

TRUNK EXTENSION AND FLEXION TORQUE IN ATHLETES

S. Ikegawa, M. Shimoda, H. Nagareda, K. Kubo, J. Okada, Y. Kawakami, K. Funato and T. Fukunaga
Tokyo Metropolitan College and University of Tokyo, Tokyo, Japan

INTRODUCTION

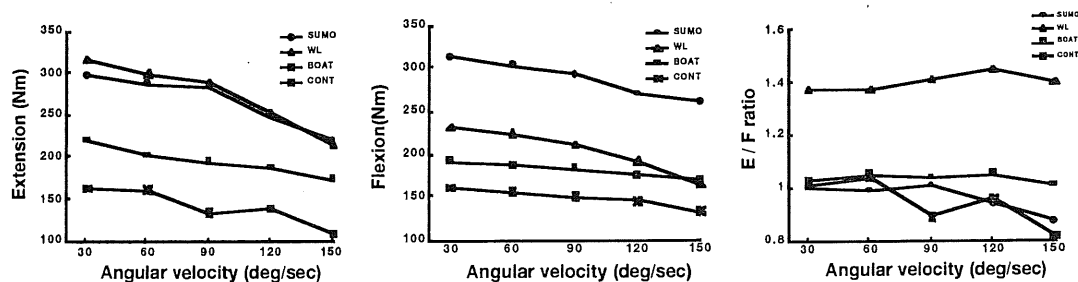
Trunk muscle strength has been used as a diagnostic parameter for low back pain. Several investigators have pointed out that patients with low back trouble have lower strength of trunk extension than normal people (Mcneil et al. 1980; Kumar et al. 1995). However, characteristics of athletes with respect to trunk muscle performance have not been widely studied. The aim of the present study is to measure and compare isokinetic trunk extension and flexion torque in various athletes.

METHOD

Subjects consisted of 6 sumo wrestlers, 7 weightlifters, 23 rowers and 4 untrained controls. They were all university students. Isokinetic trunk extension and flexion torque was measured by Cybex 770 with a trunk extension and flexion unit. Angular velocities were selected at 30, 60, 90, 120, and 150 deg/s. The range of motion was set between 0 (upright position) and 80 deg of trunk flexion.

RESULTS AND DISCUSSION

Trunk extension torque of sumo wrestlers was 297 (SD:30) Nm at 30 deg/s and decreased with increasing angular velocity to 219 (48) Nm at 150 deg/s. Weightlifters developed 315 (78) Nm at 30 deg/s which was reduced to 213 (70) Nm at 150 deg/s. Trunk extension torque of both sumo wrestlers and weightlifters was significantly greater than rowers and control subjects at all angular velocities. Sumo wrestlers had significantly greater trunk flexion torque of 312 (78) to 259 (51) Nm at 30 - 150 deg/s than the other three groups (230-130 Nm) at all velocities. When the torque was divided by body mass, however, no significant difference was observed among the groups both for trunk extension and flexion. The extension / flexion torque ratio (E/F ratio) of weightlifters (1.4) was significantly higher than the other groups (1.0 - 0.9) at faster velocities. These results would indicate that trunk extension and flexion strength is mainly determined by the mass of lean tissues, and that the E/F ratio represents athletic characteristics.



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Biomechanical Analysis of Chair-Rise and Picking up From Floor in the Pregnancy

Yu-Chi Chen, You-Li Chou, Hsieh-Ching Chen*

Institute of Biomechanical Engineering of National Cheng-Kung University

Department of Industrial Engineering and Management of

Chaoyang Institute of Technology*

INTRODUCTION

There are so many changes in physiological and psychological aspects during pregnancy. Chair-rise and picking up from floor are the most difficulty performances to the pregnant women. These changes impose postural demands and limit the performance of activities of daily living. The overall goals of this research are to quantify the effects of physiological change due to pregnancy on the performance of chair-rise and picking up from floor, and to analyze why some pregnant female have more difficulty with these tasks from the biomechanical point of view.

METHOD

Twenty-one pregnant women are recruited by staff of Cheng-Kung University Hospital. We divided the subjects to three groups which are the first stage (1st), the second stage (2nd), and the third stage (3rd). Three force-plates were used to measure the reaction forces of foot, seat and armrest. By kinematic and kinetic analysis, the angles and moments of the hip, knee and ankle joints were analyzed. The reaction forces of foot and center of pressure (COP) while picking up from floor were also analyzed.

RESULTS

Chair-rise

The chair height had great influence on knee joint and hip joint moments, but less on ankle joint. In the third stage (chair height 65% KH), because of a marked increase in abdominal depth, the maximum hip moment significantly decreased. But the maximum knee moment greatly increased.

Picking up from floor

The results indicated that the third-stage pregnant women need much time in these tasks. For the values of reaction force of foot, the third-stage pregnant women are less than the ones at the first stage whether foot is symmetric or asymmetric placement. On the maximum value of the weight shifting and COP, the third-stage pregnant women have significantly increased.

DISCUSSION

Chair-rise

In the third stage (chair height 65% KH), because of a marked increase in abdominal depth, the maximum hip moment significantly decreased. But the maximum knee moment greatly increased. This is the reason why pregnant female have more difficulty on the performance of chair-rise. If the pregnant female rise with the armrest, the maximum knee moment significantly decreased.

Picking up from floor

The maximum value of the weight shifting and COP, the third-stage pregnant women have significantly increased. This reason is the third-stage pregnant women need more weight shifting to adapt their posture change.

CONCLUSION

In the third stage (chair height 65% KH), because of a marked increase in abdominal depth, the maximum hip moment significantly decreased. But the maximum knee moment greatly increased. This is the reason why pregnant female have more difficulty on the performance of chair-rise. The pregnant female rise with the armrest, the maximum knee moment significantly decreased. Pregnant female should keep upright stance with feet placed asymmetrically while picking up from floor. This posture will reduce the weight imbalance.

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THE RELATIONSHIP BETWEEN BODY MASS INDEX AND PLANTAR PRESSURE DISTRIBUTION AND LOWER LIMB ALIGNMENT IN PREPUBESCENT CHILDREN

D. L. Riddiford, J. R. Steele, and L. H. Storlien
University of Wollongong, Wollongong, Australia

INTRODUCTION

Numerous studies have indicated that the proportion of overweight and obese children in Western industrialised societies has increased¹. An increase in body mass has been accompanied by an increase in health related problems, including musculoskeletal symptoms. However, no studies were located which have examined the effects of this increase in body mass on lower limb structure in children. Therefore, the purpose of this study was to investigate the relationship between body mass index (BMI) and plantar pressure distribution and lower limb alignment in prepubescent children.

METHODS

Height and weight of consenting Grade 3 students (n=429) from randomly selected primary schools in the Wollongong Education District were measured to calculate BMI. Footprints and slides of both lower limbs of each subject were taken in static weightbearing using a pedograph and 35 mm camera, respectively. Footprints were analysed to calculate the Footprint Angle (FA) and the Chippaux-Smirak Index (CSI)². Calcaneal Angle (CA) and Tibial Angle (TA) relative to the right-hand horizontal were measured from the posterior slide view of each subject's lower limbs³.

RESULTS

Results of the study are presented in Table 1.

Table 1: Age, BMI, plantar pressure distribution and lower limb alignment data for prepubescent children.

Variable	Girls (n=219)		Boys (n=210)		Total (n=429)			
Age (yrs)	8.4±0.5*		8.5±0.5		8.5±0.5			
BMI	17.9±2.9		17.7±2.9		17.8±2.9			
	Left	Right	Left	Right	Left	r	Right	r
FA (°)	46.1±11.6	46.2±10.7	41.6±14.1	41.2±13.6	43.9±13.1	-0.23 ^r	43.7±12.5	-0.23 ^r
CSI (%)	25.1±14.9	26.1±13.7	26.7±15.4	28.3±14.5	25.9±15.1	0.39 ^r	27.2±14.1	0.37 ^r
CA (°)	-5.4±4.1	-2.6±4.3	-5.2±4.2	-2.3±4.2	-5.3±4.3	0.02	-2.5±4.3	0.03
TA (°)	-0.4±3.6	0.8±3.9	-0.4±3.9	0.8±3.8	-0.4±3.7	-0.03	0.8±3.8	-0.05

* data are reported as the mean ± standard deviation.

^r indicates a significant correlation ($p < 0.05$) with BMI.

DISCUSSION

Although no relationship was found between BMI and lower limb alignment, there was a significant relationship between BMI and plantar pressure distribution. That is, as BMI increased, FA decreased and CSI increased (see Table 1). A decreased FA and an increased CSI have been associated with a lower longitudinal internal arch, a flatter cavity and a wider midfoot area of the footprint^{2,4}. Lower arches have also been associated with a decrease in the integrity of the foot as a weightbearing structure. Therefore, it was concluded that excess body mass may have a negative effect on the foot structure and function of prepubescent children. Further research is recommended to determine whether lowered arches in childhood are detrimental to the natural growth and development of the foot.

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BIOMECHANICAL EVALUATION OF WRIST MOVEMENT DURING MEATCUTTING.
K.SØGAARD, B.JUUL-KRISTENSEN, L.BIRCH, H.CHRISTENSEN
National Institute of Occupational Health, Copenhagen, Denmark.

INTRODUCTION

Prevalence for musculoskeletal symptoms from the wrist and hand is among Danish meatcutters about 6 times higher than the prevalence among the wage earners in general. In a field study including 48 meatcutters direct observation showed for the hand holding the knife that during a large part of the workcycle the wrist was working in an extreme position e.g. flexed or extended more than 15° or ulnar deviated more than 20°. However the repetitiveness measured as shifts between extreme and neutral positions was low. The EMG results showed a high static activity level of 8% and 9% of maximal handgrip activity for the extensor and flexor muscle, respectively. Apart from known occupational risk factors for hand wrist disorders such as extreme wrist angle, high repetitiveness and high grip force level, also dynamic parameters e.g. velocity and acceleration has been shown to be important factors (1). The aim of this study was to describe the dynamic parameters during two typical meatcutting processes in order to estimate if they add to the risk factors for hand/wrist disorders among meatcutters.

METHODS

Wrist movements from two representative meatcutters (evaluated by the observation method) were documented from video recordings obtained from two cameras placed perpendicular to each other. Three-dimensional coordinates of the joint centers were obtained using Peak Performance. From these data position as well as motion components in flexion/extension and radial/ulnar planes of the wrist were calculated. For each plane range of motion and duration in extreme positions were calculated as well as mean and maximum velocity and acceleration. EMG from the bulky part of the wrist flexor and extensor muscles were synchronously recorded using surface electrodes. From videorecordings the workcycle was divided into periods with the knife in and off the meat. EMG from these periods were compared and used as an indication of the contribution from the external force to the muscle activity.

RESULTS

Table 1.

	Flexion		Extension		Ulnar deviation	
Range of motion (°)	0-15	>15	0-15	>15	0-20	>20
% of workcycle	20	6	42	33	71	24
Mean position (°)	6	22	8	25	13	34
Mean velocity (°/sec)	47		45		46	
Max velocity (°/sec)	244		254		350	
Mean acceleration (°/sec ²)	461		421		428	
Max acceleration (°/sec ²)	3039		2946		3755	

Results are given in Table 1 as mean of the two work processes analysed. In addition it was calculated that in 18% of the workcycle time the wrist was simultaneously in extreme ulnar deviation and extreme flexion or extension. This detailed analysis therefore confirmed that during meatcutting the wrist is in extreme positions during a large part of the workcycle. This was confirmed by a significant increase in EMG activity from the extensor muscle when periods with the wrist in extreme positions was compared to periods with the wrist in normal positions. Although the repetition of movement was low, the mean and maximum velocity in all directions of movements as well as acceleration in radial/ulnar plane were of a magnitude, which has been shown to increase the risk for hand/wrist disorders in high repetition tasks without handtools (1). Compared to these work tasks the mean and maximal acceleration in flexion/extension plane was low and of similar magnitude in both directions. However, due to the mass of the knife and the external force against the knife during cutting, even hand accelerations of a magnitude as seen in this study may greatly increase the force demand on the wrist muscles. This was confirmed by a significant increase of 41% in flexor muscle activity, when the knife was in the meat compared to when it was held freely.

CONCLUSION

From the kinematic data it is concluded that the high velocity of wrist movement places meatcutters in a high risk group. Furthermore, wrist acceleration plays an important role regarding force demands during work with hand held tools. Therefore reference values for wrist acceleration indicating high risk groups in this type of work are badly needed.

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EVALUATION OF PALMAR GRASP IN PEOPLE WITH TETRAPLEGIA USING A MATHEMATICAL MODEL OF THE HAND

A. Esteki* and J.M. Mansour**

*Biomedical Engr. Dept., Faculty of Medicine, Shahid Beheshti Univ. of Medical Sci., Tehran, IRAN

**Dept. of Mechanical Engr., Case Western Reserve Univ., Cleveland, Ohio, USA

INTRODUCTION

To improve hand function in people with tetraplegia using Functional Neuromuscular Stimulation (FNS) to restore their hand function a three dimensional dynamic mathematical model was developed. In this part of the study, the model was used to evaluate palmar grasp function of the hand, in situations typical of those encountered when using FNS. Several static and kinematic models of normal hand function (1, 3) have been developed and used to evaluate prehensile capabilities of the hand. A few models of human hand dynamics (2, 4) have also been introduced to study nonprehensile functions of the normal hand. Our model includes most of the details other static and dynamic models in the literature have, as well as the unique characteristics of the hand of a person with tetraplegia. It has the capability to evaluate static or dynamic performances in palmar grasp function of both normal and abnormal hands.

