors reported previously (Ohtsuki et al. 1987, 1988) that (1) either the open or the cross step must be employed to change quickly the direction of progression during running forward, (2) when the open step is used as the first step for changing direction, the unexpectedly stopped moving target which the subject has been chasing can be caught in a shorter time. The present experiment was designed to verify the advantages of the open step over the cross step by using the ground reaction force measurements.

The force platform (Kistler 9281-B) was embedded in the center of the 4x5 meter concrete-paved square yard. Hi-speed VTR camera (SONY RSC1150, 60fps) was set at the height of 12.25 meter right above the center of the force platform. Two stimulus lamps (7.5 cm in diam.) were fixed at the end of the running course 8.8 meter away from the center of the force platform. The following three experimental tasks were prepared; (1) the subjects dashed from the starting line at their maximum speed and changed the running direction according to the stimulus indication (task ST). The starting line was pre-determined for each subject's physique and the subject started by always lifting left first so that the fourth step, i.e., right foot, was the first step for directional change and was on the force platform. Thus, when the stimulus indicated "right," the first step for change was "cross" type, and when the stimulus indicated "left," it was "open." These were named conditions STO and STC. (2) the subjects changed the direction by themselves with the fourth step on the force platform according to the specification previously given (task SF). No stimulus was given. This task also includes two conditions, i.e., STO and SFC. (3) the subjects dashed straight without changing the direction so that the fourth step was on the force platform (task CON). In tasks ST and SF, the equal number of trials was $6 \times 2 = 12$ for each condition CON, SFO, SFC, STO and STC. From the overhead VTR records, the trials in which the fourth step was not the first step for change judging from the comparison with the condition CON, and the trials in which the fourth step was not right on the force platform were omitted. Thus, seven trials for each condition were adopted for the analysis. Force records were stored in the magnetic tape data recorder and the mechanical impulse was obtained by electrical integrator.

Figure 1 shows the typical example of the three dimensional components of the ground reaction force. X indicates sideward component. Y indicates back-and-forth component and Z indicates the vertical component. The line of action of the component Y is paralleled with the designated direction of dash before arriving at the force platform. Figure 2 shows the mechanical impulses applied from the ground to the runner. For the components X and Z, O was larger than C in both SF and ST. For the component Y, O gave the larger braking impulse against the forward running than C in both SG and ST. Especially in ST, the superiority of O to C comes from longer duration of force application by the first step for change (Figure 3). As the mechanical impulse is proportional to the velocity change, it can be concluded that the open step can produce the greater angular change of the velocity vector toward the required new direction. Comparison between SF and ST tells that ST gave smaller braking impulse (component Y) than SF for both O and C, while no difference was observed in the sideward compo

for both O and C, while no difference was observed in the sideward component(X). Thus, in task ST, the subject cannot change the velocity vector of the center of mass as effectively as SF. These results suggest that when the players should change their running direction in response to the unexpected changes of situation, a considerable disadvantages will result depending on the type of the first step for directional change which the players cannot select by themselves.

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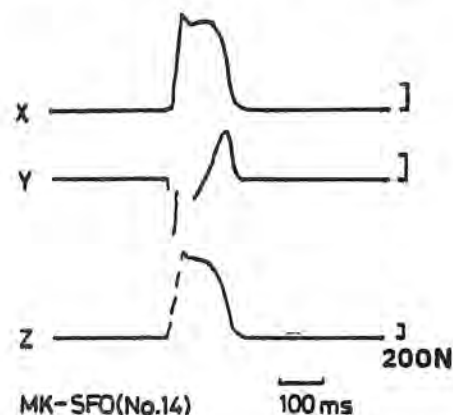


Fig. 1.

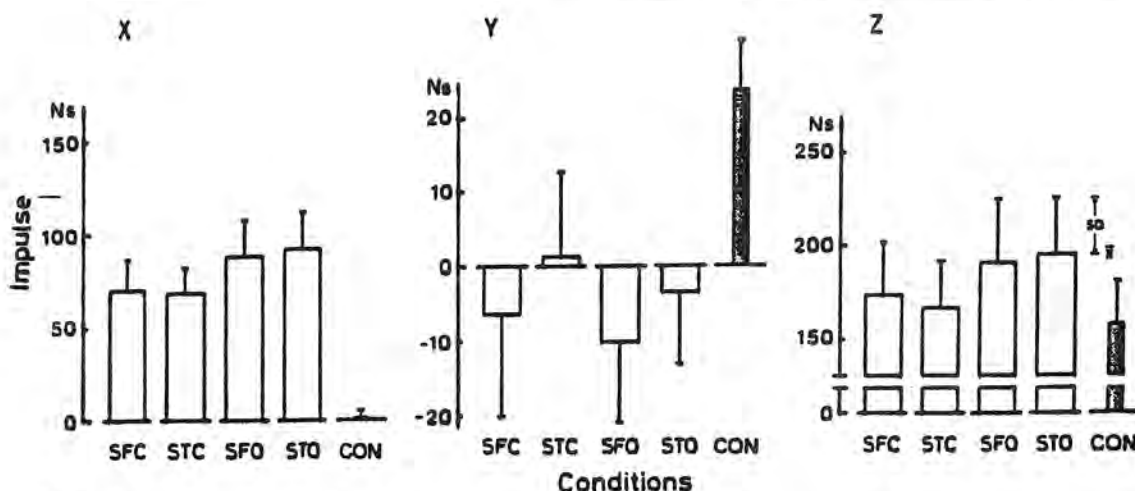


Fig. 2.

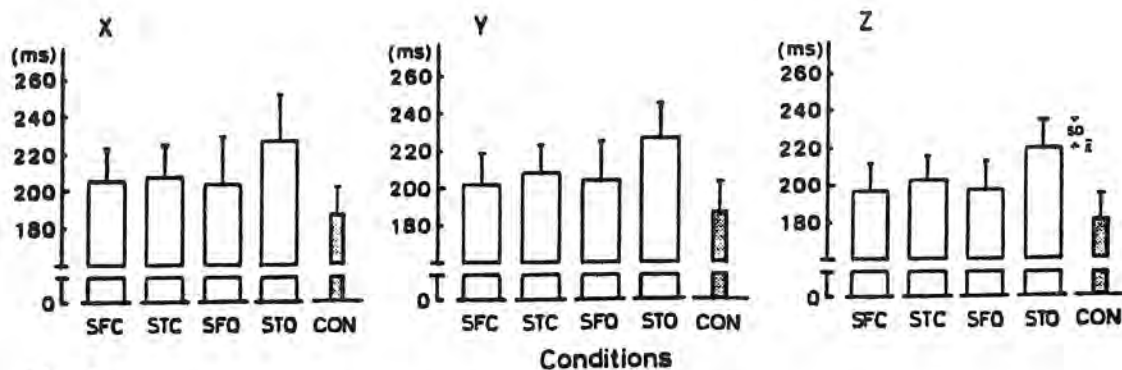


Fig. 3.

**Comparison Between Surface and Intramuscular
Wire Electrodes in Isometric and Dynamic Conditions**

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INTRODUCTION

Surface electrodes (SE) are commonly used in kinesiological investigations. They are quick and easy to apply, can be applied without medical supervision, give almost no discomfort and pain to the subject, and have a fair good reproducibility (Jonsson and Riechmann, 1968; Komi and Buskirk, 1970). SE are mainly used to record electromyographic (EMG) activity of superficial muscles and muscle groups.

Intramuscular wire electrodes (IWE) are also used in kinesiological investigations. This type of electrode is known to be very selective, according to their small leading-off area. IWE record a smaller number of muscle fibres than SE.

The purpose of this study is to evaluate the most appropriate EMG electrodes (SE vs. IWE) for isometric and dynamic conditions during an occupational cervico-brachial disorder (OCD) work task.

METHOD

Six normal male subjects performed two work conditions, isometric and dynamic. The isometric condition consisted of statically holding a known weight representing 50% of the maximal lifted weight for 2 s. The dynamic condition consisted of moving a known weight (15% of the maximal lifted weight) for a horizontal distance of 45 cm at a frequency of 40 displacements/min. The rhythm of this frequency was given by a calibrated metronome. The known weight was initially located 10 cm in front of the subject. While the subject executed the weight displacement tasks, surface and intramuscular EMG signal, and cinematographic data were synchronously collected for three trials.

Muscular activity was recorded with SE and IWE over/in the trapezius (upper part), middle deltoid, and the anterior deltoid. Pairs of silver-silver chloride electrodes (MEDI-TRACE) were placed, 20 cm apart, directly over the motor points of these muscles (Delagi et al.; 1975). IWE were implanted perpendicular to the skin and between the two surface electrodes (2 cm inter-electrode distance) for the three muscles. High input impedance (10 mega-ohms) differential amplifiers were used to record raw EMG signals after which were full wave rectified and filtered using a second order Butterworth filter with an 8 Hz cutoff frequency to produce linear envelope EMG. This EMG activity was then normalized (100% MVC).

Both work conditions were performed for 3 trials over two different sessions (two different days) in order to investigate test-retest reliability and day-to-day reliability. Both kind of electrodes (surface and intramuscular) were removed and re-inserted (intramuscular ones) for each session. The statistical models used were 2x2x3 Factorial Analysis (ANOVA) with repetitive measurements and paired t-tests.

RESULTS AND DISCUSSION

The test-retest reliability for an isometric ($r=0.75$; SE, $r=0.59$; IWE) and a dynamic task ($r=0.79$; SE, $r=0.55$; IWE) was good for both kind of electrodes (surface and intramuscular wire), which was also observed by Komi and Buskirk (1970) and Yang & Winter (1983). This test-retest reliability result was expected since the IWE and SE were not removed between the test-retest investigations (3 trials).

Day-to-day reliability was not as good as the test-retest reliability. The day-to-day reliability was better with SE ($r=0.92$) than IWE ($r=0.31$) in the dynamic condition. IWE signal was significantly different ($p<0.05$) between the two different sessions of testing, which was also observed by Komi and Buskirk (1970). The poorer reliability of the IWE could be explained by many factors. First, IWE are very fragile and they need to be handled with care during fabrication and insertion. Secondly, since IWE were re-inserted in the second session, it is very difficult to insert the wires in the same specific location and depth in the muscle as they were in the first session. This should be easily observable with the IWE since they have a very selective leading-off area (Jonsson & Komi; 1973, Perry and Beckey; 1981).

According to the statistical analysis, there was no significant differences ($p<0.05$) between SE and IWE in both dynamic and isometric conditions. IWE recorded more EMG activity than SE in both work conditions (isometrics and dynamics). This difference of EMG signal magnitude can be explain by many factors. SE recorded more EMG activity from the superficial motor units than the deeper ones (Milner Brown & Stein, 1975), the low pass filter effect created by the tissue between the electrical innervation and the SE electrode (Perry and Beckey; 1981), and IWE records muscular activity from a smaller area than surface electrodes.

In conclusion, EMG signals from both kinds of electrodes confirmed that electromyography is reliable when the electrodes are not removed or re-inserted in the same session. When the electrodes are removed or when testing in a different session, surface electrodes revealed to be more reliable than intramuscular wire electrodes.

ACKNOWLEDGEMENTS

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Data Acquisition and Analysis System on Micro Computer for Biomechanical Studies

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INTRODUCTION

With the recent advances in micro computer technology, processing speed and mass-storage capacities have increased immensely. Consequently, the use of these machines in real time data acquisition (Eldin, Alhaidar and Ali 1986; Malcolmes-Lawes 1987A, 1987B, 1987C; and Chew and Ha, 1988) has also increased in biomechanics research; however, the lack of micro-computer software to fulfill many of the biomechanical testing requirements (i.e., multi frequency channel data collection, real-time graphic display, synchronization of many devices, integrated data analysis, user friendly interface, etc.) are not available commercially. Therefore, biomechanic researchers must develop their own software in order to collect and analyze data in biomechanical settings.

A micro computer data acquisition and analysis system (BIOAD) has been developed to address these biomechanical experiment requirements.

DATA ACQUISITION AND ANALYSIS SYSTEM

Equipment

The data collection-analysis package (BIOAD) uses a MS-DOS based operating system. For data collection a Compaq 386 (16 MHz) micro computer is used with a Tecmar data collection board (LabMaster, 40 KHz, PGL). The data acquisition and analysis system requirements are a MS-DOS compatible computer.

In order to connect all input and output (I/O) lines to the computer an interface box was constructed to make full use of the data collection boards capabilities. This included 16 analog input channels, 2 analog output channels, 24 digital input-output lines, 5 programmable timer/counters, external clock line, and abort switch. Through the use of this box the computer can read analog and digital (A/D) data, control devices (LED indicators, camera, switches, etc.), time or count events, and stop data collection at any time.

Data Collection

BIOAD was designed to allow the user to configure the data collection process, enter testing information, collect data, postview trial data, save data, and test the data acquisition board. The configuration of the collection process includes selection of up to 16 A/D data channels at different frequencies, selection of different gains for each channel, counting of camera pulses, triggering on an external event, and selection of which channels to be displayed. Testing information can be entered, saved and retrieved on the subject, film, and trial observations. The data can be collected using real time graphic display or numerical display and graphically reviewed on the computer monitor after data collection on superimposed or separate axes. The data is stored on disk in an organized manner using unique algorithms to optimize memory and disk space.

The data collection process operates such that when a timing pulse is received (internal or external) all the selected channels are processed at the fastest possible speed. This eliminates much of the time lag that occurs in data collection between channels with systems that rely on the timing pulse to read each channel. Once collected, the data is stored sequentially in one buffer, making use of the minimum amount of RAM. The data can then be stored on disk in a compressed binary format. The data compression reduces the file size by down to 84%, depending on the signal, while taking a negligible amount of time for processing. It is desirable to simultaneously sample channels at different frequencies because of cases where a fast

sampling measure (EMG, etc.) is being sampled at the same time as a slower sampling measure (electrogoniometer, force platform). This way a single micro computer can run most multi device tests without synchronization problems.

A real time display is preferred since the data collection process can be examined for problems as the trial is progressing. The graphic output is in colour (16 colors, 800x560 resolution, all selected channels displayed on the same axis) or in black and white (all selected channels displayed on separate axes). This orientation allows the researcher to view a large scale image of the curves or to view smaller, individual images. To ensure that the data is collected in real time, collection is handled by an interrupt service routine, resulting in a possible lag in the real time display at very high sampling frequencies (1 channel 3000 Hz, 8 channels 1000 Hz show no discernable lag). It is also possible to view the data collection process numerically. This option is desirable when testing the LabMaster board prior to the data collection session or when a real time display is not required.