METHOD

The thumb, index finger, palm, inter-segmental joints, and fifteen muscle-tendon units typically used in upper extremity FNS systems are included in the model. Each joint is subjected to muscle moments, passive joint moment and moments due to grasp forces. Several features of a tetraplegic hand were incorporated into the model. Palmar grasp of cylindrical objects of several diameters and weights, with normal, weak and transferred muscles, and with the object either fixed to the ground, or free to slide on a horizontal flat plane with friction, were modeled. Force optimization was used to search for the activation of extrinsic flexors to provide the force needed to hold an object against gravity by frictional forces. The model was built around the computer package ADAMS (Mechanical Dynamics Inc., Ann Arbor, MI.) and can perform kinematic, static or dynamic simulations.

RESULTS

A dynamic simulation of activation of the flexor digitorum superficialis showed qualitatively the same results reported by Darling and Cole (5): initial metacarpophalangeal joint extension and proximal interphalangeal joint flexion. Model simulations revealed that potential function was highly dependent on muscle strength, joint passive moments, and object size, mass, and initial position. Simulations of palmar grasp of a fixed, cylindrical object, supported by the thumb, showed a strong interaction of muscle strength and size of the object. As the weight of the object increases, the required activation increases, while the activation decreases when the size increases, i.e. large diameter light cylinders require less muscle force to grasp than slender heavy cylinders. In cases where the object was allowed to slide, it could be grasped successfully if the extensor muscle moments could overcome the passive moments to open the hand, and the object was initially positioned properly with respect to the hand. A grasp protocol in which by activation of the flexor digitorum superficialis, the object is dragged into the palm and is entrapped, and by activation of both extrinsic flexors, sufficient force to hold the object is provided was introduced and shown successful in palmar grasp and hold of movable cylindrical objects using only extrinsic muscles.

DISCUSSION

A three dimensional dynamic model of the hand was developed and its validity was checked by comparing model predictions and available results in the literature. The model was used to investigate hand performance in palmar grasp with specific application to people with tetraplegia using FNS. Simulations of palmar grasp disclosed that, combinations of weak muscles and abnormally high passive moments associated with clawed or overly flexed index finger joints, may prevent the digits from reaching appropriate postures, even with the help of common tendon transfer techniques. It was also concluded that extrinsic muscles might be sufficient to grasp and hold cylindrical, movable objects, if prior to the grasp, an open posture is achieved, and the object is placed as close as possible to the palm. However, sufficient extension of the proximal interphalangeal joint, to permit hand opening, may be hindered by one, or a combination of, inadequate intrinsic muscle function, and increased passive moment. Passive joint moments, often neglected in many models of musculoskeletal function, were shown to be an important determinant of hand function particularly in opening the hand prior to palmar grasp.

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LOWER EXTREMITY POWER AND WORK IN LEVEL AND UPHILL CYCLING

G.E. Caldwell¹, D.G.E. Robertson^{1,2}, L. Li¹, J.M. Hagberg³

¹University of Massachusetts, Amherst, MA, USA; ²University of Ottawa, Ottawa, ON, Canada;

³University of Maryland, College Park, MD, USA

INTRODUCTION

Muscular output during ergometer cycling has been quantified using lower extremity joint moment power histories (Ingen Schenau et al., 1990). However, the power delivered to the pedal is also dependent on the rate of change of segmental energies and the joint force power contributed through the hip (Ingen Schenau et al., 1990). Recently it has been shown that simulated uphill cycling displays some distinct differences from level cycling in terms of pedal forces and joint moments, particularly when the riders assume a standing posture (Caldwell et al., 1994; 1996). Uphill cycling is of interest both due to its importance in cycling competitions and for examination of neuromuscular adaptations that may be associated with the change in environmental conditions. The purpose of the present paper was to examine changes in muscular power and work output that accompany the alteration of cycling grade and posture in ergometer cycling.

METHODS

Nine elite cyclists rode a simulated 'course' on their own bicycles mounted on a computerized Velodyne ergometer, including both level and uphill (8% grade) sections, at a workload corresponding to 80% $\dot{V}O_{2\max}$. Sagittal plane kinematics (cine at 50 fps) and pedal forces (100 Hz, 12 bit A/D) were recorded. The force and kinematic data were synchronized and used in a linked segment inverse dynamics analysis of the lower extremity. Joint moment power ($M\omega$) histories were used to calculate muscular work done at each joint for 3 conditions: seated level (SL); seated uphill (SU); and standing uphill (ST). Data were expressed as functions of crank angle (0° to 360°). Four separate trials from each condition were recorded for each subject.

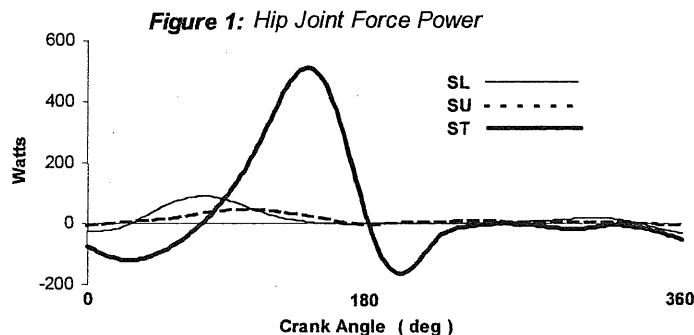
RESULTS AND DISCUSSION

The joint moment power curves for the SL condition were almost identical to those found by Ingen Schenau et al. (1990). SU power curves were very similar to SL, due to similar joint kinematics and moments (Caldwell et al., 1996). The ST condition exhibited lower peak powers at the hip and knee, due to reduced joint angular velocities, but a higher peak for the ankle due to higher joint moments (cf. Caldwell et al., 1996). Work values from the lower limb, calculated from the absolute values of the area under the power histories, are shown in Table 1. Total work was lowest in the SL condition, primarily due to shorter cycle times associated with higher cadences than the two hill conditions. The knee joint was responsible for roughly 50% of the total work in all conditions, with the hip and ankle at 30-35% and 15-20%, respectively. Significantly lower work values were seen in the SL condition for the ankle and knee compared to SU, and at the ankle only compared to ST. The higher muscular work done during the ST condition was augmented by a flow of energy through the hip joint throughout the downstroke and beginning of recovery (Figure 1). These energy transfers likely are associated with less constrained trunk and pelvis movements and with the application of forces on the handlebars while standing (Stone and Hull, 1993). The additional energy influx in the late downstroke phase between 90° and 180° coincides with the timing of the higher peak pedal force and ankle moment in the ST condition reported earlier (Caldwell et al., 1994; 1996). This additional energy source may be one reason a cyclist chooses to stand while climbing a hill.

Table 1: Muscular work (Joules)

	SL	SU	ST
Hip	48 ^a	55 ^a	46 ^a
Knee	74 ^a	90 ^b	81 ^{a,b}
Ankle	19 ^a	30 ^b	32 ^b
Total	141 ^a	175 ^b	159 ^b

*superscripts a and b represent statistically similar means



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MODELING OF RIDER'S BRAKING FORCE CONTROL IN BICYCLING BY NEURAL NETWORK WITH TIME DELAY INPUTS

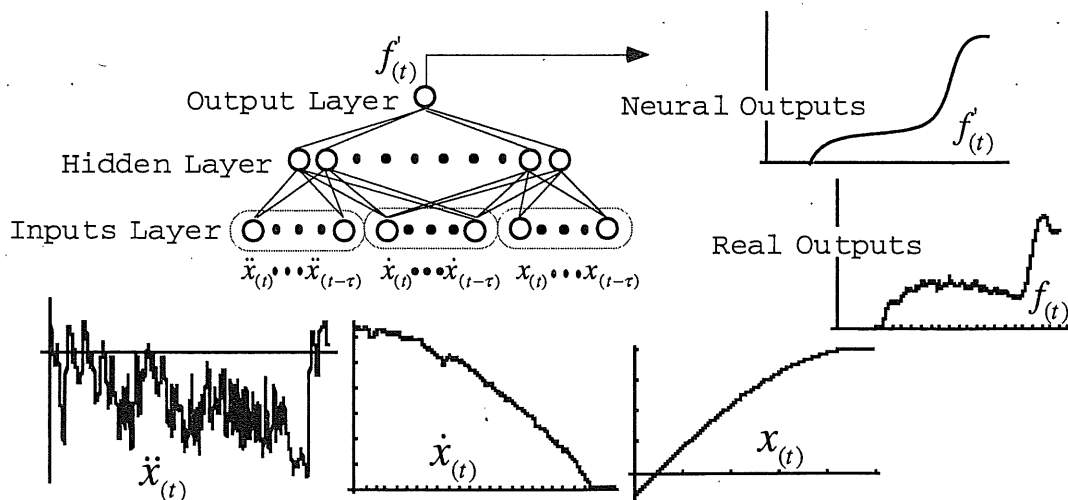
Yuji OHGI, Chikara MIYAJI, Morihiko OKADA
Univ. of Tsukuba, Tsukuba, Ibaraki, JAPAN

INTRODUCTION

In bicycle braking task, all subjects could stop very accurately on the stop line under various conditions in travelling velocity, distance of braking section and efficiency of brake mechanics[1][2]. It is obvious that this braking task had been much learned through their daily bicycling. All subjects adjusted braking force flexibly to the environmental conditions although they used experimental special bicycle in our recent studies. The purpose of this study is to identify rider's braking force control model by neural network with time delay sensory inputs.

METHOD

Time histories of acceleration, velocity, displacement, and front braking force obtained in experiment were approximated to polynomial functions. These kinematic inputs and kinetic outputs data at any time could be calculated from these functions. Rider's braking force $f(t)$ at the time of t [sec] was calculated from acceleration $\{\ddot{x}_{(t)}, \ddot{x}_{(t-1)}, \dots, \ddot{x}_{(t-\tau-1)}, \ddot{x}_{(t-\tau)}\}$, velocity $\{\dot{x}_{(t)}, \dot{x}_{(t-1)}, \dots, \dot{x}_{(t-\tau-1)}, \dot{x}_{(t-\tau)}\}$ and displacement $\{x_{(t)}, x_{(t-1)}, \dots, x_{(t-\tau-1)}, x_{(t-\tau)}\}$. Therefore the number of inputs layer will be $(\tau + 1) \times 3$, including time delay sensory inputs. Three layer model was used for this neural network's structure. The weight vector of each layer were identified by Back-Propagation method. To identify these weight vector, relative errors between neural network outputs and real outputs were evaluated. Number of hidden layer, time delay of inputs vector and weight vector of each element for every subject were determined.



RESULTS

The structure of neural network (number of hidden layer and time delay of inputs layer) and weight vector which can reproduce subject's braking control force were identified.

DISCUSSION

Between subjects variabilities can be explained by the differences of the structure (number of hidden layer and time delay inputs) or the differences of each value of weight vector of neural network. There is some possibilities, however, that outputs are different between subjects with the same structure, because of the difference of weight vector. Such a hypothesis is a new point of view to consider variabilities in human movements.

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MONOPOLAR SURFACE EMGs CAPTURE DYNAMIC FEATURES OF MUSCLE ACTIVATION UPON RAPID ISOMETRIC EXTENSIONS OF THE ELBOW

YOSHIHIKO YAMAZAKI, MASATAKA SUZUKI*, TETSUO OHKUWA, and HIROSHI ITOH

Department of Health and Physical Education, Nagoya Institute of Technology, Nagoya, Japan

* Department of General Education, Kinjo Gakuin University, Nagoya, Japan

INTRODUCTION

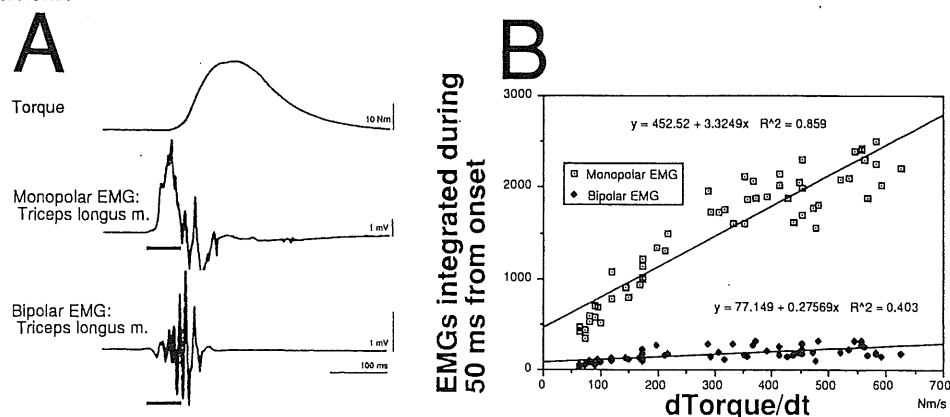
Monopolar surface EMGs (mEMGs) upon rapid contraction of the elbow were characterized by an EMG volley which showed a negative potential lasting about 50 ms and a following positive potential, whereas this is not the case with bipolar surface EMGs (bEMGs)²⁾. We examined the relationships between the EMG quantity of an initial phase and various force parameters to determine whether or not such a difference in EMGs reflected a contrast in the ability of monopolar versus bipolar methods to detect dynamic activation of muscle.

METHODS

Normal subjects (5 males) made rapid isometric contractions as rapidly as possible in the direction of extending the elbow to various visually-defined targets [9, 18, 36, and 54 % torque of maximum voluntary contraction (MVC)]. They also made continuous isometric contractions at various torque levels. EMGs were recorded from the triceps longus muscle with surface electrodes. For mEMGs, detection and reference electrodes were placed on the motor point region and the olecranon, respectively. For bEMGs, electrodes, spaced at 1-cm intervals, were placed between the two electrodes for the monopolar recording. Muscle activation was amplified with an overall band range of 7-500 Hz for mEMGs, and 20-500 Hz for bEMGs. Joint torque was measured with a strain gauge system.

RESULTS

Figure A shows torque and EMG recordings during a rapid isometric contraction targeted at 36% MVC torque. The EMG volley was apparent in mEMG but not in bEMG. Figure B shows the relationship between EMG quantities and peak values of the first derivative of torque (dTorque/dt). EMGs were rectified and integrated for quantification during 50 ms from each onset (see inset bars in Fig. A). Integrated mEMGs showed a higher correlation with peak values of dTorque/dt than integrated bEMGs. Peak values of torque and the second derivative of torque also showed a higher correlation with mEMGs (in the subject shown in Fig. B, correlations of peak torque and the second derivative of torque with integrated EMGs were $r=0.868$ and 0.884 in mEMGs, respectively, and $r=0.579$ and 0.579 in bEMGs, respectively). However, there was no such difference in correlation during continuous isometric contractions.



DISCUSSION

The results demonstrate that mEMGs are better able to capture the dynamic features of muscle activation than bEMGs. Our recent study suggested that the EMG volley was produced by an orderly recruitment of motor units in a short period as well as by a high firing rate of each motor unit^{2,3)}. Because bEMGs did not show the EMG volley, they would fail to retain such information on the dynamic features of muscle activation. On the other hand, both mEMGs and bEMGs provided equal information on muscle activation in static contractions. We propose that current methods of recording EMGs¹⁾ should be reevaluated in cases of rapid muscle contraction.

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EFFECTS OF FATIGUE INDUCED BY REPEATED DYNAMIC EFFORTS ON HAMSTRING MUSCLE FUNCTION

G. J. Pinniger, J. R. Steele, and H. Groeller
University of Wollongong, Wollongong, Australia

INTRODUCTION

Muscle fatigue is a major limiting factor to sustained physical effort and has been implicated as a predisposing factor to muscle strain injuries (Garrett, 1990). Hamstring muscle strains are incurred frequently by athletes competing in explosive activities such as sprinting (Stanton and Purdom, 1989). Although muscle fatigue has been shown to produce alterations to kinematic (Mero, Luhtanen, Komi, and Susanka, 1988) and EMG (Nummela, Rusko, and Mero, 1994) parameters of sprint running, much of this research has utilised single effort maximal sprints to elicit fatigue. No research was located investigating alterations in sprint biomechanics caused by fatigue induced by repeated dynamic efforts. Therefore, the purpose of the present study was to determine the effects of hamstring muscle fatigue induced by repeated maximal efforts on hamstring muscle function during maximal sprint running.

METHODS

Twelve skilled, male football players (mean age = 22.6 ± 2.5 years) performed three maximal 40 m sprints during which time high speed film of the subjects' sprint action and EMG of five lower extremity muscles (semitendinosus, ST; biceps femoris, BF; rectus femoris, RF; vastus lateralis, VL; and gastrocnemius, GA) were recorded (non-fatigued condition, NFC). Subjects then performed specific and general hamstring muscle fatigue tasks. The specific hamstring muscle fatigue task included three sets of 20 repetitions on a Cybex leg curl machine at 50% of 1RM with 1 min rest between sets. Hamstring muscle fatigue was then assessed using a Cybex II+ isokinetic dynamometer. The general hamstring fatigue task involved 10 maximal 40 m sprints with 30 s rest between each sprint. Three final 40 m sprints were performed by the subjects (fatigued condition, FC) during which time high speed film of the subjects' sprint action and EMG of the same five lower extremity muscles were again recorded.

RESULTS

Correlated *t*-test results indicated the following significant changes in the subjects' running action from the NFC to the FC: increased stride duration due to increased duration in the swing phase; a later occurrence of maximum knee extension in the swing phase; greater thigh extension and less knee flexion at the point of maximum knee extension in the swing phase; decreased leg angular velocity immediately prior to foot-ground contact; decreased angular displacement of the thigh segment from maximum thigh flexion to maximum knee extension in the swing phase; and decreased angular displacement of the trunk, thigh and leg segments from maximum knee extension in the swing phase to foot-ground contact. Statistical analysis of the EMG data indicated a significant increase in duration of ST activity in the FC compared to the NFC. Furthermore, the first RF burst turned off earlier, whereas the second RF burst turned on earlier in the FC compared to the NFC. Finally, hamstring muscle (ST and BF) activity began earlier relative to the onset of VL activity and a smaller time interval was evident from the cessation of RF activity to the onset of BF activity in the FC compared to the NFC.

DISCUSSION

The changes in the kinematic and EMG parameters of sprint running resulting from fatigue primarily appeared to serve as protective mechanisms to reduce stress placed on the hamstring muscles at critical phases of the stride cycle. These protective mechanisms, in turn, may have been elicited to compensate for the decreased efficiency of the fatigued hamstring muscles and may have decreased potential for hamstring muscle strain injury.

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MUSCLE DAMAGE AND MUSCLE STIFFNESS INDUCED BY ECCENTRIC EXERCISE

Mitsuyoshi Murayama¹⁾ Kazunori Nosaka²⁾ Tsugutake Yoneda³⁾

¹⁾ Institute of Physical Education Keio University, Yokohama, ²⁾ Exercise and Sports Science, Department of Environmental Science, Yokohama City University, Yokohama, ³⁾ Department of Physiology, School of Sports and Health Science, Juntendo University, Chiba, JAPAN

INTRODUCTION

Eccentric exercise induces muscle damage and muscle swelling. Previous studies have shown that increases of upper arm circumference occur after eccentric exercise in elbow flexions. Furthermore, it has been pointed out that values of joint angles of relaxing elbow after the same exercises decreased because of shortening of the elbow flexors. Muscle hardness as evaluated by means of non-invasive measurement, has been used to define muscle stiffness in human beings. It has been found that the muscle stiffness becomes greater after fatigue and contraction of muscles. The swelling of a muscle must increase its stress. But, the hardness of shortened and swelled elbow flexors following eccentric exercise has not been confirmed. The purpose of this study was to investigate the changes in muscle stiffness of the elbow flexors following eccentric exercise.

METHODS

Five male students performed 24 maximal eccentric actions of the elbow flexors of one of their arms at elbow joint angle from 100 to 180 degrees. The other arm did not perform any exercise and served as a control. Isometric force of the elbow joint angle at 90 degrees (IF), range of motion (ROM) which was determined by measuring the difference between relaxed elbow joint angle (RANG) and flexed elbow joint angle (FANG), muscle soreness (SOR), plasma creatine kinase activity (CK), circumference (CIR), and B-mode ultrasound pictures (US) of the elbow flexors were measured before and for 7 days after exercise. In each case, muscle stiffness was measured after these measurements were finished. The measurement of muscle stiffness was performed using the principle of the pushing method: the displacements of deformation of the tissue and responded forces from the tissue were recorded. Following the two-layer spring model (Horikawa *et al*), muscle stiffness was evaluated by using the data points in the range of reflecting muscle layer. These data points fit a quadratic curve rather than a straight line. Therefore, we decided to regard the responded force (N/cm^2) at the 50% muscle thickness which was calculated from the quadratic equation as the value of muscle stiffness in this study. Muscle stiffness of elbow flexors was measured at the arm when placed on a bed in a relaxed position. All measures except CK were taken from both exercise and control arms.

RESULTS

CK increased significantly ($P < 0.01$) and peaked 4 days after exercise ($16,213 \pm 4,990 \text{ IU/L}$). There were no significant changes in any measures and in muscle stiffness for control arms. For the exercised arm, all measures changed significantly ($P < 0.01$). SOR developed after exercise and IF dropped immediately to approximately 40% of pre-exercise level ($175.5 \pm 7.1 \text{ N}$) after exercise and recovered to 60% of pre-exercise level by 7 days after exercise. CIR increased from $24.7 \pm 0.8 \text{ cm}$ to $26.7 \pm 0.8 \text{ cm}$, and US showed enlargement of muscle thickness that was correlated with increase in CIR. ROM decreased after exercise and the amount of decrease in RANG was approximately 20 degrees at 4 days after exercise. Muscle stiffness values in relaxing elbow, decreased from $4.1 \pm 1.8 \text{ N}/\text{cm}^2$ (mean value of all subjects before exercise) to $3.4 \pm 1.1 \text{ N}/\text{cm}^2$ (3 days after), and then slightly increased to $4.9 \pm 2.4 \text{ N}/\text{cm}^2$ (7 days after) in exercised arm. Generally, muscle stiffness did not show a large increase in exercised arm after eccentric exercise.

DISCUSSION

It was previously thought that decreases in RANG were due to muscle shortening. However, in this experiment, the increments in muscle stiffness which were observed in muscle contraction and fatigue were not produced after eccentric exercise. According to Komiya *et al*, in non-exercised condition, muscle stiffness decreases with decreasing elbow joint angle during muscle relaxation. Therefore, muscle shortening might caused the decrements of RANG simply, without increasing muscle stiffness. If this is so, the increments of CIR and muscle thickness would be explained by water accumulation which localized outside of muscle fiber. However, it is possible to assume that this water accumulation has little effect on the remarkable increment of muscle stiffness, unlike in the case of muscle contraction and fatigue.

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IN VIVO FASCICLE LENGTH CHANGES OF THE HUMAN VASTUS LATERALIS MUSCLE DURING SQUATTING

Keitaro Kubo, Yasuo Kawakami and Tetsuo Fukunaga

Department of Life Sciences (Sports Sciences), The University of Tokyo, Japan

Introduction

The muscle length in relation to joint positions has been determined on human cadavers (3). It has been shown, however, that the compliance of the muscle-tendon unit causes internal shortening of fascicles even during isometric contractions (2). Thus prediction of muscle length change during joint movement based on cadaveric data (e.g. 1) is controversial. In this study changes in fascicle length during movement were determined in vivo in humans.

Method

Six healthy male subjects (age; 22.3 ± 1.5 yrs, height; 171.7 ± 2.9 cm, mass; 65.8 ± 5.9 kg) participated in the study. The subject bent knees bilaterally to 90° and immediately extended up to the upright position at a speed of 20 cycles/min. The knee joint angle was recorded with an electrogoniometer. B-mode ultrasonic apparatus (SSD2000) was used to obtain longitudinal images of vastus lateralis muscle at a site 50% of the right thigh and recorded every 33 msec on a videotape. The fascicle length was determined as the length of a line drawn along the echo parallel to the fascicles.

Results

The fascicle length was stretched when knee was flexed, and shortened when it was extended. Fig.1 shows relations between knee angles and fascicle length. At the knee position of $10-50^\circ$ of flexion, fascicle length in the flexion phase was shorter than that of the extension phase ($p < 0.05$).

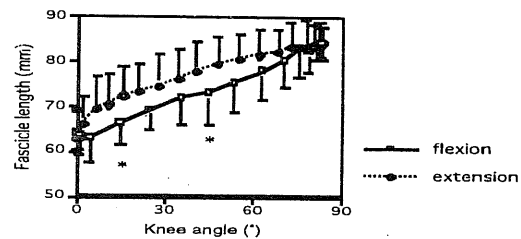


Fig.1 The relationships between knee angles and fascicle length * $p < 0.05$

Discussion

The muscle-tendon complex (MTC) is actively stretched (eccentric action) during flexion phase and then shorten (concentric action) during extension phase. The difference in the fascicle length between those two phases would be due to the series elastic component (SEC) in MTC. The SEC is stretched during eccentric phase, and recoil during concentric phase. The present study showed that in a human muscle in vivo the length of the contractile and elastic components changes dynamically when there is no apparent change in the MTC length. The effects of external loads imposed on MTC at different velocities are being investigated presently.