All user friendly screen editing facilities are designed in a multiple window format. This interface makes the data collection procedure more efficient since "set-up files" (containing information that would have to be entered each time-sampling rates, data, etc.) can be constructed, saved, loaded onto the screen, and edited when necessary.

Data Processing

The data analysis program was developed to read the data collection files directly. It is then possible to view information on the collected channels, experimental information, the film information, and comments. Eight of the possible 16 channels can be loaded for processing and the number of files to be graphically displayed on the screen can be selected. It is also possible to import an ASCII file or selected arrays from various ASCII files into the program.

Once the data is present multiple analyses can be done: Fourier analysis, power spectrum analysis, critically dampened single pass digital filtering, bias removal, slope removal, mean calculation, integration calculation, full wave rectification, normalization (to time, maximum amplitude, and/or a specific value), and ensemble averaging. Through the use of cursors these analyses can be performed on any section of the curve(s). The analysis package also takes advantage of the multiple window format user interfaces.

The output file from the analysis (ASCII format) is set up for direct input into various MS-DOS compatible graphics packages or for printing.

CONCLUSION

A user friendly, fast, and efficient data collection and processing package has been constructed for a micro computer system using unique algorithms. The integrated nature of the package such as collection, information and data storage, and processing) allows for a wide variety of applications and expansions in the biomechanical field.

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INSTANTANEOUS CENTRE OF ROTATION OF THE KNEE DURING HUMAN GAIT

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INTRODUCTION

Major attempts have been made to calculate the true centre of rotation (CR) or centroid of anatomical joints. The goal of researchers has varied from the assessment of knee joint pathologies (1), the design of polycentric knee joints (2) and the improvement of link segment models. The major technique that has been employed to date is that of Reuleaux (1,3-7), and several researchers have reported various forms of sensitivity analyses which has pinpointed inherent weaknesses of the approach (4-7). The Reuleaux method requires images of rotating segment at two different angular positions. The error in CR estimate increases drastically as angular motion decreases. The required orientation of the markers on the rotating segment is such that they most often must be close together, which further increases the error. Signal processing techniques have been suggested to reduce the error (5-7). More limiting is the protocol employed which was non-physiological: cadavers in many cases, or X-ray images while unloaded by gravitational and muscle forces, with one segment clamped.

THEORY

The technique used here is a Velocity-Radius method which has a relative ease of computation and yielded instantaneous CR's during normal gait (loaded and unloaded conditions). The method is shown graphically in Figure 1 and involves calculation of the relative motion of one segment relative to an adjacent segment. The linear velocity, $\bar{V}$, of a distal point on the relative rotating segment, its angular velocity, ω_z , and the radius of rotation, $\bar{R}$ are related by:

$$\bar{V} = \omega_z \times \bar{R} \quad \text{Eqtn. 1}$$

$$\text{or, } V_x = R_y \omega_z \text{ \& } V_y = -R_x \omega_z \quad \text{Eqtn. 1(a)\&(b)}$$

Both $\bar{V}$ and ω_z are known, thus the magnitude of $\bar{R}$ can be calculated. x_c and y_c can then be calculated at each instant in time as long as $|\omega_z|$ exceeds a threshold magnitude. In the case of the knee joint the threshold was set at .75 r/s which yielded reliable $\bar{R}$'s over about 80% of the stride period.

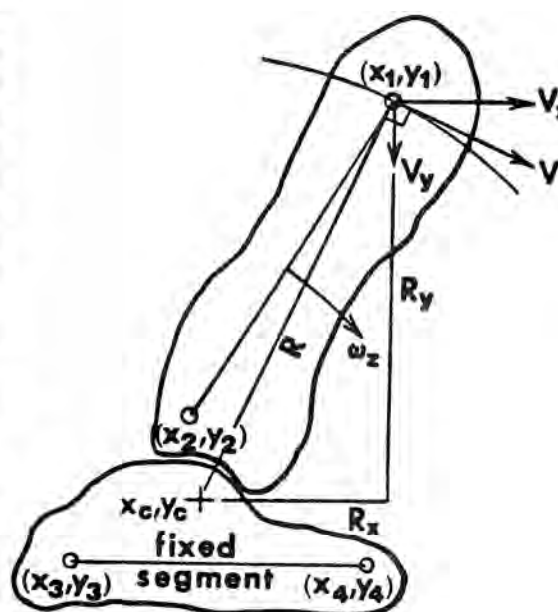


Figure 1

METHODOLOGY

Reflective markers one cm in diameter were attached to the following locations: heel, 5th metatarsal, toe, lateral malleolus, head of fibula, lateral epicondyle of femur and greater trochanter. Subjects' movement were recorded with a CCD camera and subsequently digitized at 60 Hz using a specially designed TV system. Raw x and y coordinates were digitally filtered at 6 Hz. The thigh segment for each frame was fixed on a vertical reference axis with the lateral epicondyle marker set at (0,0). V_x and V_y of the lateral malleolus and ω_z of

the leg segment were calculated. Using Equation 1(a) and (b) R_x and R_y were calculated to yield x_c and y_c . Each subject was analyzed for 10 repeat trials. The locii of x_c y_c were plotted over the four phases when knee $|\omega_z| > .75\text{r/s}$: K_1 - flexion during weight acceptance, K_2 - extension during mid-stance, K_3 - flexion during push-off and early swing, K_4 - extension during mid and late swing.

RESULTS AND DISCUSSION

The locus of CR varied slightly over the ten repeat studies for the same subject. The standard deviation of x_c and y_c varied from less than 1 mm when $|\omega_z|$ was high ($> 5\text{r/s}$) and increased to about 16 mm when $|\omega_z|$ decreased to 1 r/s. Figure 2(a) shows the locii of the CR during K_1 and K_2 phases, Figure 2(b) for phases K_3 and K_4 for one subject. All locii for a second subject were quite different.

These results are in total disagreement with a unique CR locii previously reported that was dependent on knee angle only and did not vary appreciably between subjects. The reason for such variability over the stride period can be attributed to the different muscle activity for each phase of gait. During K_1 the quadriceps are dominant for most subjects, but with a wide range of moments of force. During K_2 the moments reduce and for some subjects reverse. Then during K_3 gastrocs and quadriceps both act but differently for each subject, and during the unloaded swing phase (K_4) the hamstrings generally dominate. Thus each phase is characterized by a different angular velocity, different muscle patterns and during swing phase the articulating surfaces are not even in contact. Thus we should not expect the CR locii to be a function of knee angle alone and this study bears this out.

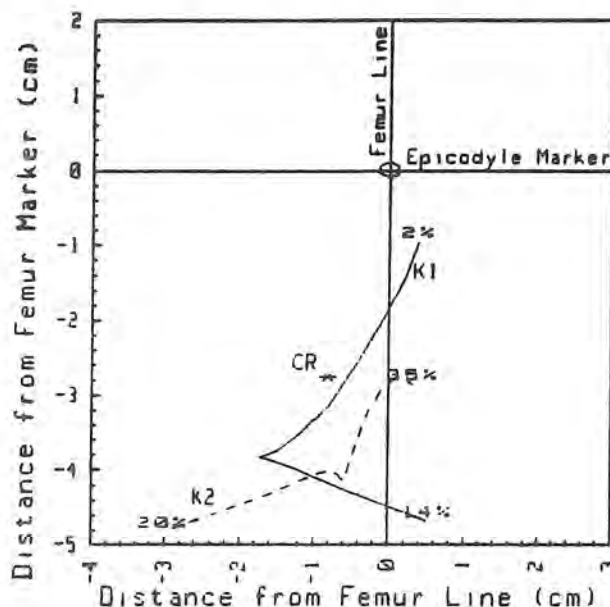


Figure 2(a)

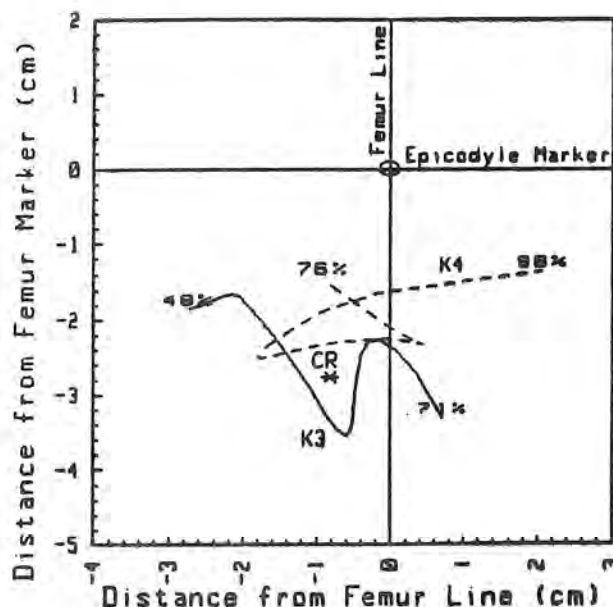


Figure 2(b)

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C OF P VS. C OF G CHANGES DURING QUIET STANCE AND VOLUNTARY SWAY IN FRONTAL AND SAGITTAL PLANES

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INTRODUCTION

Regulation of posture and balance during standing requires the precise control of the centre of gravity (C of G) about an equilibrium point within the base of support (BOS). The majority of research investigating balance control and the effects of various related pathologies have erroneously used centre of pressure (C of P) as the index representative of postural sway. Sway is a kinematic term and can only be quantified by the C of G and it has been shown (1-3) that the path of the C of P does not represent the C of G displacement. In the sagittal plane the C of P merely reflects the location of the reaction force due to the ankle moment, which, in turn is the response of the neuromuscular system to correct or cause an imbalance of the C of G. A similar interpretation can be made in the frontal plane where the C of P is an algebraic summation of the two frontal plane ankle moments. The purpose of this study was to assess the balance control system during quiet stance and voluntary postural sway by quantifying the C of P and C of G displacements in the frontal and sagittal planes.

METHODOLOGY

Two normal male subjects of similar height and weight were studied during quiet stance and voluntary sway about the ankle. The subjects stood with folded arms and the BOS was the foot length and distance between the lateral borders of the feet in the sagittal and frontal aspects respectively. Reflective skin markers were used to define a 13 segment model comprised of foot, leg, thigh, pelvis, abdomen-thorax, head-neck, arm and forearm-hand segments (4). Marker coordinates were collected using a CCD video camera interfaced with a PC frame grabber at the University of Waterloo Gait Laboratory which yielded coordinates with less than .5 cm rms noise. Kinematic data were collected over 7.16s at 60 Hz and low-pass filtered using a critically damped 2nd order double pass digital filter with a 3 Hz cutoff. Ground reaction forces were sampled at 500 Hz and low-pass filtered at 20 Hz. Ground reaction forces and the resultant C of P were used as the criterion measures against which the anthropometric model was compared. By introducing iterative changes to segment mass distribution and location the C of G modelling error was reduced. Error was quantified as the root mean square of the difference between estimated and criterion values such as horizontal forces and C of P when horizontal forces were zero.

RESULTS AND DISCUSSION

Figure 1 presents the relationship between C of P and C of G during quiet stance and voluntary, large amplitude ankle sway in both the sagittal and frontal planes. Although trends were similar the displacements were of lower amplitude in the med/lat direction during quiet stance and greater during voluntary ankle sway.

Sagittal sway during quiet stance reached ± 0.4 cm at frequencies of less than 2 Hz, while the C of P fluctuated more rapidly at amplitudes of $\pm .25$ cm greater than the C of G. In the frontal plane sway amplitudes reached $\pm .3$ cm with a $\pm .2$ cm C of P overshoot. During exaggerated voluntary ankle sway was displaced 12.0 cm in the sagittal and ± 9.0 cm in the frontal plane before returning to its equilibrium point. Throughout this movement, C of P remained in phase and overshoot sway by ± 1.5 cm and ± 2.0 cm in the sagittal and frontal planes respectively. Under these conditions, the frequency content of C of P was greater than the C of G's in both planes.

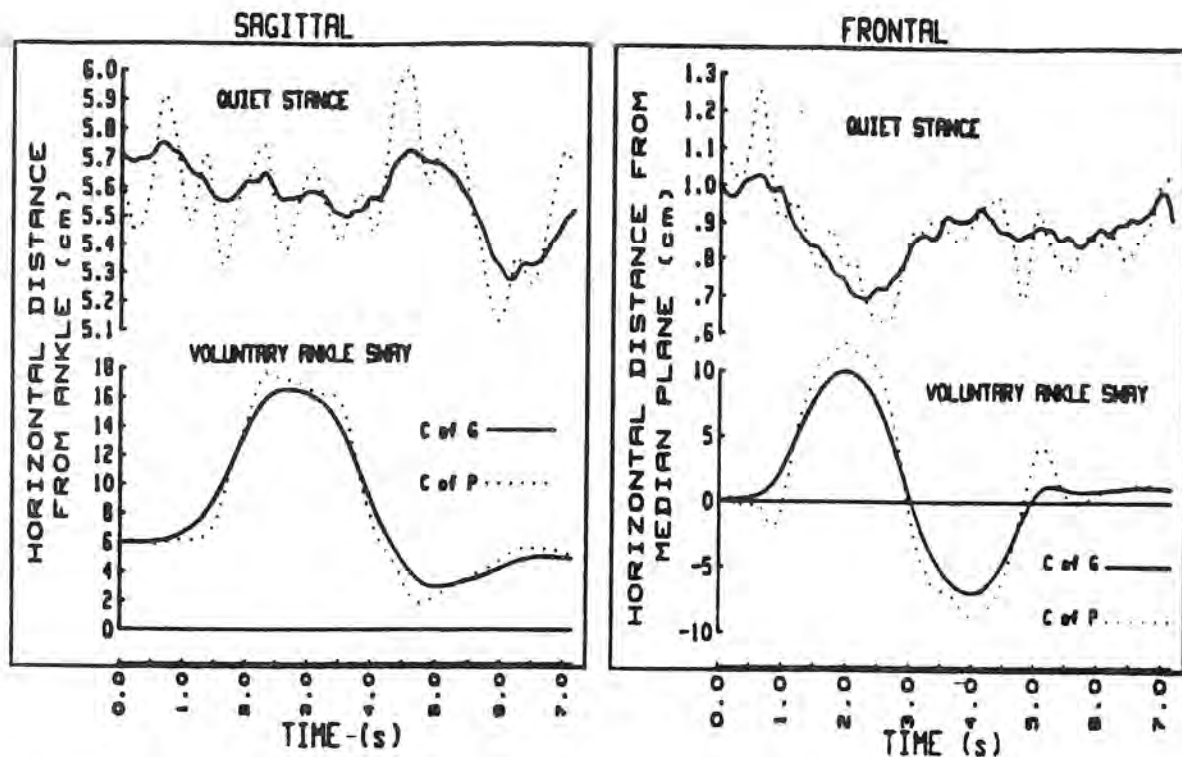


Figure 1: C of G and C of P plots in the sagittal and frontal planes during quiet stance and voluntary ankle sway.