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FUNCTION OF THE GASTROCNEMIUS - TIBIALIS ANTERIOR MUSCLE COMPLEX IN RAPID KNEE FLEXION

Keigo OHYAMA BYUN and Morihiko OKADA

Inst. Health & Sport Sciences, Univ. of Tsukuba, Tsukuba, Ibaraki, JAPAN

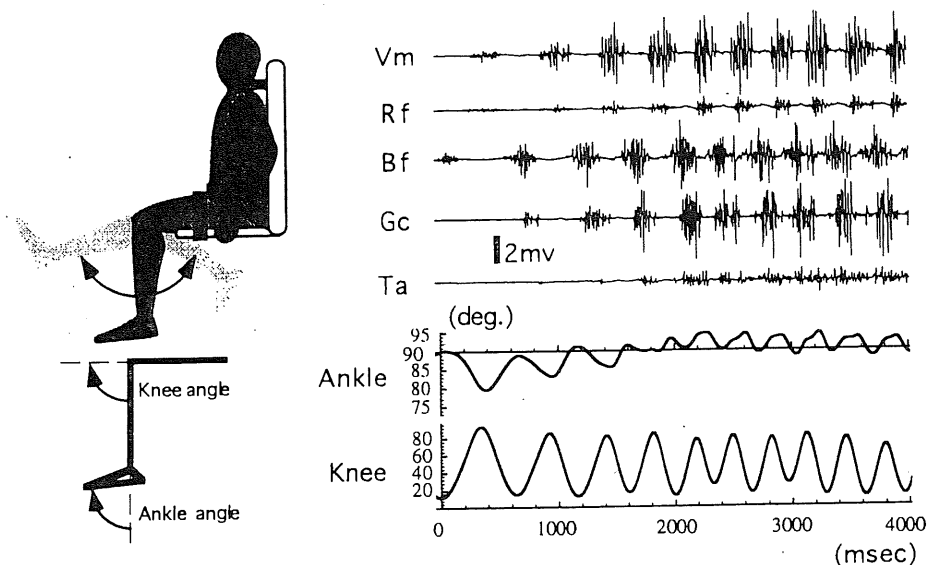
INTRODUCTION

While the gastrocnemius muscles (Gc) are powerful plantar flexors with their significant masses, due attentions have not been given to the possible functional role of these muscles in knee flexion. By virtue of it's biarticularity, Gc may function as a strong knee flexor with the aid of ankle dorsiflexors.

The purpose of this study was to investigate the functional association between Gc and m. tibialis anterior (Ta) in the flexion phase of rapid alternate extension - flexion of the knee.

METHODS

Healthy young male subjects conducted whole range alternate extension - flexion of the knee joint while sitting down on a chair, with the upper body and thighs fixed firmly. Beginning with about 2 cycles per second, the movement speed was gradually increased up to its maximum. Surface EMGs were recorded from six leg muscles, i. e. soleus, medial gastrocnemius, tibialis anterior, vastus medialis, rectus femoris, biceps femoris long head. Joint angles were measured with electrogoniometers for the knee and ankle joint.



RESULTS

Gc showed bursting activities corresponding to each bout of knee flexion. The bursts increased in magnitude proportionately with increase in the angular velocity which remained separated without fusing each other. Of the triceps surae muscles, activity in the soleus (So) was small or almost negligible compared to Gc. On the other hand, activity of Ta gradually increased toward the faster movements. As a result, the ankle joint posture gradually changed from plantar flexion to dorsiflexion.

DISCUSSION

The knee flexion moment of Gc seems to have been sustained by Ta contraction that prevent ankle plantar flexion in which the knee flexing tension may be dissipated. Moreover, Gc could utilize the passive tension raised in the stretch shortening cycle. Our results suggests that rapid knee flexions are effectively carried out by collaboration of Gc and Ta activity as a functional complex.

MUSCLE FIBER BEHAVIOR OF GASTROCNEMIUS DURING VERTICAL JUMPING

Sadao Kurokawa, Hongliang Yan, Hiroshi Nagareda, Kazuo Funato and Tetsuo Fukunaga
Laboratory of Sports Sciences, Department of Life Sciences, The University of Tokyo, Japan

INTRODUCTION

In many previous studies, the change in the length of muscle-tendon complex (MTC) during human movement have been estimated using joint angles and anthropometric data. However, because the MTC consists of contractile(muscle fiber) and elastic element(tendinous tissue), the elastic properties of tendinous tissue should be taken into consideration to estimate the change in the length of muscle fiber from MTC. The purpose of the present study was to determine the change in length of muscle fiber (fascicle) of gastrocnemius medialis (MG) *in vivo* during vertical jumping in human.

METHODS

Subjects were requested to perform two legged vertical jump with maximum effort on a force plate(KISTLER) without preparatory counter movement. A ultrasonic probe (7.5MHz) was attached on the muscle belly of MG by using specially designed apparatus, and longitudinal ultrasonic images of MG were stored in the cine-memory with a real time computerized ultrasonic apparatus(SSD-2000,ALOKA,JPN) set to operate at 40Hz. Fascicle length was determined as the distance between the ultrasonic echoes reflected from the insertions of the fascicle into the superficial and deep aponeuroses. The length of MTC of MG was calculated from the formula proposed by Grieve et al(1978). EMG activities were recorded from MG and soleus muscle by means of bipolar surface electrodes. Kinetic and kinematic data were obtained using electrogoniometers, force plate and high-speed video camera(NAC).

RESULTS AND DISCUSSION

As shown in Fig.1, the three different phases were observed in the changes of ground reaction force and fascicle length during the push-off phase in jumping(i.e. phase I is from A to B, phase II from B to C, and phase III from C to toe-off). During phase I, the MTC length remained at a constant value owing to little changes of joint angles in ankle and knee, while the fascicle length decreased by 27%. On the contrary, during phase II where vertical ground reaction force indicated maximum, the fascicle kept the constant length of 45 mm, whereas the MTC length decreased by 5%. Before toe-off(phase III), both lengths of fascicle and MTC decreased simultaneously. The decrease in length of MTC throughout phase II and III was attributed to the interaction of the joint angle changes in ankle and knee. MG muscle was activated continuously up to approximately 50msec before toe-off.

At the first half of push-off phase(phase I), the shortening of muscle fiber may be considered to be due to the lengthening of elastic element(tendinous tissue) and to take up slack in MTC. Following phase II, it is considered that the shortening of MTC length is caused by the release of elastic energy stored in tendinous tissue(recoil of tendinous tissue), while muscle fiber contracts isometrically.

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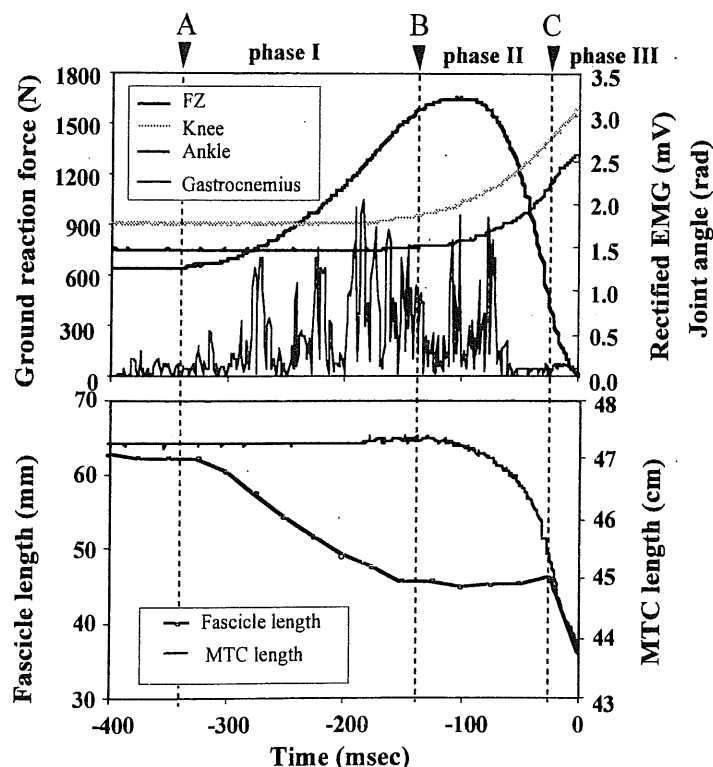


Fig.1 Typical time histories of vertical ground reaction force(FZ),knee and ankle joint angles, EMG from gastrocnemius medialis (top panel) and fascicle length, muscle-tendon complex(MTC) length from gastrocnemius medialis(bottom panel). Time is expressed relative to instant of toe-off($t=0$). A is defined as the onset of increase of vertical ground reaction force, B the onset of decrease of MTC length, C the onset of steep decrease of fascicle length.

RELATIONSHIPS BETWEEN ARCHITECTURAL CHARACTERISTICS OF KNEE EXTENSOR MUSCLES AND JUMPING POWER DURING SQUAT JUMPING

H. Akima, H. Shirakura, H. Takahashi, H. Okada, Y. Kubo, M. Ae, Y. Muraki, S. Katsuta
Department of Life Sciences, University of Tokyo, Tokyo, Japan

INTRODUCTION

Knee joints play an important role during jumps such as squat jumps. We can imagine easily that long and triple jumpers are able to produce higher power output than normal subjects during jumping. Although a large number of studies have been made on jumping, little is known about why jumpers are able to produce higher power output during jumping. In this study, we focused on the knee joints, and assumed that power output at knee joints evaluated by kinematic analysis was produced by the knee extensor muscles (KE). To clarify why jumpers can produce higher power output, we obtained architectural data of the KE. The purpose of the present study was to investigate relationships between jump power during squat jumping and architectural characteristics of the KE using magnetic resonance imaging (MRI).

MATERIALS AND METHODS

Eight male long and triple jumpers (JUMP) and five healthy men (CONT) participated in the present study. The subjects performed squat jumps and were videotaped (200 fields/sec) jumping on the force platform. Torques, angles, angular velocities, and power output were calculated for each lower limb joint based on a linked segment model. Torque and angular velocity at the knee joint were converted to tendon force (Ft) and tendon moving velocity (Vt), respectively, at the patellar tendon. Serial axial images were obtained from the anterior superior iliac spine to the head of fibula for each subjects' right leg using MRI. From the serial images, outlines of each KE were digitized, and the anatomical cross-sectional area (ACSA) and muscle volume of each muscle were calculated. Fiber length (FL) of each KE muscle was calculated as muscle length multiplied by the ratio of muscle length to FL as reported by Wickiewicz et al. (1983). Physiological cross-sectional area (PCSA=muscle volume/FL) of KE was determined.

RESULTS AND DISCUSSION

The PCSA of the KE in JUMP were significantly greater than in CONT, and mean FL of four individual muscles in JUMP were also significantly longer than in CONT (Table 1). These results suggest that the KE in JUMP would be able to exert higher muscle force, and to contract faster than in CONT. It has been reported that the maximum force exerted by the muscle, i.e. tendon force in this study, is closely related to the PCSA, and the maximum rate of shortening velocities are related to the muscle fiber length. Power output, Ft, and Vt at peak power, peak Vt at knee joint in JUMP were significantly higher than in CONT (Table 1). However, no differences were observed in peak Ft between JUMP and CONT. There was a possibility that greater values in Ft and Vt in JUMP were due to the differences in architectural characteristics, i.e., PCSA and FL, of the KE. These parameters, i.e., Ft and Vt, should be normalized by PCSA and FL, respectively.

Table 1 Physiological cross-sectional area (PCSA), fiber length (FL), power, tendon force (Ft), and tendon velocity (Vt)

	PCSA (cm ²)	FL (cm)	Peak power (W)	Peak Ft (N)	Peak Vt (cm/sec)	Ft at peak power (N)	Vt at peak power (cm/sec)
JUMP (n=8)	275.5** (19.7)	7.2* (0.3)	334.7** (180.8)	3369.7 (721.6)	46.6* (3.3)	1628.1* (456.0)	19.2* (6.9)
CONT (n=5)	220.4 (17.2)	6.6 (0.5)	93.9 (55.0)	2654.2 (654.1)	39.2 (4.6)	1088.3 (318.7)	9.4 (5.5)

Values are means and SD. *, p<0.05, **, p<0.01. Fiber length value was average of four muscles.

Table 2 shows normalized values of Ft and Vt by architectural properties. There were no significant differences in peak Ft/PCSA and peak Vt/FL between JUMP and CONT. There was no significant difference in Ft/PCSA at peak power between JUMP and CONT, while Vt/FL at peak power in JUMP was still significantly greater than in CONT. These differences would be associated with intrinsic speed of shortening of sarcomeres and training jumpers have done for a long time.

Table 2 Peak power, tendon force (Ft), and tendon velocity (Vt) at knee joint during squat jumping.

	Peak Ft/PCSA (N/cm ²)	Peak Vt/FL (1/cm)	Ft/PCSA at peak power (N/cm ²)	Vt/FL at peak power (1/cm)
JUMP (n=8)	12.3 (2.7)	6.5 (0.6)	5.9 (1.6)	2.6* (0.9)
CONT (n=5)	11.9 (2.1)	6.0 (0.7)	5.0 (1.5)	1.4 (0.9)

Values are means and SD. *, p<0.05, **, p<0.01.

It is concluded that JUMP have greater power output at the knee joint during jumping, and this is based on faster muscle contraction at peak power when normalized by the architectural characteristics of knee extensor muscles.

SEX DIFFERENCES IN LIMITATION OF FAT-FREE MASS AND CHARACTERISTICS OF SKELETAL MUSCLE ARCHITECTURE

T. Abe, W. F. Brechue*, S. Fujita†, and J. B. Brown*

* Dept. of Kinesiology, Indiana Univ., Bloomington, IN, Dept. of Exerc. Sport Sci., Tokyo Metropolitan Univ., Tokyo, Japan, and † Dept. of Exerc. Sci., Univ. of Southern California, Los Angeles, CA

INTRODUCTION

Skeletal muscle is the largest fat-free tissue component at the tissue-system level of human body composition. Therefore, fat-free body mass (FFM) is often used to estimate total skeletal muscle mass. Forbes [1] speculated that the upper limits of FFM are about 100 kg in men and 60 kg in women. Our previous study, however, indicated that male professional Japanese Sumo wrestlers whose FFM averaged 109 kg, including the largest one of 121.3 kg [2]. To our knowledge, the upper limit of FFM in female athletes has not been investigated. Furthermore, the association between sex differences in FFM limitation and characteristics of the skeletal muscle architecture has not been examined. Therefore, the purpose of this study was to investigate 1) the upper limit of FFM in female athletes, and 2) whether sex difference exist in architectural characteristics of hypertrophied muscle.

METHODS

Twenty-five female athletes (mean age 21.6 ± 5.2 yr and mean stature 1.73 ± 0.06 m) and 22 frame size-matched male athletes (mean age 19.6 ± 1.3 yr and mean stature 1.74 ± 0.08 m) volunteered for the study. Body density was measured by the hydrostatic weighing [2] to estimate FFM. Muscle thickness was measured by B-mode ultrasound at 13 sites of the body [3]. Isolated muscle thickness and fascicle angles of the long head of the triceps (T-long), vastus lateralis (VL) and gastrocnemius medialis (MG) muscles were also measured *in vivo* using B-mode ultrasound. Muscle fascicle lengths of T-long, VL and MG were estimated [4].

RESULTS

Four of the female athletes had more than 60 kg of FFM, including the largest one of 82.1 kg. On average, male athletes had a significantly greater FFM than female athletes (71.5 ± 10.5 kg vs 56.1 ± 8.9 kg). Muscle thickness was highly correlated with cross-sectional area (CSA) of the muscle and was significantly larger in male athletes at all sites except for the anterior thigh. Male athletes had a significantly higher isolated muscle thickness and fascicle angles of the T-long, VL and MG muscles than female athletes. A significant correlation was observed between muscle thickness and fascicle angle for T-long ($r=0.84$, $P<0.001$) and MG ($r=0.38$, $P<0.01$) muscles, but not for VL muscle. No significant difference was found between sexes in fascicle length of the T-long, VL and MG muscles.

	Muscle Thickness (cm)			Fascicle Angle (deg)			Fascicle Length (cm)		
	T-long	VL	GM	T-long	VL	GM	T-long	VL	GM
Female	2.3 ± 0.3	2.7 ± 0.3	2.1 ± 0.3	21 ± 3	18 ± 3	20 ± 2	6.3 ± 0.7	8.9 ± 2.0	6.2 ± 0.9
Male	$3.0 \pm 0.3^\dagger$	$2.9 \pm 0.3^\dagger$	$2.3 \pm 0.4^*$	$27 \pm 3^\dagger$	$21 \pm 3^\dagger$	$23 \pm 3^\dagger$	6.5 ± 0.6	8.6 ± 1.7	5.9 ± 1.0

[Values expressed as mean $\pm$ SD; $^\dagger P<0.01$, * $P<0.05$ versus female athletes]

DISCUSSION

Our data revealed the existence of the essential sex difference in FFM reflected by upper limit of skeletal muscle mass. Factors which affect upper limit of skeletal muscle mass include fiber size (CSAs of the type I and type II fiber), the amount of non-contractile tissue (mainly connective tissue), muscle fiber number, fascicle length and possible muscle fiber pennation angle. Previous studies [5,6] indicated that male athletes had significantly greater mean fiber CSAs than the female. However, no significant sex difference was found in biceps and VL muscle fiber numbers. Our data indicate that fascicle length was similar between sexes in the T-long, VL and MG, but male had a significantly greater fascicle pennation angles. These results suggest that the greater absolute skeletal muscle mass of males is due primarily to differences in fiber size. Moreover, the difference in fascicle angles between sexes may be associated with level of muscle fiber hypertrophy.

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SHORTENING VELOCITY OF MUSCLE FIBERS IS NOT CONSTANT DURING 'ISOKINETIC' JOINT ACTION

Yoshiho Ichinose, Masamitsu Ito, Yasuo Kawakami, and Tetsuo Fukunaga

Laboratory of Sports Sciences, Department of Life Sciences, University of Tokyo, Tokyo, Japan

INTRODUCTION

To determine force-velocity relationships of human muscles, many authors have utilized isokinetic dynamometers which employ constant angular velocities. However, because of anatomical factors, a constant angular velocity over the range of motion may not represent a uniform shortening velocity of muscle fibers. Especially, in pennate muscles like the quadriceps femoris, fascicle length change does not parallel that of joint angles (1). The present study aimed to determine the fascicle shortening velocities *in vivo* during voluntary 'isokinetic' knee extension.

METHODS

Six males participated in this study as subjects. The subject was seated on a test bench of an electric dynamometer (MYORETRZ-400, Asics) with the hip joint angle flexed at 80°. The subject extended the leg with the maximal effort from 110° flexion to full extension at 150°/s. Longitudinal images of the vastus lateralis muscle (VL) at a level 50% of the thigh length were obtained using an ultrasonic apparatus (SSD-2000, 7.5MHz, Aloka) and recorded on a video tape. Knee joint angles were synchronously recorded by an electrogoniometer. A fascicle was traced to measure its length on the ultrasound images which were printed out from the video tape every 30 ms, and fascicle shortening velocity was calculated.

RESULTS

During an 'isokinetic' action of knee extension the fascicle shortening velocity of VL varied with joint angle (Fig.1). Highest shortening velocity was observed at the knee joint angle of 50-70°. At the onset (flexed position) and toward the end of motion (extended position), the shortening velocity was approximately 60% lower than in the middle of the motion.

DISCUSSION

Fascicle length and pennation angles of VL have been shown to vary with knee joint angles (1). In addition, moment arms of the knee extensors change with knee joint angles (2). These anatomical factors would contribute to the nonuniform fascicle shortening velocity during 'isokinetic' joint movement. The results indicate that the assumption of a constant muscle shortening velocity in an 'isokinetic' testing is not satisfied. Many studies have used peak torque without considering joint angles to study force-velocity relationships of human muscles. However, in that case force-velocity characteristics of muscle fibers may not reasonably be reflected.

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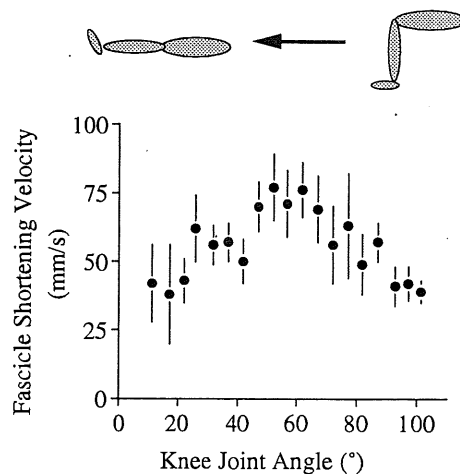


Fig. 1 Variation of fascicle shortening velocities due to changes in knee joint angles during 'isokinetic' knee extension

IN VIVO MOMENT ARM ESTIMATIONS OF HUMAN VASTUS LATERALIS DURING CONTRACTIONS

M. Ito, Y. Ichinose, Y. Kawakami and T. Fukunaga

Department of Life Sciences (Sports Sciences), University of Tokyo, Tokyo, Japan

INTRODUCTION

Moment arms have been measured with geometrical methods (2, 3), e.g. CT and MRI, in vivo. However the difficulty in determining muscle's line of pull during contractions does not allow us to straightforwardly apply moment arms obtained from images to biomechanical calculations. Moment arms obtained from other methods performed with cadavers such as tendon travel method (1, 3) cannot be thought to reflect faithfully those of living humans. In this study, ultrasonography was used to measure the distance of the tendinous movement of the human vastus lateralis muscle to estimate the moment arms under tensed conditions, together with the tendon travel method (1, 3).

METHOD

Real-time ultrasonography (SSD-2000, Aloka, Japan) was applied to measure the tendon travel distance of the human vastus lateralis muscle (VL) in vivo. The center of rotation of the knee joint was visually aligned with the axis of rotation of a dynamometer (Myoret RZ-450, asics, Japan). Subjects ($n=6$) performed isokinetic knee extension exercises with submaximal effort (30 - 40%MVC). The range of motion was 100 to 0° (0° = full extension). Ultrasonic images were recorded at 30Hz using video, with time codes for synchronization to angle and torque data. The images were printed on papers and identification of fascicles and aponeurosis were executed on them. A cross point (P) between the echoes from the aponeurosis and a fascicle was determined as an attaching point of the fascicle onto the aponeurosis. The traveled distance of P was measured to the nearest millimeter with a ruler. The traveled distance of P vs. knee joint angle data were fit with a polynomial that minimized the sum of the squared residuals. The moment arm of VL was calculated by differentiating the polynomial fit of the traveled distance of P vs. knee angle data.

RESULTS

The cross point (P) moved proximally by 59mm with decrease in knee flexion angle from 100 to 0° (Fig.1A). The estimated moment arms of the VL was largest at the knee flexion angle of 36° and the range was from 17 to 43mm in this subject.

DISCUSSION

When the moment arm curve calculated in this study is compared with the data from previous studies (Fig.1B), the values, range and knee angle at which

the peak of the moment arm curve appears are in good agreement with the previous studies (1, 2). In this study, the subjects tried to keep the constant level of force throughout the range of motion. However, since controlling the muscle's force constant throughout the range of motion in vivo is impossible, the effects of tendon elasticity may affect the traveled distance of P and thus the calculated moment arms. Although some problems will have to be solved yet, the method employed here can give us individual moment arms of many muscles during contractions in vivo.

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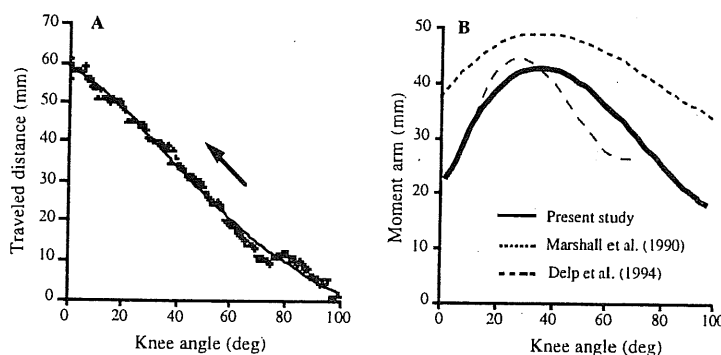


Fig.1 A: Traveled distance of P during knee extension from 100° flexion to full extension. B: Calculated moment arms vs. knee angle, comparing with the data from the previous studies.

SPECTRAL DECOMPOSITION OF MECHANOMYOGRAM (MMG) ASSOCIATED WITH LATERAL EXPANSION AND BULK MOVEMENT

Y. Ito ¹, K. Akataki ², M. Watakabe ², K. Mita ², K. Ando ³ and M. Ito ⁴

¹ Suzuka University of Medical Science and Technology, Suzuka, Japan

² Institute for Developmental Research, Aichi Human Service Center, Kasugai, Japan

³ Chubu University, Kasugai, Japan

⁴ Bio-Mimetic Control Research Center, Riken, Nagoya, Japan

INTRODUCTION

Mechanomyogram (MMG) has been considered to be originated from pressure waves due to lateral expansion of contracting skeletal muscle fibers and to reflect the mechanical activities of muscle fibers (Gordon and Holbourn 1948). However, the MMG derived by a microphone or a contact acceleration sensor seems to include two components associated with the lateral expansion of individual muscle fibers and the longitudinal bulk movement of the muscle related to the nonsimultaneous activation of the fibers (Orizio et al. 1990). The aim of the present investigation was to develop a spectral decomposition method of the MMG into two components associated with the lateral expansion and the bulk movement in contracting muscles.