If the body is modelled as an inverted pendulum the product of body weight and the difference between C of G and C of P is the controlling moment at the ankle. The controlling moment is therefore directly related to the body's angular acceleration ($M = \gamma I$), as was supported by these results ($R = .85$). Thus it appears that the CNS must be acutely aware of the trajectory of the C of G at all times.

CONCLUSION

The difference between C of P and C of G depicts the magnitude and direction of the neuromuscular system's regulation of postural sway, thus both must be measured to assess the net control of balance by the CNS.

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SEPARATION OF THE IMPACT AND ACTIVE COMPONENTS OF DROP JUMPS USING FOURIER ANALYSIS. Ross H. Sanders and Barry D. Wilson, University of Otago, Dunedin, New Zealand.

INTRODUCTION: Nigg (1986) has described the force-time profiles of activities such as running and jumping as comprising an 'impact' component and an 'active' component. The active component is actively produced by the subject and consists of low frequencies. The impact component results from the differential acceleration of the rigid parts and non-rigid parts (wobbling mass) and is mainly comprised of high frequencies. This paper presents a method of separating these components using a Fourier transform and inverse transform.

METHOD: Two male and two female volunteer subjects (S1 to S4) performed 10 drop jumps (from a height of 40cm) for maximum height from a Kistler 9281B force platform. Force data was sampled at 500Hz and stored in a Magnum PC-2 computer. The change in force from the resting level and the return to this level was identified from the force record. The data between these two points represented the period of contact of the subject with the force platform. To permit comparisons between subjects the force data was normalised by dividing by subject mass.

Mathematical Representation of the Force Data. A half range expansion was performed on the data corresponding to the period of contact. This technique reflects the data, thereby doubling the period of the force function (see for example Alexander and Jayes, 1980). The data can then be simply represented as a finite Fourier series of sine terms.

Time Normalisation of Force Data. Each expanded set of force data was then normalised to 201 samples by employing the Fourier coefficients to regenerate the data with the required number of data points. The first 101 data points then represented percentiles of the period of contact.

Separation of Active and Impact Components. The normalised force for the period of contact was reconstructed by performing an inverse transform from the force record retaining frequencies up to the Nyquist limit. This was then termed the 'actual force'. The contact period was divided into 10 equal parts (deciles). RMS error of the reconstructed force record for each harmonic cutoff from the actual force record was calculated for each decile of the contact period to provide a measure of the difference between the functions at different times of the jump. Since the high frequency components of the force record associated with the impact were expected to decay, the harmonic cutoff which produced a close fit (small RMS error) to the actual function in the latter deciles was selected as the most appropriate to separate the impact and active components. Residuals from the selected function and the actual force were then computed to represent the impact component.

RESULTS: The harmonic cutoffs selected to define the active component ranged from 5 harmonics (6.22Hz) to 11 harmonics (highest frequency =13.48 Hz). However, there was little variability in the selected harmonic or the frequency range within subjects. Mean RMS errors for the deciles on which the cutoff selections were based are shown for S3 in figure 1. RMS errors of less than 0.4m.s^{-2} represent an error of less than approximately 1% in the force estimates and thereby can be considered a good fit to the data at these deciles. For this subject the 7th harmonic was commonly selected as the cutoff frequency. Figure 2 shows a typical force-time plot of S4 and the active component determined by the method described. It is apparent that as the wobbling mass settles there is a close match between the two functions. In figure 3 residuals are shown for different harmonic cutoffs of a typical drop jump of S3.

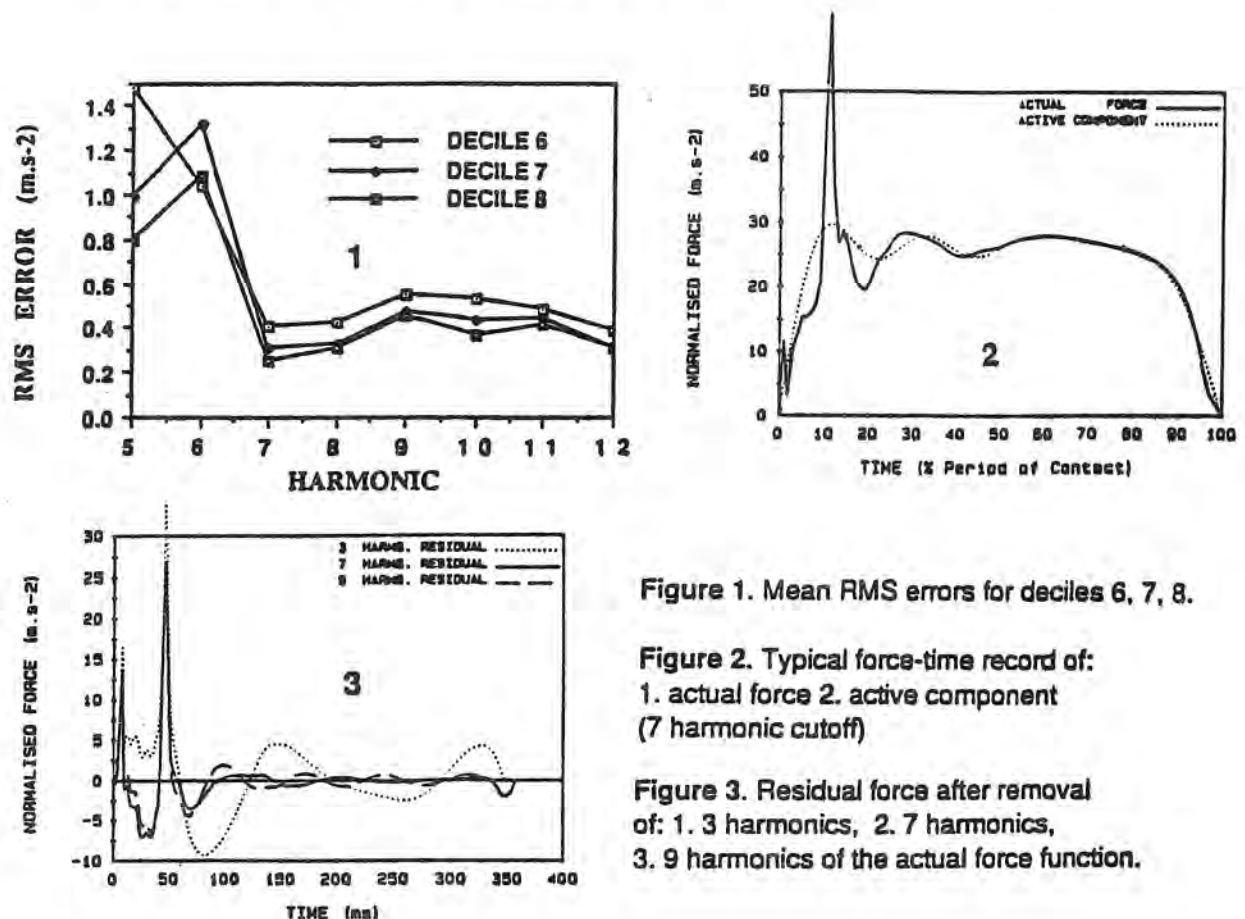


Figure 1. Mean RMS errors for deciles 6, 7, 8.

Figure 2. Typical force-time record of:
1. actual force 2. active component
(7 harmonic cutoff)

Figure 3. Residual force after removal
of: 1. 3 harmonics, 2. 7 harmonics,
3. 9 harmonics of the actual force function.

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THE RELATIONSHIP BETWEEN TECHNIQUE AND GROUND REACTION FORCES AT LANDING
IN NETBALL

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Netball is a fast and skilful game in which players frequently perform short bursts of acceleration to "break free" from a defender, followed by rapid deceleration which, by the rules of the game, is limited to one-and-a-half steps while in possession of the ball. Previous studies (Steele and Milburn, 1987; 1988) indicated that this period of rapid deceleration can result in excessively high braking forces at foot-ground contact with values as high as 6.5 times a player's body weight (BW) recorded. Such forces may subject the lower extremity ligaments, especially at the knee, to excessive stress. Peak vertical ground reaction forces (VGRF) at landing in netball under various footwear conditions (mean = 3.9 to 4.3 BW) were also shown to be greater than the values of 2 to 3 BW demonstrated in activities such as running (Cavanagh and Laforune, 1980). Although players may be able to withstand high ground reaction forces if their musculoskeletal system is properly aligned, an individual with a skeletal malalignment or unusual footfall pattern may risk potential injury. The purpose of the present study was to evaluate the relationship between technique and ground reaction forces generated at landing in netball to identify a technique which would minimize stress on the musculoskeletal system.

Ground reaction force data were collected for 10 skilled players (mean age = 19.6 years) landing on a KISTLER force platform after receiving a series of high and standard chest level passes while defended. The landing action of each subject was filmed from a posterior and lateral view at 200 Hz using two 16 mm high speed phase-locked cameras located 12 metres from a calibrated zone. A calibration cube (0.85 m x 0.85 m x 0.85 m) with 44 known 1 cm diameter spheres was filmed prior to each initial trial after loading or disturbing the cameras. This enabled later calculation of the Direct Linear Transformation (DLT) camera constants necessary to compute the three-dimensional coordinates of the 23 targets located on each subject. Slide pictures were taken of each subject immediately following attachment of the targets and prior to filming to enable later calculation of the three-dimensional location of the segmental targets in relation to the respective anatomical frame of references (F.O.R.).

Six representative trials per subject were digitized and smoothed using a three-point moving chord average. The spatial location of the target markers on each subject were determined using the DLT method while the orientation and motion of the trunk and the segments of the landing limb were calculated for the frames representing initial foot-ground contact and peak resultant force using a series of transformation matrices. The kinematic variables quantified included the relative angle of flexion-extension, abduction-adduction, and internal-external rotation at the ankle, knee and hip of the landing limb; the angular velocity at each joint in the same three orthogonal planes of motion; and the horizontal displacement of the foot from the hip of the landing limb.

Force-time characteristics of the peak VGRF, peak resultant mediolateral and anteroposterior forces (braking forces), and the peak resultant force were determined for each trial. A multiple correlation matrix was applied to the data to identify any significant ($p < 0.05$) relationships among the kinematic and kinetic variables calculated at landing.

Numerous significant relationships were demonstrated between the technique used by a player and the ground reaction forces generated at landing. A lower peak VGRF was most strongly related to decreased ankle adduction at foot-ground contact and increased hip flexion throughout the landing action. Increased time to the peak VGRF was predominantly related to increased knee flexion at peak resultant force. Although significantly related to lower limb and trunk alignment, lower peak braking forces were most strongly correlated with decreased foot-hip displacement. An increased foot-hip displacement was associated with an exaggerated "striding out" position characteristically adopted at landing by netball players. Lower peak resultant forces were related to a reduction in the degree of ankle adduction at foot-ground contact. Conversely, increased time to the peak resultant force was related to increased knee and hip flexion in combination with a decrease in knee flexion-extension velocity at peak resultant force. However, decreased knee flexion-extension velocity was significantly related to higher peak braking forces.

Results of the present study indicated that the technique used by a player after receiving a pass was related to the forces generated at landing. It is therefore concluded that players should adopt the following technique modifications: land with the foot neutrally aligned to eliminate excessive ankle adduction-abduction, internal rotation or dorsiflexion (characteristic of an exaggerated heel-strike footfall pattern); ensure adequate hip flexion (approximately 33 degrees at foot-ground contact and 45 degrees at peak resultant force) and adequate knee flexion (approximately 17 degrees at foot-ground contact and 40 degrees at peak resultant force); and reduce the foot-hip displacement by eliminating an exaggerated "striding out" position. Such technique modifications could decrease the ground reaction forces experienced at landing and, in turn, decrease the stress placed on the musculoskeletal system.

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INFLUENCE OF TIMING ON VERTICAL JUMPING PERFORMANCE: A SIMULATION STUDY.

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Introduction

Recently, Bobbert et al. (1988) performed a study of coordination in vertical jumping. One of their findings was, that muscle groups of the lower extremity were not activated simultaneously, but in a proximo-distal time sequence. This pattern of timing was explained using mechanical considerations dealing with transformation of rotational segmental movement into translational movement of the center of gravity. It was shown that the bi-articular rectus femoris and gastrocnemius muscles play an important role in this transformation. Using inverse dynamics it is not possible to establish the exact influence of timing, nor to find its true optimum with respect to a mechanical movement goal. To achieve such aims, direct dynamics is more promising. In this paper results of preliminary simulation studies using a highly simplified model are presented, showing a relationship between timing and jumping performance.

Model description

The model consists of three rigid segments: foot, lower leg and upper leg, all having uniform mass distribution. Mass of the upper body is concentrated at the hip. The hip is allowed to move in vertical direction only, thus eliminating balance problems. A knee extensor is modelled using a linear torque-angle characteristic. The bi-articular gastrocnemius muscle is modelled as purely tendinous, behaving as a stiff linear spring when above rest length.

In this study, rest length of the tendinous structure (henceforth called 'tendon') and thus time at which the tendon became taut (henceforth called 'tendon timing') was used as the independent variable. For all lengths, tendon became taut during the final part of the push-off only.

All simulations started from a static position at which knee angle was 50 degrees.

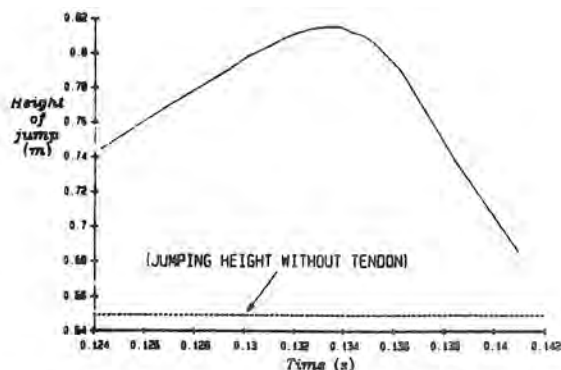


Fig. 1. Jumping height as a function of time at which tendon becomes taut.