METHODS

Ten healthy male adults ranging in age from 21 to 24 years participated in this study. Each subject was required to hold different levels of isometric contractions from 10 to 90% MVC of right knee extensors. The MMG was detected by a contact acceleration sensor (MT-3T, Nihon Koden) attached at the surface of the quadriceps muscle. The acceleration wave which was considered to reflect only the bulk movement was recorded by the same acceleration sensor placed on the tuberositas tibiae. The power spectral densities of the MMG ($P_{MM}(f)$) and the acceleration over the tuberositas tibiae ($P_{AA}(f)$) and the coherence between two signals ($Coh^2(f)$) were estimated by the FFT method. The lateral expansion component ($P_L(f)$) and bulk movement component ($P_B(f)$) of the MMG power spectral density were decomposed according to the following equations;

$$P_L(f) = P_{MM}(f) \cdot (1 - Coh^2(f)) \quad (1)$$

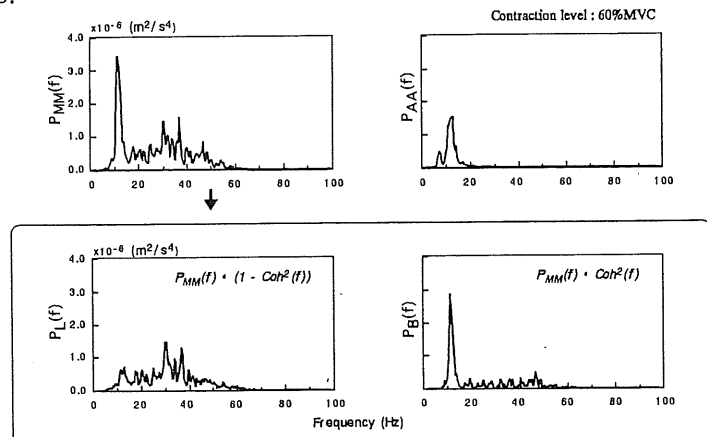
$$P_B(f) = P_{MM}(f) \cdot Coh^2(f) \quad (2)$$

RESULTS & DISCUSSION

The $P_{MM}(f)$ changed from unimodal to multimodal distribution as the muscle contraction increased. The first peak of the multimodal $P_{MM}(f)$ at strong contraction levels occurred in the bandwidth from 5 to 15 Hz as well as the unimodal $P_{MM}(f)$ peak at weak contraction levels (Barry 1987). This bandwidth was similar to that of the $P_{AA}(f)$ peak frequency. The maximal coherence between two signals in the range of 5 to 15 Hz varied from approximately 0.3 to 0.9 with the increasing level of force. Figure represents typical patterns of $P_{MM}(f)$ and $P_{AA}(f)$ obtained from experimental recordings at 60% MVC and $P_L(f)$ and $P_B(f)$ calculated according to equations (1) and (2). The results of the spectral decomposition showed progressive decrease in percentage of the lateral expansion component ($P_L(f)/P_{MM}(f)$). The ratio was $87.0 \pm 2.9\%$ at 10% MVC and reduced to $59.5 \pm 12.7\%$ at 80% MVC and $62.3 \pm 12.0\%$ at 90% MVC. The present decomposition method may be useful to estimate accurately the mechanical activities of muscle fibers from the MMG.

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NON LINEAR BEHAVIOUR OF THE MUSCULAR EQUIVALENT VISCOSITY FACTOR AT DIFFERENT MUSCLE LENGTHS.

A. Martin, B. Morlon, M. Pousson and J. Van Hoecke.

Groupe Analyse du Mouvement, Université de Bourgogne, Dijon, France.

INTRODUCTION

In a previous study (Martin et al. 1994), it has been shown that the mechanical behaviour of the muscle during maximal concentric actions can be described using a muscle model. This model consists of a contractile component (CC) in parallel with a nonlinear viscous element (B), both in series with an elastic component (SEC). Relations derived from the model associated with experimental data permitted the determination of the maximal shortening velocity of the muscle ($V_0(x)$) and of the a_f parameter which reflects the curvature of the force-velocity relationship. The aim of this study was therefore to examine the effects of the muscle length on the B factor and on the a_f parameters for the elbow flexor muscles.

METHODS

Modelling : The force that a muscle can develop during a concentric action at a given muscle length (x) is derived from the muscle model. In the relation (1), $F_0(x)$ represents the maximal isometric force of the muscle, $F_C(x,v)$ the concentric force, $B_C(x,v)$ the damping factor which reflects the overall loss of muscle capacity to produce force at high shortening velocities, and V_{CC} the shortening velocity of the CC. From the relation (1) and by using the classical Hill equation of the force-velocity relationship, the theoretical expression of the nonlinear damping factor $B_C(x,v)$ can be derived (2) :

$$F_C(x,v) = F_0(x) - B_C(x,v)V_{CC} \quad (1); \quad B_C(x,v) = \frac{F_0(x)}{V_0(x)} \frac{(1+a_f)}{(a_f+V^*)} \quad (2); \quad F = \frac{(F_0 - F)b}{V_{CC}} - a = Bb - a \quad (3)$$

In equation (2), a_f is the dimensionless parameter which describes the curvature of the force-velocity relation. ($a_f = a/F_0 = b/V_0$) and V^* is the normalised velocity ($V^* = V/V_0$). From the equation (1) and the Hill equation, a linear relation between the force and the B factor can be derived (3). In this relation, a and b are the constants of force and of velocity. The experimental data $F(v)$ will be used in the relation (3) to determine the maximal shortening velocity of the muscle $V_0(x)$. The a_f parameter is determined by fitting the experimental data of the B factor with the theoretical values derived from the relation (2).

Experimental procedure : Biomechanical tests were carried out using a Biodex type isokinetic dynamometer. The subject was asked to perform 2 maximal concentric elbow flexions at different angular velocities (60 to 360 °s⁻¹), and to hold 2 maximal isometric actions at elbow flexion angles of 30°, 60°, 90° and 120° (0° corresponds to full elbow extension). The isokinetic torque was registered at angles of 30°, 60°, 90° and 120°. To determine the force and the shortening velocities from torque and angular velocity, the musculo-skeletal model of the elbow joint was used. By using the experimental $F(v)$ data and the equation (1), the B factor was calculated.

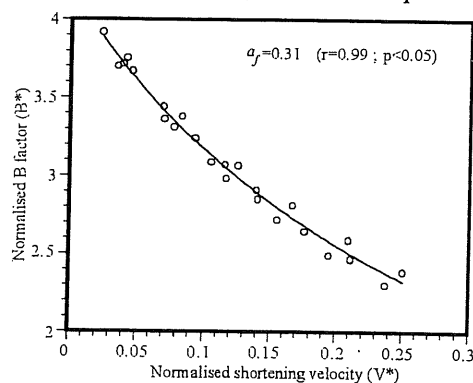


Fig. 1 : Determination of a_f by fitting the theoretical B factor with the experimental data.

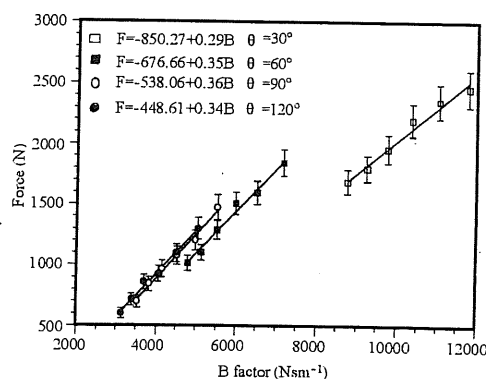


Fig.2 : Linear relationship between the force (F) and the experimental B factor ($0.96 < r < 0.98$; $p < 0.05$)

RESULTS AND DISCUSSION

At any muscle length the experimental F and B data are in agreement with the theoretical relation (3) (fig. 2), and at a given force F, the value of B increases with the muscle length. The maximum shortening velocities decrease with an increase of the muscle length (from 1.17ms⁻¹ at $l=0.285$ m to 0.93ms⁻¹ at $l=0.355$ m). Regardless of muscle length the parameter a_f was the same. For the elbow flexor muscles the value $a_f = 0.31$ has been found by fitting the experimental data of B with the theoretical nonlinear variations $B(v)$ (fig. 1, relation 2). It follows that the muscle model can be used for any simulation process, providing the dependences of B with muscle length and shortening velocity are taken into account and that the correct value of a_f is introduced in the model.

HUMAN MUSCLE STIFFNESS: *IN-VIVO* MEASUREMENT USING CONVENTIONAL DYNAMOMETRY

Rachel Skoss¹, Geoffrey R. Strauss², Peter J. Hamer¹, Graeme A. Wood¹ and Barry J. Gibson³

Department of Human Movement, The University of Western Australia¹, School of Physiotherapy, Curtin University of Technology² and Department of Human Movement, Edith Cowan University³, Perth, AUSTRALIA

INTRODUCTION

Stiffness measures of the series elastic structures of human muscle are functional correlates of muscle efficiency and effectiveness (Shorten, 1987), and as well can provide useful insights into neuromuscular control and injury mechanisms. Several techniques are described in the literature for *in-vivo* measurement of this muscle property, including the quick-release method (Goubel *et al*, 1971), added compliance (Wilkie, 1950) and damped oscillation (Cavagna, 1970) techniques. Each, however, has its own shortcoming (*viz*: untestable assumptions; specialised apparatus; band-limited assessment range). This paper describes a procedure which uses a conventional strength assessment dynamometer (Kin-Com, Chattecx Corporation) to obtain measures of hamstring muscle stiffness throughout the full range of knee flexor torque capability. The reliability of measures so obtained are reported, together with recommendations for future research.

METHOD

Maximal knee flexor isometric torque was assessed in 12 moderately active male subjects. Nine submaximal intensities (from 10 to 90%, with 10% increments) were then established. An eccentric/concentric isokinetic protocol on the Kin-Com was formulated to perturb the leg in an eccentric muscle action once a specified knee flexor torque (the start torque) was achieved by the subject. The leg was perturbed through a 10 degree range of movement with a high initial acceleration and a pre-set angular velocity of 180 degrees per second. Three trials at each relative intensity were performed, with the order of intensities presented randomly. Subjects were made aware of the trial intensity, and had 30 seconds recovery between trials. Following the trials at the submaximal intensities, three trials at 100% of the pre-test maximal isometric torque were attempted. If these efforts resulted in perturbation of the limb, further trials were performed with 10% incremental loads. Subjects were tested on three occasions, one week apart.

Measures of muscle stiffness were obtained from concomitant changes in joint torque and lower limb angular displacement during the iso-acceleration phase of the perturbation. Measures of joint torque were adjusted for gravitational and inertial factors, but no allowance was made for viscous damping which has minimal affect on the results in any event.

RESULTS

Individual subject:day stiffness-torque data sets were mathematically approximated using a variety of functions (polynomials, exponentials, logarithms) in order to obtain the best overall expression for the underlying nature of stiffness measures (program KURV+). The best least squares fit was found to be provided by a function of the form: $Y = aX^{bx}$, where Y is the stiffness (Nm/degree) and X is the isometric knee flexor torque (N.m), and the mean value of the coefficients were $a=32.97$ and $b=-13.19$. The reliability of function values (coefficients a and b) were then examined using ANOVA techniques. It was found that there was no significant difference ($p<0.05$) between testing days, and that the reliability was acceptable (intra-class $R_{xx} = 0.6$). Reliability estimates were influenced by the relatively small between-subjects variance, particularly in the case of coefficient b , and were higher for days 2 and 3 alone.

DISCUSSION

Measures of muscle stiffness provide essential data for musculo-skeletal modelling, and could also contribute to understandings of physiological change arising from exercise factors such as warm-up, fatigue, muscle conditioning and delayed-onset muscle soreness. The Kin-Com testing protocol implemented in this study provided a means of obtaining reliable measures for future examination of these effects. The protocol is particularly useful in that it will yield reliable measures at high torque levels, and that measures can be obtained under known loading circumstances which enables the stiffness-load behaviour to be accurately characterised. The best mathematical function was of an expected form, but was nevertheless superior to the traditional exponential formulae given to stiffness-torque curves (Shorten, 1987). The greater inter-subject variability observed for coefficient a suggests that the initial part of the curvi-linear relationship could reflect important morphological and physiological differences within the group. These too warrant future investigation.

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STIFFNESS CHANGES DISPROPORTIONAL TO MUSCLE FORCE AFTER CONCENTRIC CONTRACTIONS OF RAT MEDIAL GASTROCNEMIUS MUSCLE

K. Meijer*, H.J. Grootenboer*, H.F.J.M. Koopman* and P.A. Huijing*^s

*Institute for Biomedical Technology, University of Twente, Enschede, The Netherlands

^sInstitute for Fundamental and Clinical Movement Sciences, Vrije Universiteit, Amsterdam, The Netherlands

INTRODUCTION

Muscle shortening induces considerable changes in the muscle length-force characteristics (Meijer et al., 1997). These changes have been attributed to a decrease in the number of attached cross-bridges (due to fatigue or deactivation) as well as to distributions of sarcomere lengths in series. If these mechanisms are indeed active, then post-shortening force decrease is expected to be accompanied by a decrease in muscle stiffness. The purpose of the present study was to determine if there is such a coexistence.

METHODS

Experiments were performed on the medial gastrocnemius muscle (GM) of the rat ($n=2$). Pairs of fully isometric contractions (supramaximal stimulation) and isokinetic contractions ($v = 10 \text{ mm/s}$, $\Delta l = 4 \text{ mm}$) were performed at muscle lengths in the range between active slack length and optimum muscle length. The muscle was allowed to redevelop isometric force after the concentric phase of the isokinetic contractions. For each pair of contractions the target length of the concentric contraction was equal to the length at which the fully isometric contraction was performed. Muscle stiffness was determined using quick releases (0.2 mm in 3 ms), performed at the end of the fully isometric contractions and during the post-shortening isometric force plateau of the isokinetic contractions.

RESULTS

Post-shortening muscle force as well as stiffness were decreased with respect to their fully isometric values. This is in accordance with previous observations (Sugi & Tsuchiya, 1988). For a given decrease in force there was a larger decrease in stiffness (Fig. 1a). Comparison of the fully isometric and post-shortening force-stiffness curves indicated that the muscle had become more compliant after shortening (Fig. 1b). Analysis of the stiffness data with the alpha method (according to Morgan, 1977) indicated that the increased compliance was force dependent.