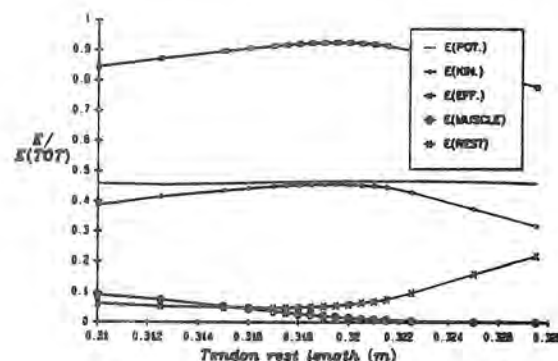


Fig. 2. Energy terms at take-off as a function of tendon length.

Simulation method

Behaviour of the model was simulated using SPACAR, a finite element method with deformable links, developed at the Technical University of Delft (van der Werff, 1977). Integration of the differential equations was performed using a variable stepsize variable order predictor-corrector method (Shampine & Gordon, 1975).

Results

Total energy in the system is constant and can be separated into five terms: $E(KIN.-Y)$, kinetic energy of the center of gravity in vertical direction, $E(POT.)$, potential energy of the center of gravity, $E(EFF.)$, the sum of the two terms mentioned, $E(MUSCLE)$, energy in the knee extensor and $E(REST)$, a rest term comprising rotational energy of the segments, kinetic energy in horizontal direction and energy in the tendon. $E(EFF.)$ at take-off of course determines jumping height.

As can be seen from fig. 1 jumping height varied appreciably with tendon timing, having an optimum when tendon became taut at $t=0.133$ s., corresponding to a tendon rest length of 0.3195 m. At optimum timing, 93% of total energy contributed effectively to jumping height reached (see fig. 2).

In fig. 2 total energy at take-off is separated into the terms described above. A clear optimum exists for $E(KIN.-Y)$, whereas $E(POT.)$ is relatively insensitive to changes in timing. $E(MUSCLE)$ decreases with greater tendon length, whereas $E(REST)$ increases sharply at tendon lengths above 0.32 m.

Discussion

Jumping height at optimum tendon timing was 1.26 times jumping height for a model without tendon. Thus, a passive structure adding no energy to the system increased efficiency of the movement from 0.74 to 0.93.

This optimum can be explained as follows:

- At tendon timing below optimum, tendon starts decelerating the rotation of the upper leg relatively early. Therefore, the knee is not extended far enough, leaving too much energy in the knee extensor at take-off.
- At tendon timing above optimum, tendon becomes taut relatively late. Therefore, angular velocity of upper and lower leg remains too high, creating a significant increase in $E(REST)$ at take-off.

Of course, the explanation for the existence of an optimum in tendon timing offered here cannot be directly related to timing problems for human subjects performing maximal vertical jumps. Therefore, at present a model is being developed in which dimensions are more realistic and in which knee extensor and tendon are replaced by Hill-type muscle models for which force is a nonlinear function of length, velocity and activation. Nevertheless, results presented in this paper lend further support to our view that timing should be understood in the light of the mechanical context and the dynamics of the entire system.

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DETERMINATION OF THE COMPOSITE AND INDIVIDUAL JOINT SINUS CONES OF THE HUMAN SHOULDER COMPLEX

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Research and studies on the human shoulder complex have more than a century long history, and a brief summary of the subject matter is presented by the senior author [1]. Modelling of the human shoulder is essential for the multi-segmented mathematical models as well as design of the shoulder mechanism of anthropometric dummies. The most striking feature of the human shoulder is the presence of independent articulations among the bones of the shoulder complex, i.e., the clavicle, scapula, and humerus. It is possible to model the human shoulder by utilizing the concept of the joint sinus cones. In general, the joint sinus cone is the conical surface enveloping the joint sinus which is defined as the total range of angular motion permitted by a moving segment of a joint when the other segment is rigidly fixed. In the case of the human shoulder one can define individual joint sinus cones for various articulations of the human shoulder complex as well as a composite joint sinus cone describing the total range of the angular motion permitted by the upper arm (humerus) with respect to the torso.

In this paper first the parameters of an elliptical composite joint sinus cone, and second the individual joint sinus cones associated with the sternoclavicular, claviscapular, and glenohumeral joints of the human shoulder complex are quantitatively determined by introducing the "minimum joint motions" criterion and utilizing in-vivo data base established by Engin and Chen [2].

To determine the composite joint sinus cone the data of reference [2] is first converted from a format of single-valued functional relationship, expressed in a local joint axis system, into a set of 32 unit vectors describing the circumductory motion of the upper arm in a torso-fixed coordinate system. This set of unit vectors is then employed in determining the five parameters of the composite joint sinus cone, namely, three Euler angles which locate the cone in a torso-fixed coordinate system and the two half apex angles of the elliptical cone. Thus, the problem now becomes one of finding the five parameters. The redundant data permits an optimization scheme to be used to find the best-fitted elliptical cone surface in the least squares sense. This is a highly nonlinear optimization problem and requires numerical solution. To check both accuracy and stability two different methods, namely, the flexible tolerance method and the direct method are considered and their results are compared.

Since there are three elliptical joint sinus cones representing sternoclavicular, claviscapular, and glenohumeral articulations the number of cone parameters to be determined becomes 15 in the case of individual

joint sinus cones. In view of the fact that from the data of reference [2] one can only extract the unit vectors defining the orientation of the humerus with respect to the torso during circumductory motion of the upper arm, the system is redundant and unique solution is possible by introducing an appropriate optimization criterion, which is taken as the minimization of the rotations of the clavicular, scapular, and humeral links as well as the axial rotation of the clavicular and scapular links from their respective reference positions. The natural choice for these reference positions is the one corresponding to the relaxed position of the arm. In this way, the problem is one of optimization with equality constraints corresponding to the required humerus orientations.

Two different optimization techniques, namely, the Lagrange multipliers method and the flexible tolerance method are considered. The former is a direct method which makes use of Lagrange multipliers to convert a constrained optimization problem into an unconstrained one. By setting the partial derivatives of the objective function of the unconstrained problem to zero, a set of highly nonlinear strongly coupled algebraic equations are obtained, and solved through the use of IMSL library subroutine. The flexible tolerance method, being a direct search method, can accommodate any type of constraints without the need for partial derivatives. The accuracy and stability of the solutions are checked by utilizing both methods.

In view of the foregoing the following results were obtained. The apex angles for the composite sinus cone are 147° and 168° . The apex angles of the sternoclavicular cone (Cone I) were found to be 28° and 46° . The corresponding angles quoted by Dempster [3] based on cadaveric measurements are 35° and 44° . The other apex angles in the present investigation are 30° and 39° for the claviscapular cone (Cone II) and 95° and 135° for the glenohumeral cone (Cone III). Dempster [4] gives 30° and 60° for Cone II and 100° and 160° for Cone III. The aforementioned figures show a reasonably good agreement between the theoretical results obtained in this study and those measurements made on cadaveric specimen preparations. One should also note that the theoretical values are based on in-vivo data base, whereas the ones obtained on cadaveric preparations did not include muscles, thus giving rise to larger cone sizes.

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THREE-DIMENSIONAL ASSESSMENT OF HUMAN GAIT

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The main goal of this study is to propose a coefficient describing motion of the ankle, knee and hip, during walking in 3-D space. The muscular moment about the axis of the ankle, knee and hip joint permits the computation of the following coefficient:

$$P_{3-D} = \sum_{i=1}^3 \sqrt{\frac{1}{T} \int_0^T (M_{x_i} \cdot \omega_{x_i} + M_{y_i} \cdot \omega_{y_i} + M_{z_i} \cdot \omega_{z_i})^2 dt} \quad /1/$$

where: T - is the duration of the support phase

$M_{x_i}, M_{y_i}, M_{z_i}$ - is the muscular moment about the i-th joint projected onto the OX, OY, OZ axis, respectively.

$\omega_{x_i}, \omega_{y_i}, \omega_{z_i}$ - is the angular velocity about the i-th joint projected onto the OX, OY, OZ axis, respectively.

In order to show an application of the formula /1/, the results, obtained during the investigation of a few normal walking subjects /of both sexes/, have been compared. The compared subjects were walking with approximately the same velocity. The value of the coefficient P_{3-D} demonstrates energy saving during the support phase. The obtained coefficient, for normal walking subjects, permitted the detection /more precisely than it has been done before/ of imperfections of leg's motion in the support phase. The transport phase of leg's motion was investigated using non-conventional phase diagrams of lower and higher order: $\dot{q}/q$, $\ddot{q}/\dot{q}$, and $\dddot{q}/q$, where: q - is the relative angle between the chosen segments of the leg. Hip, knee and ankle joint phase diagrams in three planes OXY, OXZ, OYZ

of the orthocartesian coordinate frame OXYZ, were obtained for the investigated subjects. The phase diagrams are superior in comparison with conventional diagrams q/t , $\dot{q}/t$, $\ddot{q}/t$, because they show perturbations in the smoothness of motion, especially in the transport phase of motion. It has been found that the phase diagrams are very useful in investigation of the support phase, as it is much easier to detect this phase on the phase diagram than on the q/t diagram.

The proposed power coefficient /1/, as well as the lower and higher order phase diagrams describe bipedal locomotion to an extent that has not been known up till now.

The motion was registered on an automatic movement analyser named ELITE /installed at Centro di Bioingegneria in Milan, Italy/.

The phase diagrams were obtained on an IBM-PC/AT microcomputer using a special set of programs named Human Locomotion /HL/.

INTRA-ABDOMINAL PRESSURE AND PATTERNS OF ABDOMINAL MUSCLE ACTIVATION IN ISOMETRIC TRUNK FLEXION AND EXTENSION

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INTRODUCTION

In many motor tasks there is an increase of pressure within the abdominal cavity, however, the functional role of the elevated intra-abdominal pressure (IAP) and the patterns of muscle activation accompanying it are not well understood. The complicated coordination of the anterior abdominal wall, glottis and diaphragm makes isolating individual contributions difficult. In trunk extension, an increased IAP could provide an assisting extensor torque (cf. Thomson, 1988). In this situation it would seem beneficial to raise the IAP by means other than involvement of the rectus abdominis and oblique muscles, as they could also create a counteracting flexor torque (c. McGill and Norman, 1987). In trunk flexion, on the other hand, a strong abdominal muscle contraction, as necessary to produce flexor torque, would also result in an elevated IAP, which in turn could be counter-productive to the task. It therefore seems reasonable that other mechanisms, such as relaxation of the glottis and diaphragm, would take place to minimize the IAP. This investigation was undertaken to explore these possibilities by measuring abdominal and back muscle activity, IAP, and produced torque during standardized trunk flexion and extension tasks.

METHODS

Intra-abdominal pressure (IAP) was measured with a tip pressure transducer inserted into the gastric ventricle via the nose (Grillner et al., 1978). The position of the transducer was confirmed by monitoring pressure changes during sub-maximal Valsalva maneuvers.

Myoelectric activity (EMG) from four muscle sites were recorded using bi-polar surface electrodes placed over: i) rectus abdominis (RA), 3.0 cm lateral to the umbilicus; ii) obliquus externus (OE) between the anterior superior iliac spine and the caudal border of the rib cage; iii) obliquus internus (OI) between the anterior superior iliac spine and the symphysis pubis and iv) erector spinae (ES) 3.0 cm lateral from the mid-line at the L3 level. EMG signals were pre-amplified, filtered through a bandwidth of 5Hz - 10kHz, rectified and integrated.

Torque measures for trunk flexion and extension were obtained in a side-lying position using a swivel table attached to a dynamometer (Thorstensson and Nilsson, 1982). The axes of rotation of dynamometer and swivel table were aligned to the L3 level of each subject. Torque, IAP and EMG were obtained during maximal voluntary contractions in seven equally incremented trunk positions ranging from +60° (flexion) to -30° (extension) with 0° being a straight body. Type of contraction and position were randomized. In each situation two consecutive trials were performed and the one with the highest mean torque for a one second period was taken for analysis. For this same period, mean IAP and integrated EMG values were calculated. Values presented are means for four male subjects. The ranges for age, height and weight being 23 - 27 (yrs) 183 - 190 (cm) and 77 - 86 (kg) respectively.

RESULTS

Extension tasks (Fig. 1A). IAP levels were markedly elevated, typical values were in the range 7.2 - 10.6 kPa (54 - 80 mmHg) and tended to rise with an increased flexion angle. EMG activity of the abdominals were generally less than 10% of their recorded maximum with RA being the least active in all positions. Erector spinae activity was at 70% or more of its recorded maximum over the range of trunk positions. The highest torque was produced at +60° (288 Nm) and declined successively with increased trunk extension angles.

Flexion tasks (Fig. 1B). IAP ranged from 7.3 kPa (at +60°) to 15.5 kPa (at -15°). The EMG activity of RA, OE, and OI were at all times above 60% of their recorded maximum. Erector spinae showed low activity, averaging below 10% of its recorded maximum. Trunk flexor torque was typically lower than that recorded in trunk extension (about 40% of the recorded maximum) with little change due to position.

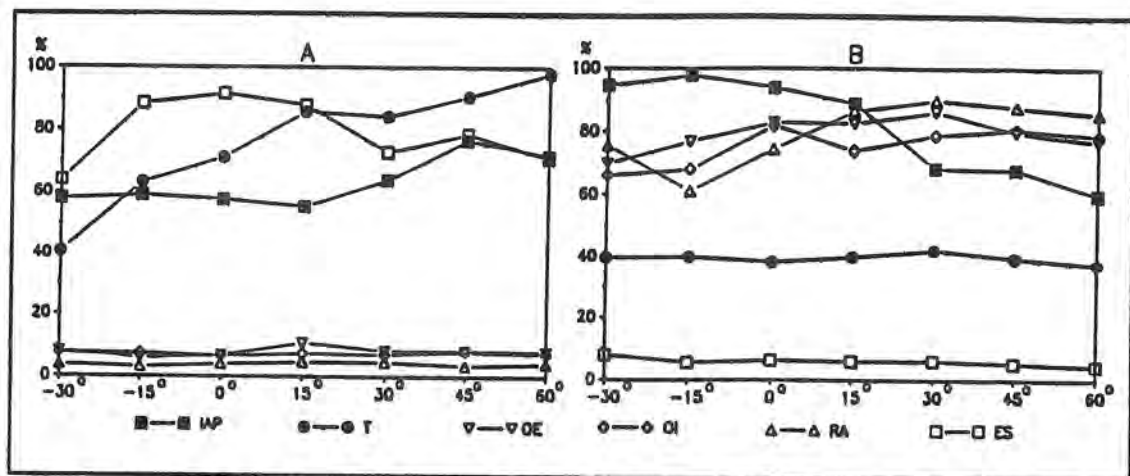


Figure 1: Mean values ($n=4$) for maximal isometric contractions (A - extension; B - flexion) at different trunk positions. All values are expressed as a percentage of their recorded maximum.