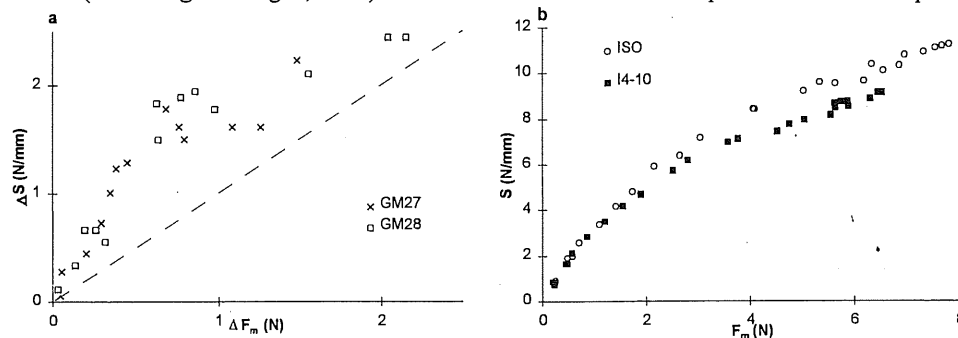


Figure 1 a) relationship between post-shortening force decrease (ΔF_m) and stiffness decrease (ΔS) for two muscles. b) force-stiffness curves for the fully isometric condition (ISO) and the post-shortening condition (I4-10).

DISCUSSION

Our results indicate that muscle shortening affects stiffness more than that it affects muscle force. The reduced stiffness may be a sign of hysteresis in the length-force properties of the extra-cellular tendinous tissue (i.e. tendon or aponeurosis). This could explain why, in rat GM, post-shortening force decrease is not accompanied by a change in muscle geometry (Meijer et al., 1997). Possible intra-cellular sources of reduced muscle stiffness could be a reduction in the number of attached cross-bridges or the presence of "popped" sarcomeres (Morgan, 1990). Potentially, the alpha method can be used to distinguish between cross-bridge elasticity and non cross-bridge elasticity. If the assumptions of the alpha method are valid, then the observed increase in force dependent compliance could be attributed to changes in tendinous tissue elasticity or the presence of "popped" sarcomeres. However, due to the non-linearity of the various elastic components the correctness of such a morphological distinction is questionable (Ettema & Huijing, 1994). It is concluded that post-shortening force decrease is accompanied by a disproportionate decrease in the series elastic stiffness of the muscle. Unfortunately, we cannot draw any definite conclusions regarding its source.

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OXYGEN KINETICS IN TWO DIFFERENT THIGH MUSCLES DURING DYNAMIC KNEE-EXTENSION EXERCISE

Toshiyuki Homma, Sachiko Homma, and Atsuko Kagaya

Research Institute of Physical Fitness, Japan Women's College of Physical Education.

Tokyo, Japan.

PURPOSE

The oxygen kinetics in individual muscles exercising simultaneously during regional exercise remains unclear. This study was therefore conducted to clarify the time course of changes in oxygen consumptions in two different thigh muscles recruited for dynamic knee-extension exercise and its relationship to pulmonary oxygen uptake.

METHODS

Six female subjects performed dynamic knee-extension exercise at 40% and 80% $\dot{V}O_{2peak}$, which was continued for 10 minutes (40% $\dot{V}O_{2peak}$) or until exhaustion (80% $\dot{V}O_{2peak}$). Pulmonary oxygen uptake ($\dot{V}O_2$) was measured by the 10-s mixing chamber method. Changes in oxygenated hemoglobin (HbO_2), deoxygenated hemoglobin (Hb) and total hemoglobin (HbT) contents were measured in the vastus lateralis (VL) and rectus femoris (RF) muscles using near-infrared spectroscopy (OM100A, Shimadzu) and oxygen saturation (SO_2) was calculated as HbO_2/HbT . The oxygen supply and oxygen consumption in the VL were calculated from the rates of increase in HbT and Hb under venous occlusion. The surface electromyogram (EMG) of the VL and RF was also recorded.

RESULTS AND DISCUSSION

Muscle oxygen supply, oxygen consumption and $\dot{V}O_2$ did not change after the initial increase during exercise at 40% $\dot{V}O_{2peak}$. The integrated EMG (iEMG) of the VL and RF was unchanged during exercise at 40% $\dot{V}O_{2peak}$, whereas it gradually increased with time in both muscles during exhaustive exercise at 80% $\dot{V}O_{2peak}$ (Fig. 1.). SO_2 in the VL continued to decrease until the subjects were exhausted. In contrast, it was unchanged in the RF after the initial decline (Fig. 2.). Oxygen supply as a ratio relative to the resting value in the VL reached 3.62 ± 0.72 (A.U.) in the first few minutes, but did not increase further as exercise at 80% $\dot{V}O_{2peak}$ was prolonged. However, the oxygen consumption and $\dot{V}O_2$ increased gradually from 20% of maximal duration (9.28 ± 0.72 A.U. and 324.1 ± 18.9 ml/min) until exhaustion (13.77 ± 1.15 A.U. and 562.3 ± 22.8 ml/min, respectively, $p < 0.01$; Fig. 3.), suggesting that extraction of oxygen from blood to muscle increased as the exercise at 80% $\dot{V}O_{2peak}$ was prolonged. The oxygen consumption in the VL was significantly correlated with $\dot{V}O_2$ during exercise ($r = 0.893 \sim 0.978$, $p < 0.001$). Therefore the present findings suggest that the oxygen kinetics are different in two thigh muscles and pulmonary oxygen kinetics are more closely related to muscle oxygen kinetics in the VL than in the RF during dynamic knee-extension exercise.

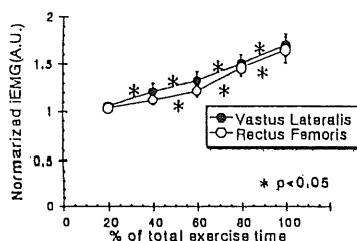


Fig. 1. Changes (mean $\pm$ SE) in iEMG as percentage of total exercise time during exercise at 80% $\dot{V}O_{2peak}$.

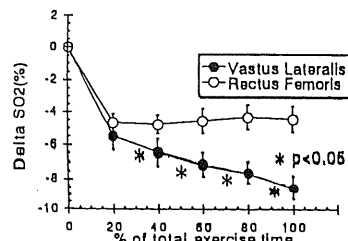


Fig. 2. Changes (mean $\pm$ SE) in SO_2 as percentage of total exercise time during exercise at 80% $\dot{V}O_{2peak}$.

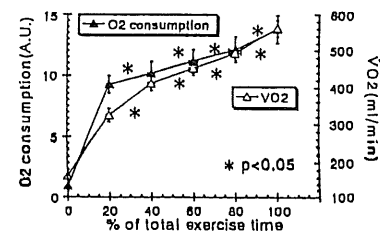


Fig. 3. Changes (mean $\pm$ SE) in oxygen consumption in the vastus lateralis muscle and $\dot{V}O_2$ as percentage of total exercise time during exercise at 80% $\dot{V}O_{2peak}$.

THE ILIOTIBIAL TRACT AND ASSOCIATED MUSCLES

J. Gavin Reid** and J. Mark Johnson*

*School of Physical and Health Education and *Department of Anatomy and Cell Biology
Queen's University, Kingston, Ontario, K7L 3N6, Canada.

INTRODUCTION

The iliotibial tract described as 'the ligament of idleness' and 'the ligament of boredom' by Evans (1979) is located between the iliac crest and the lateral tubercle of the tibia. It has two muscles inserting into it, tensor fascia lata (TFL) and gluteus maximus (GM), giving it the possibility to act over both the hip and knee joints. The functions of the iliotibial tract and its associated musculature are not well defined (Paré et al., 1981) and the function of the GM on the iliotibial tract has not been well documented. The purpose of this study was to measure the electromyographic (EMG) activity of the TFL and GM during a variety of movements, while associated force production at the hip and knee was evaluated.

METHODS

Ten male subjects performed knee extension, hip flexion (with the knee extended and flexed), hip abduction and hip extension on a KIN/COM while recording EMG.

Reliability of the exercise tests was accomplished by a second test session on five of the tested subjects. Results were reliable on total work, peak torque, and EMG measurements of GM. However, results were less reliable on angle of peak torque and EMG measurement of TFL (ANOVA showed no significant difference, but correlation $r < r_{crit}$ for both).

Peak force, the angle that peak force occurred, and work was retrieved from the KIN/COM data for both concentric and eccentric phases of the exercises. Peak force data were converted to moment data by multiplying force by the distance of the strain gauge pad from the centre of rotation (moment arm length). The resulting data were averaged for the two trials performed by each subject and evaluated using one-way repeated measures ANOVA.

RESULTS AND DISCUSSION

Concentric data were consistently lower than eccentric data for total work and peak torque, showing significant differences between the two contraction types ($p < 0.05$) for all exercises. Both TFL and GM were found to be active for all exercises measured. During measurements involving the knee extension both muscles were active, indicating the iliotibial tract had some affect on knee extension. The activity of both muscles was higher in hip flexion with the knee straight, as compared to having the knee bent, further strengthening that observation. Both muscles were shown to aid in hip abduction, especially at hip angles of greater than 15° abduction. During hip extension, GM reached its peak activity at 10° of hyper-extension, and TFL increased activity steadily as the hip moved to that hyper-extended position. As the hip joint moves from a flexed to a hyper-extended position, GM ability to laterally rotate the thigh increases.

When considering the coactivation of TFL and GM in all exercises, it would seem that the two muscles could act to reduce the unwanted thigh rotation in cases where one muscle is acting to produce an extension or flexion action without rotation of the thigh.

CONCLUSION

The evidence presented shows the wide range of functions of the iliotibial tract and the two affiliated muscles, TFL and GM. The training of this mechanism by strengthening the muscles could be an important technique for the correction or rehabilitation of knee injuries, due to its knee stabilizing qualities or possibly to help to prevent hip fractures in the elderly, by its femur-splint function.

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EFFECTS OF SLANTED CONTACT SURFACE ON NEUROMUSCULAR CONTROL OF THE HUMAN TRICEPS SURAE MUSCLE IN STRETCH - SHORTENING CYCLE EXERCISES

S. Yoon, K. Ohyama Byun, H. Okada, N. Fujii, K. Takamatsu
Institute of Health and Sport Sciences, University of Tsukuba
Tsukuba, Japan

INTRODUCTION

In order to execute an effective stretch-shortening cycle (SSC) exercise, it is important that muscle stiffness must be maintained throughout the cycle (Gollhofer et al. 1992). Muscle stiffness can be adjusted actively by neuromuscular activation such as stretch reflex (Nichols and Houk, 1976). Therefore, for increasing muscle stiffness during SSC exercises, it is necessary that the stretch reflex system be activated by enhancing the stretching length and stretching velocity of activated muscles during the eccentric phase.

The purpose of the present study was to investigate the neuromuscular control of the human triceps surae muscle during rebound drop jump which is as a typical SSC exercise, under slanted contact surface conditions.

METHODS

Subjects. Twelve male students (28.3 ± 3.5 yrs)

Experimental protocol. Rebound drop jump from a height of 30cm (RDJ30) and 50cm (RDJ50) under five surface gradient conditions (Uphill type 3° and 8° , Flat type, Downhill type 3° and 8°)

Measurements. During RDJ, the subjects were filmed with a 16mm high speed camera. Simultaneously, ground reaction force was recorded using a force platform. Electromyographic signals from the triceps surae muscle (medial head of gastrocnemius and soleus), m. tibialis anterior and m. vastus lateralis were registered with bipolar surface electrodes. From ankle and knee angle recording, the length change of m. gastrocnemius during contact phase was calculated according to the model of Grieve et al. (1978). The stretching length and stretching velocity of m. gastrocnemius during eccentric phase were obtained on the basis of this data.

RESULTS AND DISCUSSION

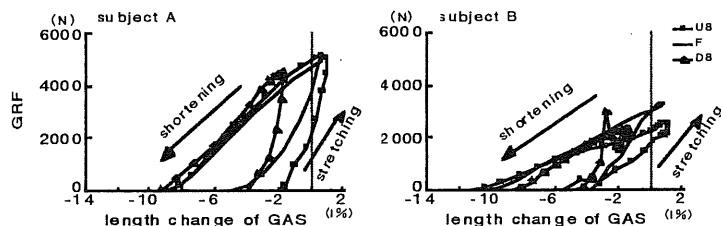


Fig. 1 Relationships of maximum ground reaction force and length change of the medial head of m. gastrocnemius during contact phase in RDJ30 - in case of Subject A and B -

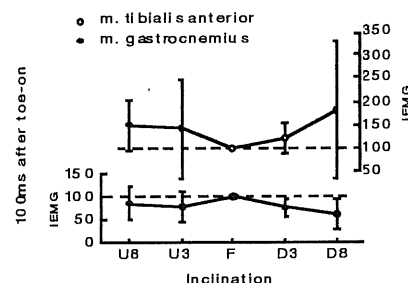


Fig. 2 The iEMG of the medial head of m. gastrocnemius and m. tibialis anterior during 100ms after toe-on in RDJ30

The results of this study of the rebound drop jump performed under slanted contact surface conditions show that jumping performance (jumping height and contact time) and ground reaction forces decreased compared to flat conditions in most subjects. The iEMG of m. gastrocnemius during 100ms after toe-on was lower than that during 100ms before toe-on under slanted contact surface conditions. These results may suggest the possibility of reduction of muscle stiffness and elastic energy utilization because stretch reflex can not be activated sufficiently under slanted contact surface conditions. Furthermore, verifying this possibility, mean stretching velocity and rate of ground reaction force to muscle length change of m. gastrocnemius during eccentric phase were examined in the present study. These parameters under slanted contact surface conditions were lower than those under flat conditions. From these results, RDJ under slanted contact surface conditions causes a reduction of muscle stiffness and elastic energy utilization because stretch reflex may be inhibited in order to protect the tendomuscular system and to maintain jumping balance.