DISCUSSION

The main findings were that a marked elevation of the intra-abdominal pressure (IAP) was present both in maximal isometric trunk extension and flexion and that a pressure increase could be produced with or without abdominal muscle activity, depending on the task.

The sizable levels of pressure recorded in trunk flexion were likely a result of the reduction in abdominal volume through increased abdominal muscle activity. The possible extension moment produced by this pressure increase would appear to be a disadvantage to the movement as a whole. Evidently, a compensation from the diaphragm or glottis to reduce this pressure did not seem to occur. The advantage, if any, that this increased IAP may provide in active flexion is unclear. However, its function may be to provide support for the spine and/or abdominal musculature.

In trunk extension IAP was elevated with little or no activity from the obliquus externus, obliquus internus and rectus abdominis muscles. The low levels of abdominal activity were of particular interest indicating that the increase of pressure within the abdomen was elevated and sustained by the diaphragm and/or the transverse abdominis muscle. The involvement of these muscles over the obliques and rectus abdominis is important, as the assisting IAP extensor moment could be produced without the development of a counter moment. This mechanism has implications in lifting tasks where an increased IAP may assist in trunk extension and help in the reduction of spinal load.

Acknowledgements. This study was supported by a grant from the Swedish Work Environment Fund.

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Reaction time and postural responses during arm movements in the frontal and sagittal planes

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INTRODUCTION

Prior to a voluntary movement there is a feed-forward control of the equilibrium (cf. Oddsson 1988). When performing arm movements in the sagittal plane postural muscles of the legs and trunk are pre-activated in a distal to proximal order to counteract the perturbing effect on the equilibrium caused by the voluntary movement (e. g. Belenkii et al. 1967, Bouisset & Zattara 1981). Earlier studies have primarily dealt with postural responses to different arm movements in the sagittal plane which mainly will disturb joints with an axis of rotation in the frontal plane. Thus, movements around the ankle, knee and hip joints are involved in the postural adjustment to these perturbations (Friedli et al. 1984). This study was undertaken to investigate postural responses and reaction times for arm movements in the frontal plane and compare them with reaction times of sagittal arm movements. It was postulated that the reaction time of frontal as compared to sagittal arm movements would be shorter due to higher postural stability (cf. Cordo & Nashner 1982)

METHODS AND PROCEDURES

Six physically active male subjects participated in the study. In response to the onset of a visual stimulus (white light) each subject was instructed to perform (in a randomized order and one at a time) four different forms of arm movements, 1) left arm to the side, 2) right arm to the side, 3)

both arms to the side and 4) both arms forwards. The subjects performed 10-12 movements of each type and the 8 best trials were selected and included in an analysis of variance (ANOVA). Reaction time was measured as time from onset of the visual stimulus to start of muscle activity in the prime mover muscle.

In three of the subjects LEDs (light emitting diodes) were attached to the shoulder and the hand. The position of each LED was detected by a SELSPOT system and software was used to calculate the angle of both arms with respect to the vertical.

Muscle activity (EMG) was recorded from the prime movers of the task (left and right deltoid muscles, LD & RD) in all subjects and from muscles controlling movements of the left side of the body including ankle, knee, hip and trunk muscles in four of the subjects (peroneus longus (PL), tensor fascia latae (TFL), obliquus externus (OE), rectus abdominis (RA) and erector spinae (ES)). Activity in the soleus (S) tibialis anterior (TA) and hamstring (HAM) muscles was recorded in one of the subjects. The EMG signals were preamplified 100 times, high-pass filtered and further amplified 50 times before being low-pass filtered and rectified.

RESULTS AND DISCUSSION

Frontal arm movements. Postural responses of muscles controlling movements in the frontal plane are shown in Fig. 1. During arm raising

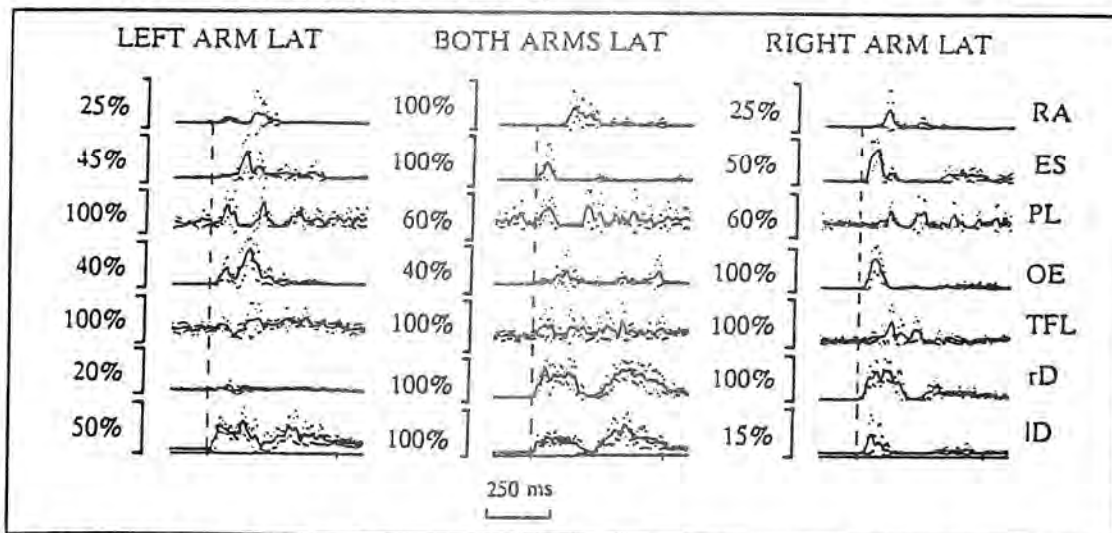


Fig. 1. Mean and standard deviation of 8 superimposed EMG recordings for one subject performing different lateral arm movements. Dotted line indicates onset of prime mover activity. Amplitudes shown are relative to peak amplitude (100%) in each muscle during the different arm movements. For abbreviations see text.

movements to one side the OE muscle of the opposite side was activated together with the RA or ES muscles of the opposite side. These muscles were activated in a well defined burst simultaneously or slightly after the prime mover muscle (from 10 ms before to 40 ms after prime mover onset). The individual difference in the activation of ES or RA was probably due to small variations in the initial posture and/or that the movement was not performed purely in the frontal plane. Trunk muscles on the same side of the body as the prime mover were activated somewhat later and to a lesser degree than the corresponding trunk muscles of the opposite side (Fig. 1). The PL muscle on the same side as the prime mover showed an increase in activity shortly after the onset of the prime mover (cf Fig. 1). This increase was, however, very inconsistent and not present in all trials. The response in PL varied considerably between subjects and trials. This variation may also be due to small individual variations in the posture prior to the execution of

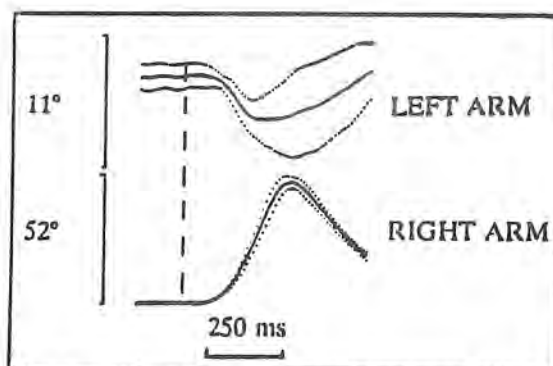


Fig. 2. Mean and standard deviation of 8 superimposed recordings of angular displacements of the arms when performing a right lateral arm raise. Dotted line indicates onset of prime mover (right deltoid).

a movement. Activity in TFL was generally very low and in many trials absent. During arm raising to one-side the HAM muscles on the opposite side were activated 10-40 ms before the prime movers. The HAM muscles were also pre-activated during a bilateral arm movement.

During lateral movements of one arm all subjects in most trials showed an activation of the opposite deltoid muscle. This activation appeared either simultaneously with or up to 100 ms after the onset of the prime mover (Fig. 1). This activation resulted in a marked displacement of the arm (Fig. 2). This was probably an associated postural adjustment since it could counteract the perturbing effect on the equilibrium caused by the primary movement.

Sagittal arm movements. Earlier results showing a preactivation of postural muscles of the legs and trunk were confirmed in this study. Activity

in ankle extensor muscles was low and often absent, whereas the hamstring muscles were consistently activated with a marked burst 20-80 ms before the activation of the deltoids (during contralateral and bilateral arm raising). The erector spinae muscles (ES) were activated almost simultaneously with the prime movers.

Reaction times. Mean values of the different

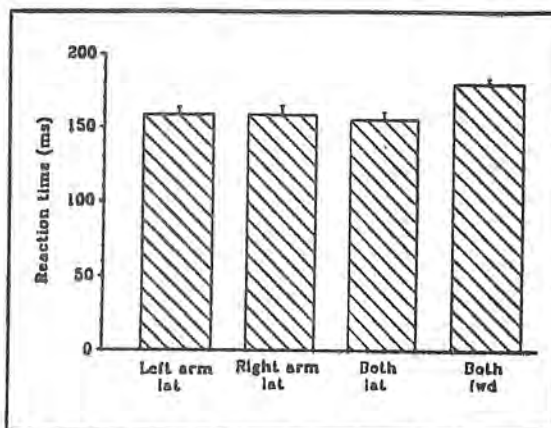


Fig. 3. Mean and 99% confidence limit for the reaction times of four different forms of arm movements.

reaction times are shown in Fig. 3. The reaction time during arm raising movements forwards was significantly longer than all other values ($p < .01$).

It is concluded that this difference may be due to a better mechanical stability of the body in the frontal as compared to the sagittal plane (cf. Cordo & Nashner 1982). The larger base of support and the fact that movements in the frontal plane are restricted in the ankle, knee and hip joints support this conclusion.

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ABDOMINAL AND HIP FLEXOR MUSCLE INVOLVEMENT IN VARIOUS TRAINING EXERCISES

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INTRODUCTION

Different forms of training exercises for the flexor muscles of the trunk and hip have an interest in many sport contexts and rehabilitation programs, e.g. for low back pain. It is often desirable to design these exercises to specifically engage only one or a few of the muscles controlling the movements of the hip, pelvis and trunk. A possible beneficial effect of abdominal muscle training on the lumbar lordosis could, for example, be counteracted by undue involvement of hip flexors, such as the iliopsoas muscle. Information on the complex coordination between the muscles of the trunk and hip flexor synergies is scarce. This is particularly true for the iliopsoas muscle due to its relative inaccessibility for electromyographic recordings (cf. Linden and Delhez, 1986).

The purpose of this study was to describe the patterns and relative magnitudes of activation for three abdominal and three hip flexor muscles, including the iliopsoas (iliacus), during a variety of common training exercises.

METHODS

Three healthy male subjects (25-31 years, 67-87 kg, 178-195 cm) participated in the study. They performed repeated sit-up exercises (hands on the chest) at the rhythm of a metronome. Each movement cycle (up and down) lasted 4 seconds. Three different types of movements were used: (i) sit-up without specific instructions, (ii) isolated trunk bendings, and (iii) isolated hip flexions. In each type of movement four different combinations of supported or nonsupported feet, straight or flexed legs were carried out.

Electromyographic (EMG) recordings were made with surface electrodes from the hip flexor muscles: sartorius (SA) and rectus femoris (RF) and from the trunk flexors: rectus abdominis (RA), obliquus externus (OE) and obliquus internus (OI). The EMG of the hip flexor iliacus (IL) was picked up with indwelling bipolar fine-wire electrodes (teflon insulated stainless steel, diameter 75 μ m). The wires were inserted with a 0.7 mm needle at a point 1 cm medial to the sartorius muscle, 1 cm below the inguinal ligament and 3 cm lateral to the femoral artery. The localization of the wires was controlled in one subject using ultrasound technique. All EMG signals were sampled at 2 kHz, bandpass filtered at 10-1000 Hz and analyzed via an on-line computer system (Moritani et al., 1986). The angular movements of the hip and upper trunk were recorded with two electrogoniometers.

For each exercise, integrated and averaged EMG were measured for the whole movement cycle (starting 200 ms prior to the start of the movement). The mean amplitude value was taken and expressed as a percentage of the highest mean amplitude value for each muscle found in any of the different exercises (cf. Fig. 1).

RESULTS AND DISCUSSION

When the subjects were asked to perform the sit-up exercise without specific instructions ("spontaneous sit-up", 1 in Fig. 1) they started to flex the trunk first and then the hip. The level of activity in both trunk and hip flexors was intermediate, between 35% (IL) and about 70-75% (RF, OI) of the recorded maximum (Fig. 1).

In the more specific exercises studied, the iliacus and the other hip flexor muscles showed low or no activity when the sit-ups were performed as pure trunk flexions (e.g. 2 in Fig. 1). There was no movement at the hip joint nor did there appear to be a need to stabilize the pelvis in these situations. Changes in leg position or whether the feet were supported or not, did not change the pattern. During hip flexion, the hip flexor activity was high and increased markedly with anchored legs (4 and 6 versus 3 and 5 in Fig. 1). The activity of both IL and SA reached its highest value in the bent leg, feet supported situation. This can probably be attributed to the decrease in force potential due to the overall decrease in muscle length.

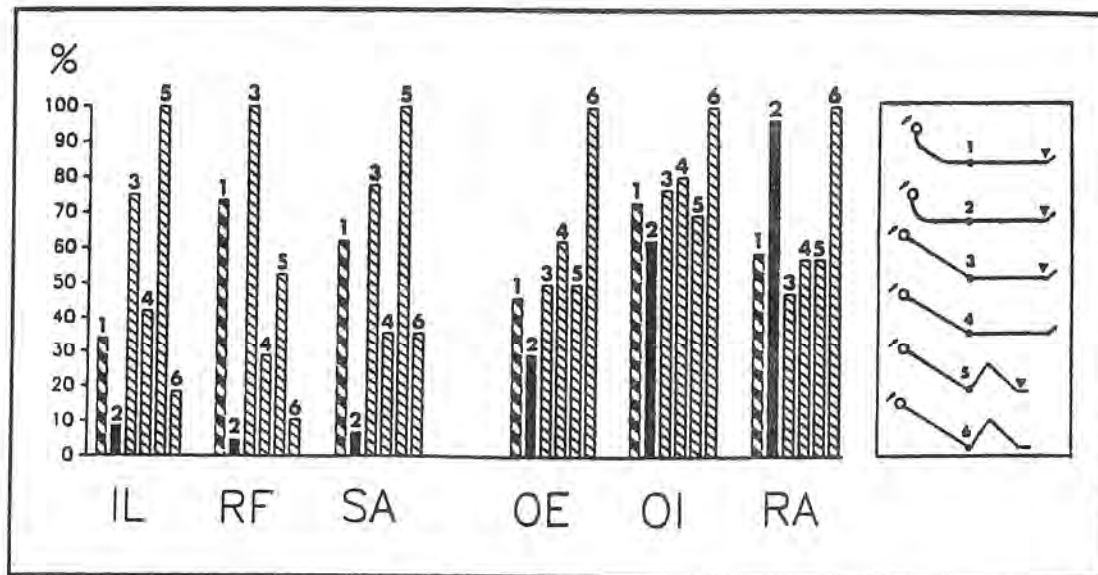


Figure 1. Mean EMG amplitude values for hip (IL, RF, SA) and trunk flexor muscles (OE, OI, RA) during six different forms of sit-up exercises (1-6, illustrated to the right). Average values for three subjects are given, expressed in percent of the recorded maximum. The exercises were (1) spontaneous sit-up, (2) trunk flexion, and (3-6) hip flexions. Filled triangles indicate support of the feet. For further details and explanation of the abbreviations see Methods.

The abdominal muscles were selectively activated in trunk flexion exercises where they act as prime movers. Particularly RA showed high activity in these situations. Interestingly, the highest activity in all the three abdominal muscles was seen when the instruction "hip flexion with bent legs and no support" was given (6 in Fig. 1). It should be noted that this exercise actually resulted only in a trunk flexion since the hip flexion could not be executed due to the relatively low moment of inertia of the legs. In all hip flexion exercises there was an activation of the abdominal muscles. Mean values averaged 45-100% of the recorded maximum (3-6 in Fig. 1). The function of the abdominal muscles was to keep the trunk stabilized (isometric contraction) during the raising of the upper body via hip flexion (with the exception of exercise no. 6, cf. above).

Thus, it is possible to modify the design of a sit-up exercise to selectively activate the trunk and not the hip flexor muscles, whereas the reversal appears difficult due to the need for trunk stabilization. Modifications, such as bent legs or support to the feet, will effect the relative activation of primarily the hip flexors by changing the moment counteracting the raising of the upper body.

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TORSIONAL STRENGTH AND MUSCLE ACTIVITY DURING AXIAL TWISTING OF THE TRUNK

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INTRODUCTION

Although twisting has been found to be associated with low back pain and injury (Frymoyer et al 1980), little is known about the mechanics of the lumbar spine during the development of axial twists. Several years ago, we constructed a dynamic model of the trunk to predict loads on the intervertebral joint, and the 50 muscles and ligaments that cross L4/L5 during sagittal plane lifting of loads. (see McGill and Norman, 1986). The model was driven from biological responses measured from the subject (ribcage and pelvis position to orientate the lumbar vertebrae and EMG to estimate muscle activity). We have recently begun to extend this model to predict tissue loads during dynamic axial twisting efforts. However, several issues pertaining to the modelling of twisting remain unresolved, two of which constitute the purpose of this study: Firstly, little is known about the maximum torsional strengths of the trunk and the possible modulating effects of rotation position, direction, and velocity of rotation. Secondly, we must understand the relationship of abdominal and erector EMG to the development of torsion, or perhaps another, as yet undefined, mechanical function.

METHOD

Ten male and five female university students participated in this study.

Axial torques were measured from a jig which was mounted to the subject's shoulders and was linked to an isokinetic dynamometer (CYBEX II) mounted overhead. The pelvis was rigidly fixated while subjects performed isometric twisting efforts in a ramp increment to maximum effort within 4 seconds. Both clockwise and counter clockwise isometric efforts were recorded with the trunk at 0° and pre-rotated at +30° and -30°. Maximum dynamic efforts at both 30°/s and 60°/s in both directions were also recorded. Six EMG sites (rectus abdominis, external and internal obliques, latissimus dorsi, T9 erector spinae, L3 erector spinae) monitored activity during all tasks. EMG channels, torque magnitude and

position were digitally recorded and processed. Several strategies to obtain maximum EMG amplitudes for normalization were performed.

RESULTS AND DISCUSSION

The maximum torsional strengths are shown in table 1. Females were able to generate approximately 45% of the torque that the males generated. Velocity during the dynamic twists did not appear to significantly modulate torque output. The magnitude of isometric torques were similar to those reported by Pope et al (1987) (range: 33-167 N.m; 84 N.m average at 0°).

Investigation of the torque dependance on EMG suggests that there is some relationship between activity of the obliques and latissimus dorsi and torque during isometric exertions although there was great variation between subjects. Our subjects did not consistently produce the curvilinear relationships that have been reported by others. The activity of other muscles suggested very little relationship with torque production as their amplitudes tended to saturate at a level below 50% MVC during isometric efforts. Nonetheless, this evidence of extensive co-contraction suggests that the often used modelling assumption of negligible co-activity is invalid for twisting.

Since EMG activity during twisting in most cases did not approach the maximum amplitudes observed during the normalization trials one may suspect that an inhibitory phenomenon may be involved during twisting efforts. Schultz et al (1983) predicted 192 N.m. of torque assuming the musculature was contracting at full intensity while our subjects only generated half of this amount. Furthermore, this finding suggests that muscle activity cannot be "calibrated" to twisting exertions and that another procedure must be sought for modelling.

Dynamic trials which started at a fully pre-rotated position and finished at the opposite side show generally higher torque during the first half of the range (up to the midpoint of

rotation) while EMG activity lagged behind. Perhaps elastic recoil of passive tissues assisted torque generation. However, ligament lengths calculated during a full twist did not indicate that they were under strain suggesting that other elements such as the facets provided passive forces. Perhaps as well, in the pre-rotated position at the beginning of the trial all muscles can cause twisting due to their changing line of action, for example, rectus abdominis, which would not be parallel with the compressive axis of the spine. However, as the neutral position is approached these muscles lose their potential to generate twisting torque and the prime twisters must assume a greater responsibility. Preliminary runs of our dynamic model show that the line of action of the muscles associated with the amount of twist is far more important in modulating torque production than muscle length itself. Indeed, length excursion of any of the major extensors or abdominals during full twists did not exceed 7% of the length when the torso was not twisted.

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TABLE 1 Maximum torsional strength (N.m)

MALE								FEMALE							
								ISOMETRIC							
Position	0°		+30°		-30°			0°		+30°		-30°			
Direction	CW	CCW	CW	CCW	CW	CCW		CW	CCW	CW	CCW	CW	CCW		
Average	94.4	100.2	70.6	102.2	109.2	73.7		41.2	44.2	28	47.6	46.8	30.8		
S.D.	—	21.1	15.9	16.1	19.2	16.5	22.4	7.3	11.4	8.7	4.9	6.4	8.7		
								DYNAMIC							
Speed	30°/s		60°/s					30°/s		60°/s					
Direction	CW	CCW	CW	CCW				CW	CCW	CW	CCW				
Average	92.3	87	83.3	79.5				44.8	44	40.2	45				
S.D.	16.5	15.2	18.8	12.8				15.4	10.5	14.5	15.5				

RESPONSE OF BIOMECHANIC CORRELATES OF CUMULATIVE TRAUMA DISORDERS IN THE CARPAL TUNNEL AND EXTRINSIC FLEXOR MUSCULATURE TO SIMULATED WORKING CONDITIONS¹

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Introduction

Epidemiological studies have examined the differences in prevalence of Cumulative Trauma Disorders (CTD's) between job classifications. Silverstein et al (1987) categorized tasks based on their force and repetition characteristics using arbitrary classification guidelines designed to best separate low levels from high levels. This approach has shown that high forces and high repetition rates together have a multiplicative effect on the prevalence of CTD's. However, it is still unknown how force, repetition and posture interact to cause injury, nor at what level of interaction will prevalence of injury increase significantly. There is a need for a study that links the prevalence of injury with quantification of the stresses placed on the musculo-skeletal system by doing a task. A system was developed to predict internal factors related to loading of the soft tissues of the carpal tunnel and the extrinsic flexor musculature throughout the duration of a task (Moore et al, 1989). A preliminary study using the system was conducted with simulated grasping tasks designed to match the four categories used by Silverstein et al (1987).

Methodology

Two basic tasks were performed by four subjects. The subjects were healthy males ages 24-29. All subjects were right hand dominant and all testing took place with the right hand. The Low Repetition task (LOR) simulated glue gunning while the High Repetition task (HIR) simulated staple gunning. Both tasks were done with a caulking gun instrumented to allow direct monitoring of grasp force and modified to allow adjustment of hand force required. Both tasks were done at two different force levels (High-HIF and Low-LOF) and with two different wrist postures. Finger joint angles were collected using a lightweight instrumented glove (VPL Research, Redwood City, California). Wrist joint angle was measured using two potentiometers mounted on the wrist. EMG was monitored from the centre of the flexor musculature of the forearm. The signal was amplified, rectified, then low pass filtered with a 6 Hz upper cut off frequency.

Table 1: Combined Results for Internal Factors
Normalized to LOF.LOR

Factor	LOF.LOR	LOF.HIR	HIF.LOR	HIF.HIR
Peak Axial Force	1.00	1.14	1.98	2.34
Axial Impulse	1.00	1.08	2.51	3.27
Excursion	1.00	3.34	1.05	3.19
Ave. Exc. Velocity	1.00	3.89	1.13	3.83
Peak Exc. Velocity	1.00	1.16	1.23	1.12
Ave. Normal Press.	1.00	0.97	2.50	3.14
Peak Normal Press.	1.00	1.34	2.16	2.81
Work	1.00	3.29	2.05	8.39
Peak Power	1.00	1.35	1.64	4.34
5th Percentile APDF†	-	-	-	-
50th Percentile APDF	1.00	2.99	4.99	8.98
95th Percentile APDF	1.00	2.29	1.61	5.01

†All cases included a static component = 0

¹This work has been supported by NSERC Grant #A2785 and the Ontario Ministry of Labour.

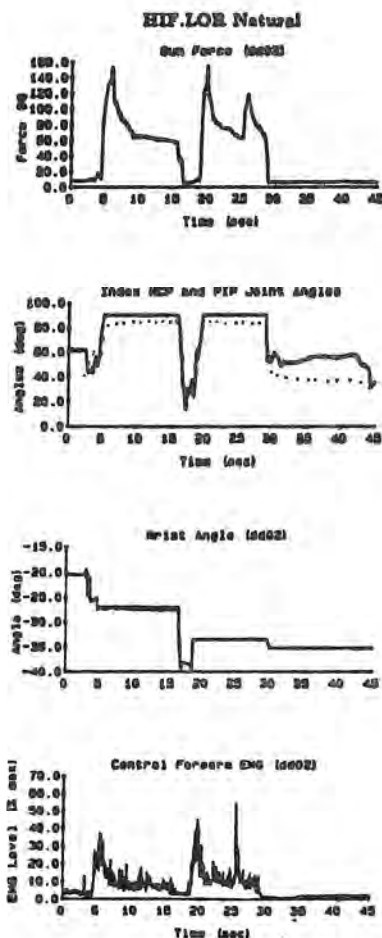


Figure 1: Measured Variables (HIF.LOR Natural Wrist Posture)

Results and Discussion

A typical plot of measured variables over one cycle can be found in Figure 1, for the LOR, "natural" wrist posture task. In the natural wrist posture the subjects were allowed to hold the gun in the most comfortable position for grasping. All subjects adopted an extended wrist posture. In the "forced" posture the subjects were asked to flex their wrists while performing the task. The subjects then adopted a neutral or slightly flexed posture. The added physiological load with the forced wrist posture was best illustrated in the APDF of the EMG (Figure 2). The forced posture required higher activation levels for longer periods of time without an increase in grasp force. For all tasks, all subjects used higher grasp force then required to perform the tasks, especially in the HIR cases. The results of the internal factors normalised to LOF.LOR, collapsed across subjects and wrist pos-

tures, can be found in Table 1. The results indicate that the work factor best reflects the odds ratios of injury found by Silverstein. This energy transfer factor includes elements of excursion (repetition), normal force (force and posture) and time. Repetition effects alone were best shown by excursion. Peak velocity of excursion did not reflect a change across conditions. Force effects were shown in both the normal and axial force measures. The dynamic component of the EMG also reflects the odds ratios of injury. The values for this component for the HIF.HIR case were below the guidelines suggested by Jonsson (1978) even though all subjects complained of fatigue at the end of three repetitions of this case for both wrist postures.

Conclusions

Preliminary results indicate that the system presented by Moore et al (1989) does quantify the effect of force, repetition and posture over the duration of a task on the musculo-skeletal system in a manner that should give useful insight into the causes of cumulative trauma disorders.

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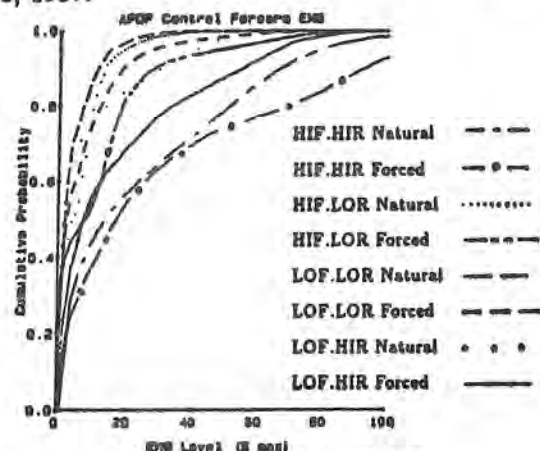


Figure 2: APDF of Forearm EMG

A SYSTEM TO PREDICT INTERNAL FACTORS RELATED TO THE DEVELOPMENT OF CUMULATIVE TRAUMA DISORDERS OF THE CARPAL TUNNEL AND EXTRINSIC FLEXOR MUSCULATURE¹

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Introduction

Cumulative trauma disorders of the upper extremity have been linked to tasks with high repetition rates, extreme or sustained postures and/or high or sustained forces. During task analysis for reduction of cumulative trauma disorders (CTD's), qualitative guidelines such as presented by Armstrong et al (1979) can be used. This approach allows identification of gross injurious single variable factors (ie injurious postures or high forces) but quantified guidelines are nonexistent and it is unclear how the risk factors interrelate. A system was required to account for the contribution of all three factors force, repetition, and posture over the duration of a task, a measure not previously possible. The carpal tunnel and extrinsic flexor musculature site was chosen because it represents a large part of CTD claims, and also contains examples of several different types of CTD's: nerve compression (Carpal Tunnel Syndrome), tenosynovitis, tendinitis, muscle soreness (myalgia) and chronic pain.

System Development

A system was developed that takes data on hand and wrist postures, grip forces, hand function and forearm EMG levels used throughout performance of a task, and creates internal factors from predictions of soft tissue loading from a model of structures in the carpal tunnel (see Figure 1). The data is collected using a system described elsewhere (Wells et al, 1988). The internal factors studied were developed logically from the known risk factors involved in CTD's and possible injury causing mechanisms for each of the different CTD's investigated (see Table 1). To calculate the factors, the wrist is modeled as a pulley with the finger flexor tendons passing over it (Armstrong and Chaffin, 1979). The radius of curvature and the tendon wrap angle both vary as a function of wrist angle. The radius is calculated using a geometric function created from the tendon excursion associated with the current wrist angle and whether the worker is in flexion or extension. The axial force in the tendons is calculated using the normalized results by An et al (1985) for different hand functional grips. The normal pressure factor represents the average pressure of the flexor tendons on the surrounding tissues as they pass through the carpal tunnel, calculated using a standard belt/pulley analysis. The results for flexion and for extension are accumulated separately in the impulse measure due to their effect on different tissues. Tendon excursion, as seen at the proximal end of the carpal tunnel, is calculated by summing the effect of movement of the appropriate wrist and finger joints, using cadaver measurements by Armstrong and Chaffin (1978). Two factors (frictional work and power) reflect the transfer of energy from the tendon to the surrounding tissue. They are calculated using the normal pressure and excursion results, and best quantify the combined effect of force, repetition and posture. The amplitude probability distribution function of the EMG from the forearm flexor musculature is used to describe the physiological load in the muscle. Many of the factors are summated temporally (ie impulse, total excursion) as CTD's are the result of repeated or continuous low level loading over a long period of time.

Conclusion

The system described bridges the gap between study of the task and study of hand function. It will be used to analyze a variety of injurious and non-injurious tasks, to statistically predict a loading level which might be used as a quantified guideline in task analysis. It will also allow study of the relationship between force, repetition and posture and give insight into injury mechanisms for various cumulative trauma disorders.

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Table 1: Relationship Between Internal Factors, CTD's and Injury Mechanisms

CTD	Possible Injury Mechanisms	CTD Risk Factor	Possible Internal Factor
Carpal Tunnel Syndrome	-Force of Tendons on Median Nerve	-Force -Posture -Time	-Peak Normal Pressure -Impulse of Normal Pressure During Flexion
Tenosynovitis	-Force of Tendons on Synovial Sheath	-Force -Posture -Time	-Peak Normal Pressure -Impulse of Normal Pressure During Flexion -Impulse of Normal Pressure During Extension
	-Movement of Tendon with respect to Tendon Sheath	-Repetition -Time	-Tendon Excursion -Peak Excursion Velocity -Average Excursion Velocity
	-Friction Between Tendon and Tendon Sheath	-Force -Repetition -Posture -Time	-Frictional Work Factor -Peak Frictional Power
Tendinitis	-Strain in the Tendons	-Force -Time	-Peak Tendon Axial Force -Impulse of Axial Force
Myalgia	-Muscle Fatigue and Overuse	-Force -Posture -Repetition -Time	-Static Muscle Load (5th Percentile APDF) -Dynamic Muscle Load (50th Percentile APDF) -Peak Muscle Load (95th Percentile APDF)

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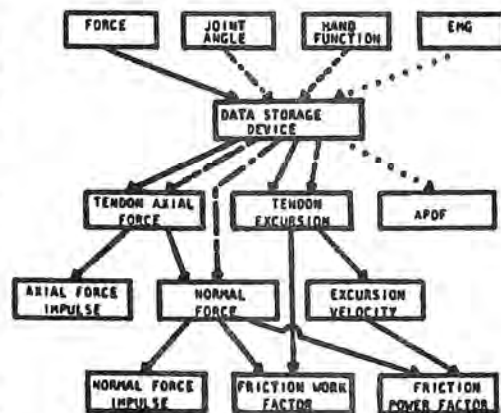


Figure 1 System to Predict Internal Factors

L4-L5 SHEAR FORCE REDUCTION BY LOW BACK MUSCULATURE DURING LIFTING

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Introduction

Biomechanical models have been used as a non-invasive means of estimating the compression and shear forces experienced by the lumbar intervertebral discs during lifting. Models proposed by Chaffin (1969) and others have represented the extensor tissues of the lumbar spine with a single equivalent force generator parallel to the line of compression and at a distance 5cm posterior to either the L4-L5 or L5-S1 disc. These models, although highly simplified anatomically, have served a valuable function in providing a method for the assessment of risk for disc compression types of injuries. However, cadaver studies (Macintosh and Bogduk, 1987, McGill and Norman, 1987) have indicated that the lumbar laminae of the longissimus thoracis (LT) and iliocostalis lumborum (IL) have lines of action which are actually oblique to the compressive axis of the spine (Fig. 1). The forces generated by these muscles have posterior components which would tend to shear the superior vertebrae posteriorly on its inferior counterpart. This could partially compensate for anterior reaction shear forces that result from gravitational forces acting on the flexed upper body and load. A multiple muscle, biomechanical model proposed by McGill and Norman (1986) has incorporated this obliquity into the representation of the lumbar musculature.

The purpose of the present study was to compare shear forces acting on the L4-L5 motion unit estimated by this model with results from a single equivalent model. Knowledge of the joint shear forces will enable further assessment of the link between shear loads and low back injury.

Methods and Analysis

Fifteen subjects performed five squat (bent knee) lifts with loads of 6, 14, 22, 29 and 32 kg. A sixth lift was performed with one of the loads repeated. A multiple muscle, EMG driven, dynamic, sagittal

plane model (McGill and Norman, 1986) was used to estimate the forces in trunk muscles, ligaments, and on the L4-L5 disc during each lift. Joint forces were also estimated from the single equivalent model. The peak shear force for each trial was determined and averaged for all of the subjects at each of the 5 load conditions. Ensemble average curves were calculated with force-time history data of shear force outputs from both models and the lower erector spinae force estimates from the multiple muscle model (32 kg load condition). An attempt was made to eliminate subject size effects in the ensemble analysis so only 9 subjects of similar height and weight were used.

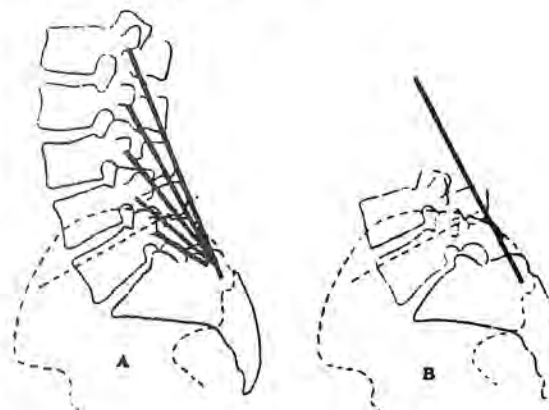


Figure 1. Anatomical representation of the lower erector spinae in the a) multiple muscle model and b) single equivalent model.

Results and Discussion

The average peak shear force calculated with the multiple muscle model remained at a constant value of approximately 200 N regardless of load but rose, predictably, almost linearly with load when calculated with the single equivalent model (Fig. 2). The difference between the means from each model was significant ($p < 0.005$) for each of the 5 load conditions. The average shear force over the entire

lift duration ($n=9$) was observed to increase to approximately 700 N with the single equivalent model but it never exceeded 200 N with the multiple muscle model (Fig. 3). Figure 4 illustrates the relationship between the external reaction shear force and the lower erector spinae forces which allows the shear load on the disc to remain relatively constant throughout the lift. The shear force remained low for all loads tested because EMG activity, and corresponding muscle forces in each of the oblique, shear offsetting laminae, increased with load.

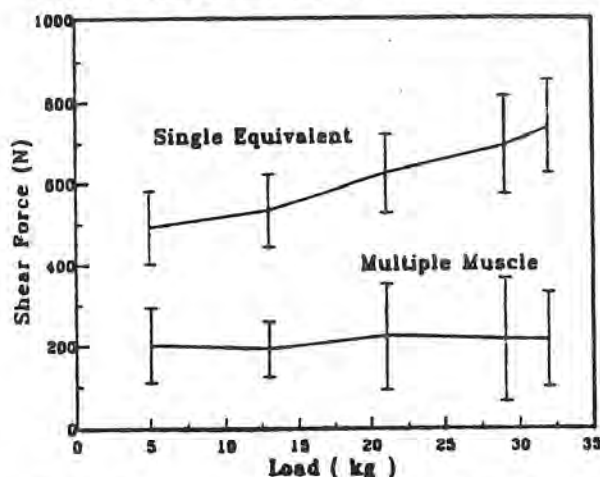


Figure 2. Average peak shear force at each load condition as estimated with both models.

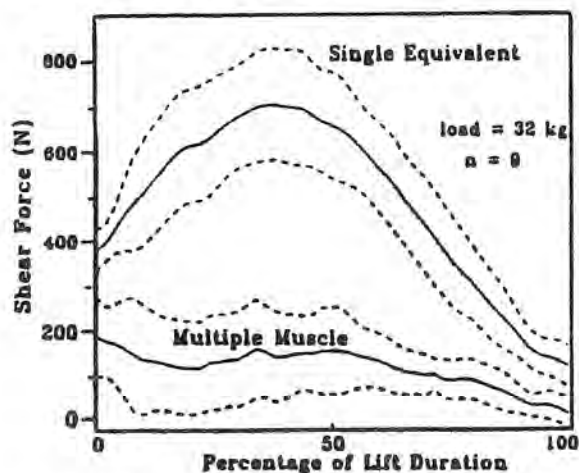


Figure 3. Ensemble average of the shear force over the entire lift duration as estimated with both models.

The finding that joint shear loads remained virtually constant and quite low regardless of the load lifted is important for those interested in determining mechanisms of injury. The disc itself and other structures of the motion unit may be vulnerable to

anterior shear forces but with the range of subjects and load conditions used in this study the multiple muscle model estimates indicate that the large shear forces produced by the weight of the load and the upper body are effectively offset by forces in the low back musculature. Thus, the motion unit was never heavily loaded in shear. This does not mean that shear force cannot play a role in injury under extreme conditions but it does indicate that either because of the design of the musculature in the lumbar spine, neurological control in activation, or both, the shear forces on lumbar motion units is reduced to low levels. Further, it would be erroneous to speculate on the role of shear forces in injury when calculated from the single equivalent model.

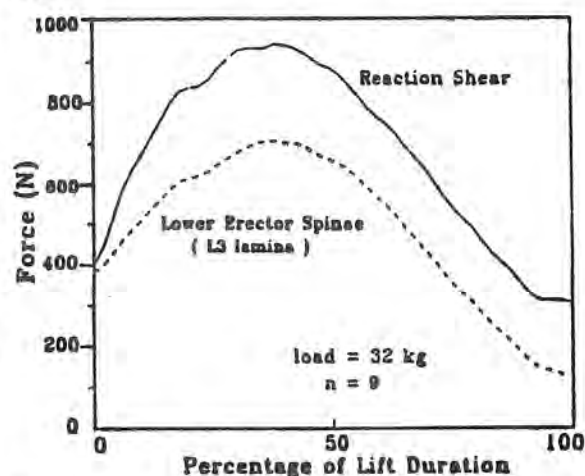


Figure 4. Illustration of the relationship between the forces in the lower erector spinae musculature and the anterior reaction shear force.

Acknowledgement

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A COMPARISON OF SELECTED HEADFORMS UNDER IMPACT CONDITIONS

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Several standards and testing protocols exist for evaluating protective helmets used in activities such as football and baseball (NOCSAE), ice hockey (CSA, ASTM) and bicycling and motorcycling (ANSI, CSA, ASTM). The criterion measure of helmet performance is usually peak headform acceleration, but different headforms are used by each standards agency. This is a particularly acute problem when the development of an International Standard, such as that for ice hockey helmets, is concerned.

PURPOSE:

The purpose of this study was to determine differences in the performance characteristics, as measured by peak headform acceleration, of several headforms under a standardized test protocol.

METHODOLOGY:

Two models of 4 headform types were used: ASTM models C and D, an aluminum-magnesium alloy; ISO models M and J, an aluminized epoxy conforming to ISO sizing standards; Hodgson-WSU, sizes MED and LARGE, with humanoid features; Swedish Ice Hockey Federation, sizes MED and LARGE, made of hardwood and conforming to ISO sizing standards. Each headform was subjected to 2 successive impacts, using a monorail drop test rig at 7 test locations, including Front, Rear, Right and Left Side, Crown, Right Front Boss (RFB) and Lower Left

Rear Boss (LLRB). Headforms were dropped from heights of .20m, .40m and .60m, against a flat rigid anvil covered with a modular elastomer programmer pad (MEP), with a Shore D hardness of 35. Headform acceleration was determined with a Bruell and Kjaer uniaxial accelerometer and the signal was displayed on a Nicollette 3090 storage oscilloscope. A Vibrametrics triaxial accelerometer was used for the WSU headform tests. Impact velocity was recorded electronically.

RESULTS:

Peak headform accelerations for the two successive impacts were averaged for analysis. The response to impact between headforms, and within headforms at different test locations, was not uniform (see Figure 1). Highest peak accelerations were recorded on the Crown and Side and lowest on the Right Front Boss and Lower Left Rear Boss.

Because the mass of the various headforms ranged from 4.74 to 6.51 kg., it was felt that the differences noted in the headforms may have occurred as a result of differences in impact energy. To overcome this, the data for

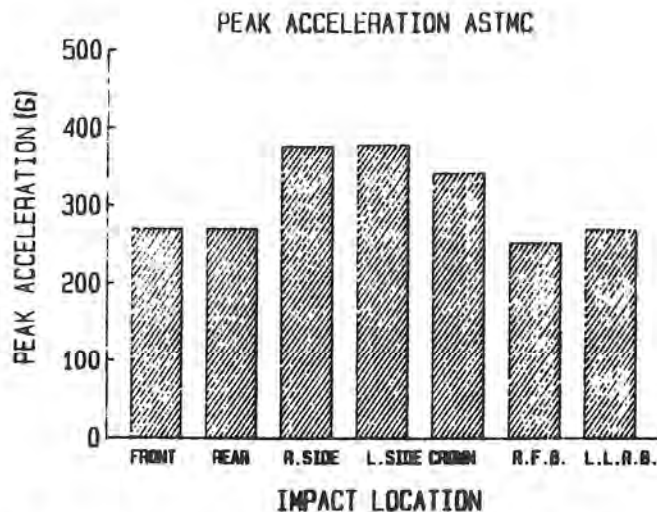


Fig. 1. Peak Acceleration by Impact Location.

each headform type was combined and the peak headform acceleration was then plotted as a function of impact energy, and the appropriate linear regression lines were drawn using the GRAPH PAD software package (see Figure 2).

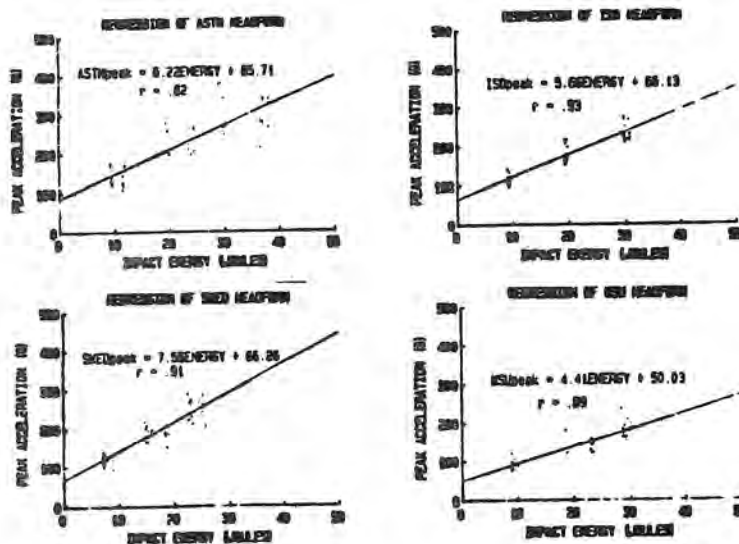


Fig. 2. Regression Equations for 4 Headform Types.

Using this approach it was then possible to arrive at a reasonable estimate of an

appropriate tolerance criterion for each headform type when subjected to a standardized impact energy.

For example, for an impact energy of 51J (i.e., 5.1kg headform dropped from 1.0m), as recommended in the proposed ISO hockey helmet standard, a pass/fail criterion of 275g has been suggested. This value represents 68% of the predicted headform acceleration for the unhelmeted ASTM headform (403g). In other words, the helmet is expected to reduce the peak headform acceleration by at least 32%. Assuming that this percent reduction is desirable, the pass/fail criterion for each headform type, estimated from the regression lines in Figure 2 is: ASTM (275g); ISO (250g); WSU (190g); SWED (315g).

DISCUSSION:

These results were obtained on only a small sample of headforms and must be regarded as preliminary. Determining and plotting peak headform acceleration versus impact energy takes into account differences in actual impact energy which would not be apparent if only drop height was considered. A pass/fail criterion should not be a universal value but should be considered and determined in conjunction with the headform used.

ACKNOWLEDGEMENTS:

This study was sponsored and supported by the Canadian Standards Association.

ANTICIPATORY POSTURAL SUPPORT RESPONSES: EFFECT OF DIRECTION AND TYPE OF INTENDED MOVEMENT

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INTRODUCTION

In an unpredictable environment, we are often subjected to perturbations that can destabilize our balance. Because of our body structure that includes a large mass of the head, arm and trunk supported on two narrow structures, the legs, balance control is particularly critical. How the neuromuscular system responds to unexpected perturbations to maintain balance has been studied by several researchers. A common tenet that has emerged, to simplify control of multiple muscles, is the notion of synergies: muscle responses organized into distinct and discrete units for specific perturbations (cf. Nashner & McCollum, 1985). Recently (McPherson, 1988a,b; Rushmer et al., 1988) researchers have examined in detail postural responses in cats subjected to translations of platform in various directions. They have shown that muscle responses are modulated such that no distinct muscle synergies exist for each direction. McPherson (1988a) from the analysis of ground reaction force data suggests that the system controls these forces and the muscle responses are organized as task dependent units at a lower level. In contrast to these studies, the study of anticipatory postural control has been limited. The majority of the work in this area has been limited to examining responses to voluntary perturbations of the upper body (cf. Cordo & Nashner, 1982; Bouisset & Zatarra, 1981). The focus of this study is to describe the anticipatory postural support responses when the subjects are required to move their lower limb in different directions.

METHODS

Subjects (n=9) were required to stand with the ipsilateral limb on the force plate. In response to a light cue, they were instructed to flex the ipsilateral lower limb at the hip forcefully and quickly in one of seven directions (0°, 45°, 90°, 135°, 180°, 225°, 315°) as shown in Figure 1. Ten trials per condition which were blocked were collected. The conditions were randomized. Rectified and low pass filtered (100 Hz) EMG activity from the following ipsilateral (rectus femoris - IRF; biceps femoris - IBF; medial gastrocnemius - IMG) and contralateral (rectus abdominus - CRA; erector spinae - CES; rectus femoris - CRF; biceps femoris - CBF) muscles along with ground reaction forces (Fy - vertical; Fz - medio-lateral; Fx - anterior-posterior) were collected online (sampling rate = 300 Hz).

DATA ANALYSIS

Each trial was windowed from the onset of the light cue, and the first five trials of each condition were ensemble averaged. A threshold file of quiet standing was used to determine the normal variability in each channel. From this file the band representing $\bar{X} \pm 2$ S.D was calculated and used to determine the onset of the response. From the averaged record the latency of responses in each channel, the average magnitude and duration of the first response were determined. An ANOVA on the time when the limb was lifted off the force plate revealed no effect of various conditions ($\bar{X}$ - 634 ms). Therefore the responses were set with respect to this event, and ANOVA was done within each condition to determine the order of muscle responses. ANOVA was also done on the latency, magnitude and duration across condition. Posthoc analysis using Neuman-Keul method ($p < .05$) was done to identify common groupings among variables.

RESULTS AND DISCUSSION

The ordering and magnitude results are summarized in Figure 1 and Table 1. In Figure 1, the order of responses within each condition are shown in groupings that were statistically similar. The discussion will concentrate on the first group for each condition. In Table 1, the magnitude of the responses are grouped by letters across conditions. Similar magnitudes are identified by common letters, while the relative magnitude can be seen in the ordering of the alphabets (A > B > C...).

The onset of change in Fy component did not vary across condition. Thus whatever measure one uses, the reaction time across condition did not vary. The increase in Fy component is necessary to initiate movement while change in Fz component represents the weight transfer to the contralateral limb. The magnitude and latency of both of these were identical across conditions. This invariance is similar to those found by Mcpherson (1988). The Fx component on the other hand showed a direction dependency as expected. The first grouping of muscle responses reveal two important points. First, the IMG response occurs for all conditions (its latency was the same across condition) to initiate the limb movement.

The magnitude of IMG in contrast is modulated: it is bigger in the forward and sideward direction and smaller in the backward direction. This may be due to the changes in Fx component across condition. The second observation is that the first postural muscle to be activated was CES in the forward direction and CRA in the backward and sideward direction: representing anticipatory action to counter balance trunk movements that will be caused by the intended movements. Other appropriate postural muscles are also activated in specific directions to assist in this stabilization. The results concur with recent observations that postural control at higher levels may not be implemented at the muscle level, but rather a higher order parameter such as overall force.

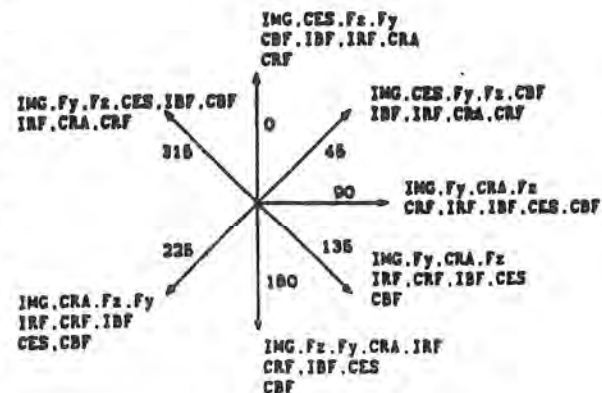


Figure 1: Order of responses.

Table 1: Summary of the magnitude of the responses.

	0°	45°	90°	135°	180°	225°	315°
IRF	A	A	C	C	C	C	A
IBF	C	D	D	C	C	A	C
IMG	A	A	A	D	D	D	A
CRF	A	B	B	A	A	A	A
CBF	A	A	C	A	C	C	A
CES	B	B	B	B	A	A	B
CRA	A	A	A	A	A	A	A
Fy	A	A	A	A	A	A	A
Fz	A	A	A	A	A	A	A

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ACKNOWLEDGEMENTS

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VISUAL CONTROL OF LOCOMOTION: STRATEGIES FOR CLEARING OBSTACLES

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INTRODUCTION

Numerous studies on gait have provided a rich data base on "normal" unobstructed gait motor patterns (cf. Winter, 1987), and has improved our understanding of how these very important movements are controlled (cf. Patla, 1985). Many researchers have also examined how these patterns are modulated in response to unexpected (cf. Nashner, 1980) and expected perturbations (cf. Patla, 1986). But how we use vision to guide and adapt locomotor patterns over irregular terrain has received limited attention. Visual control of step length (Patla et al., 1988) during locomotion has shown the complex task specific nature of the adaptations to the locomotor patterns. Regulation of step length is one way to navigate in an irregular environment. Often when faced with obstacles, we modify our gait to go over them and continue with our task. The focus of this study has been to examine the strategies adapted by subjects to modify gait to go over obstacles.

METHODS

Subjects (n=6) were instructed to walk at their normal self-paced speed and go over the obstacle with the ipsilateral limb that was introduced randomly in the trials. The obstacle were two aluminum plates (L - 2 cm, H - 8 cm) that were raised by solenoids. The solenoids were triggered by mats placed along the walkway at two different times (N - 1 step duration in advance; E - 2 step duration in advance). The obstacles were placed after the force plate approximately in the middle of the step. Telemetered raw EMG signals from the following ipsilateral (biceps femoris - IBF; rectus femoris - IRF; tibialis anterior - ITA) and contralateral (biceps femoris - CBF; rectus femoris - CRF; soleus - CSO) muscles along with ground reaction forces (vertical - Fy and horizontal - Fx), and foot mat signals were collected online (sampling frequency = 1000 Hz). Ten trials for each of the following conditions were collected: control (NC) and five experimental conditions (NN, LN, HN, LE, HE). All the experimental trials were randomized. The subjects were also videotaped.

DATA ANALYSES

The EMG signals were full wave rectified and low pass filtered (10 Hz). The successful trials were rubberbanded (IHC to IHC) and then ensemble averaged. To quantify changes in gait patterns, the following values were determined: pushoff and breaking impulses in Fy (IyP, IyB) and Fx (IxP, IxB) as guided by the Fx curve; stance, step and double support duration; the normalized (to NC condition) average EMG values for three phases - 30-50%, 50-70%, 80-100%; the absolute trunk and relative hip, knee and ankle angles when the limb was above the obstacle. An ANOVA was done on these variables and posthoc analysis was carried out using Neuman-Keul method ($p < .05$).

RESULTS

The results are summarized in Tables 1, 2 and 3. Subjects were generally successful in clearing the obstacles (approximately 8-9% of leg length for the high obstacle).

Table 1: Summary of significant muscle responses. Arrows show the direction of change.

Phase Condition	0-30%	50-70%	80-100%
LN	—	↑ ITA	↑ CBF ↑ IRF
HN	—	↑ IBF ↑ ITA	↑ CBF
LE	—	—	—
HE	↑ IBF	↑ IBF	↑ CBF ↓ IBF

Table 2: Summary of impulse components and temporal data. Same letters are used to group conditions with statistically similar values; the order of the alphabet is used to signify relative magnitude.

Condition Parameter	NC	NN	LN	HN	LE	HE
IyB	D	D	A	A	A	A
IyP	A	A	A	A	A	A
IxB	A	A	C	C	C	E
IxP	B	B	B	B	B	A
Stance	E	E	E	A	E	A
Db. Support	A	A	A	A	A	A
Step Dur.	E	E	E	A	E	A

Table 3: Summary of limb position above the obstacle. Arrows indicate direction of change.

	LN	HN	LE	HE
Trunk Angle	--	--	--	--
Hip Angle	--	↑ flexion	--	↑ flexion
Knee Angle	--	↑ flexion	--	↑ flexion
Ankle Angle	--	↑ flexion	--	--

DISCUSSION

It is clear that the strategies adopted by the subjects are both dependent on the height of the obstacle to be cleared, and the time when the subject sees the obstacle. For the low size obstacle the results suggest minimal changes in ipsilateral muscle activation profiles (ITA), while for the high obstacles subjects were actively flexing at the knee (IBF) which will through mechanical coupling also cause hip flexion to assist in the task. Activation of hip flexor (IRF) would be inappropriate because of its extensor action at the knee. Thus the system selects functionally appropriate response. The increase in first component of the vertical impulse for all conditions suggests raising of the center of mass. Thus for low height obstacles, this change in body posture was enough, but for higher obstacles ipsilateral limb also has to be flexed to clear the obstacle. The effect of different times when the obstacle is presented is clearly seen in the ITA response for "N" conditions compared to the "E" conditions. Therefore later the cue time, subjects chose to increase their safety margin by flexing at all three joints compared to flexing at the hip and knee for the early cue condition. The task dependent modulation of the locomotor patterns suggest that these patterns are accessible and modifiable online.

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