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NEUROMUSCULAR FATIGUE OF INDIVIDUAL MUSCLES DURING WALKING IN HEALTHY YOUNG WOMEN

M. SHIBATA¹, Y. OHMURA¹, T. TAKAISHI², Y. YOSHITAKE³, and T. MORITANI³

¹College of Nursing Art and Science Hyogo, Akashi; ²Nagoya City University, Nagoya; ³Kyoto University, Kyoto, Japan

INTRODUCTION

During walking exercise, many different muscle groups of lower limb are active. During walking at high speed, we are not sure either neuromuscular fatigue occurs in all the active muscle groups at the same time or with a time-lag. Therefore, the purpose of this study was to investigate each individual muscle fatigue among different muscle groups during walking by means of electromyography (EMG).

METHODS

Ten young female college students (age: 21.0 ± 1.1 yrs, stature: 1.6 ± 0.07 m, body mass: 52.2 ± 5.1 kg) served as subjects for this study. Each subject performed a walking exercise on the treadmill at $110 \text{ m} \cdot \text{min}^{-1}$ for 2 minutes. Myoelectric (EMG) signals recorded by bi-polar surface electrodes from vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA), medial gastrocnemius (MG), and soleus (SL) muscles were band-pass filtered (5-1000 Hz) and amplified. From these data, mean power frequency (MPF) and root mean square (RMS) were calculated each period of 12 sec. According to the previous studies of neuromuscular fatigue, the neuromuscular fatigue in each individual muscle group was determined by both the significant decrease in MPF and the significant increase in RMS for time course.

RESULTS & DISCUSSION

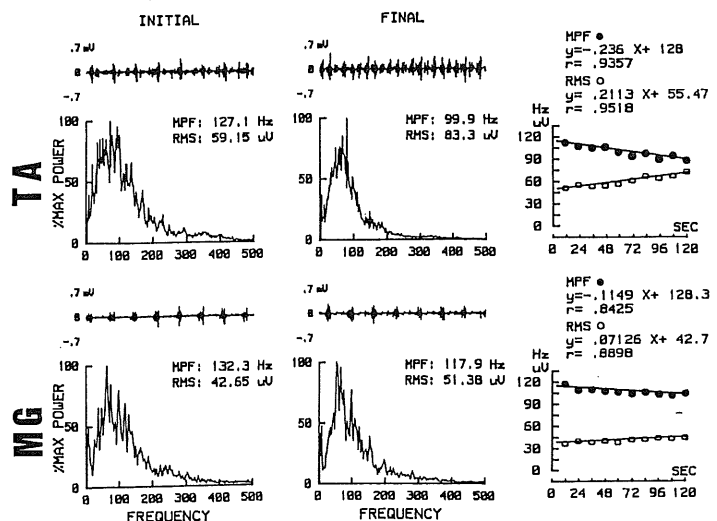
Figure 1 shows typical power spectral data and the change in MPF and RMS both in TA and MG during walking at a speed of $110 \text{ m} \cdot \text{min}^{-1}$. The spectral data in TA clearly indicate a decline in frequency toward the lower frequency band in the time course of muscle fatigue development. The RMS in BF was clearly increasing and MPF decreasing. On the other hand, there was a little change in the spectral data of MG, compared with those in TA. The absolute value of slope coefficient in RMS and MPF was also lower than that in TA. The grand mean in the rate of increase in RMS was greater in order of TA(.14), BF(.03), VL(.01), RF(.01), and others(minus). The grand mean value of slope coefficient of RMS in TA was significantly ($F_{5,54}=3.607$, $p<.01$) greater than that in other muscles. The grand mean in the rate of decrease in MPF was greater in order of BF(-.051), TA(-.037), RF(-.032), VL(-.031), SL(-.029) and MG(-.026), respectively.

These results would indicate that muscle fatigue occurs in each individual muscle group at different time period of walking exercise, not at the same time. TA and BF muscles tend to fatigue earlier at high speed. Our data also suggest that MPF and RMS are suitable indicator for evaluating individual muscle fatigue in total body exercise with many muscles being active.

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Figure 1 A typical set of data showing the changes in the raw EMG, spectral data in TA and MG muscles at initial and final periods of constant walking exercise for 2 min, and the changes in MPF and RMS obtained from EMG analysis for time course.



ADAPTIVE CHANGES IN RESPONSES TO REPEATED LOCOMOTOR PERTURBATIONS

M. K. Rand, D. A. Wunderlich, P. E. Martin, G. E. Stelmach, J. R. Bloedel
Arizona State University; Barrow Neurological Institute

Tempe; Phoenix, USA

INTRODUCTION

Compensatory changes of gait pattern have traditionally been examined by using unexpected, brief and relatively small perturbations (Nashner 1980; Dietz et al. 1987). When such a perturbation was applied during treadmill walking, gait was modified in a functional way to preserve stability. Both supraspinal and peripheral neural mechanisms are involved in generating such necessary modifications (Quintern et al. 1985). While most research in this area has focused on brief perturbations, effects of relatively long duration perturbations have not been well studied. In the case of brief perturbations, subjects may not have enough time to modify their gait during the time course of the perturbation, thus modification during the perturbation may be minimal. On the other hand, in the case of longer perturbations subjects may have enough time to modify their gait during the perturbation, thus subjects may develop a different strategy compared to that seen in a short perturbation. The present study examined responses to repeated treadmill perturbations of a relatively long duration in order to determine the adaptive mechanisms to stimuli of this type.

METHODS

Seven normal subjects (mean age: 42.5 years old; range 20-56) participated. During treadmill walking at 2.3 miles/hour, a perturbation consisting of a sudden slowing of the treadmill (S-Dec) followed by a sudden increase (S-Inc) back to the original speed was applied randomly. The duration from S-Dec to S-Inc was about 1500 ms. The perturbation was repeatedly applied at a specific time of the step cycle (at two thirds of the stance phase of the right leg). Thirty of these timed perturbations were applied in each of two trial blocks separated by a third trial block in which randomly timed perturbations were applied. To record body position using an Optotrak 3D system during gait, infrared light emitting diodes were placed over the big toe, fifth metatarsal, heel, ankle, knee, thigh, hip, and shoulder, neck, and upper back. EMGs were recorded from the tibialis anterior (TA) and gastrocnemius (G) of both legs. Pressure sensitive devices were placed under the big toe and heel for each foot as foot contact switches in order to record the time of heel-strike and take-off during gait. Responses in the two sessions with timed perturbations were analyzed. Angular motion of the trunk (medial-lateral and anterior-posterior), step length, EMG latency and EMG duration were measured.

RESULTS

In the early trials, subjects either shortened or lengthened their step length just after the treadmill S-Inc in order to compensate for the perturbations. They gradually adjusted step length so that it was relatively unaffected by later perturbations. The subjects also gradually reduced their trunk motion, which was large in the initial trials. Large variabilities in EMG patterns for lower limb muscles were observed during and after the perturbations in the early trials. As the subjects experienced more perturbations, the onset and duration of TA over the time course of the perturbation approached normal ranges for these measurements. The subjects co-contracted G of both legs immediately after the treadmill S-Inc in the early trials. However, the extent of this co-contraction was reduced in the later trials. The subjects adjusted the timing of locomotor related G activity to the treadmill S-Inc, so that burst activity immediately after was temporally integrated into the locomotor related muscular activity. Thus by the time of sixty trials, the effects of the perturbations during locomotion were substantially attenuated.

DISCUSSION

When a long-duration perturbation was repeatedly applied, locomotion was disrupted in the early trials. However, after experiencing more perturbations, trunk motion and stepping were close to the near normal level. This indicates that the subjects developed a strategy to maintain close to their normal locomotor pattern, by compensating for the mechanical and temporal features of this perturbation during and after the perturbation. These adaptive mechanisms were similar to that seen in the EMG pattern. The compensatory muscular activities in response to the perturbations seen in the early trials were minimized and approached a near normal locomotor EMG pattern which was maintained despite the dramatic changes in treadmill speed. The results suggest that locomotor adaptive mechanisms operate to establish a near normal locomotor pattern despite the presence of intermittent perturbations.

This research was supported by the Flinn Foundation and NIH grant PO1 NS30013.

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Computer Simulation of Curve-Ball Kicking using Coupled Euler-Lagrange Approach

T. Asai*, M. Nasako**, O. Murakami** and T. Akatauka***

* Fac. of Ed., Yamagata University, Koshirakawa, Yamagata, 990, Japan.

** ASICS Co., Takatsukadai, Nishi-ku, Kobe, 651-22, Japan.

*** Fac. of Engi., Yamagata University, Yonezawa, Yamagata, 990, Japan.

Introduction

The purpose of this study is to clarify the ratio of revolution of the curve-ball and the ball-foot interaction during the curve-ball kicking in football using a high speed video camera (4,500 fps) and fluid-structure interaction Finite Element Analysis.

Methods

Experiment

Three university football players were chosen as the subjects. The players give the curved kick toward left side to a regular football goal 25m away. A high-speed camera was set up 1.5m away from the side direction. The ball used in this experiment was an official FIFA ball (434.6 g, 90,000 Pa). This experiment was photographed using the high-speed camera (FASTCAM-ultima), which can take 4,500 frames per second with 256×256 pixels, and was recorded on a VTR. The ball velocity and the ratio of revolution were numerically calculated from the value of the digitized data. The contact time of the instep with the ball was obtained from the number of frames in which the contact between them was observed.

Simulation

The foot model and the surface of the ball were defined using Lagrangian frame of reference and the Lagrange Finite Element Method was used. The air inside the ball was defined using Eulerian frame of reference and the Euler Finite Volume Method was used (Fig. 1). The air model of that was defined by the gamma law equation of state. This solution used explicit time integration codes (MSC/DYTRAN) of Finite Element Analysis. Five different offcenter distances in impact were set in computer simulation as Case 1 ~ Case 5.

Results and Discussion

The average value of ball velocity in experiment was 23.28 m/sec (S.D.=1.01 m/sec), that of contact time in experiment was 8.16 msec. (S.D.=0.78 msec.) and that of the ratio of revolution in experiment was 8.72 r/sec. (S.D.=0.91 r/sec.). The spinning ratio of the curveball in baseball was about 30 r/sec. (R. G. Watts et al, 1990). It seems that the ratio of revolution in curveball kicking was fewer than that of baseball. As compared the Case A (coefficient of friction = 0, offcenter distance = 0.06m) with the Case B (coefficient of friction = 0.7, offcenter distance = 0.06m) in computer simulation, the ratio of revolution in the Case A was 8.8 r/sec., and that in the Case B was 12.8 r/sec. It is considered that the important factors to spin a ball in curveball kicking are the effect of friction and the moment of revolution by not co-axial impact with deformation of a ball.

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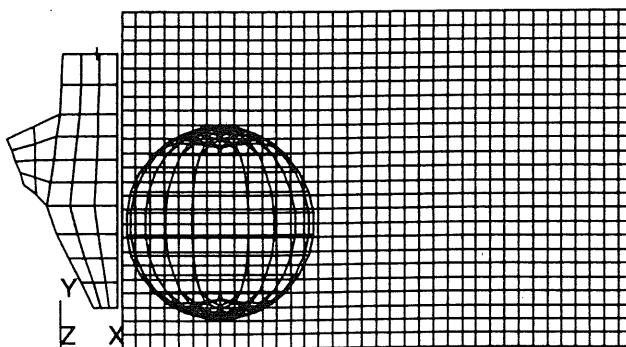


Fig. 1. The ball-foot interaction model using a coupled Euler (ball) - Lagrange (foot) approach.

Computer Simulation of the V Style Technique in Ski Jumping using Computational Fluid Dynamics

T. Asai*, M. Kaga** and T. Akatauka***

* Fac. of Ed., Yamagata University, Koshirakawa, Yamagata, 990, Japan.

** Fac. of Ed., Okayama University, Tsusimanaka, Okayama, 700, Japan.

*** Fac. of Engi., Yamagata University, Yonezawa, Yamagata, 990, Japan.

Introduction

Recently, the V style (skis together or crossing in back and spread in front) technique in flight phase is one of the most important and fundamental technique in the ski jumping. A lot of studies were performed about the V style technique in Biomechanics. For example, Watanabe (1992) analyzed the V style technique using a wind-tunnel experiment. Arndt et al (1995) investigated about the transition from take-off to early flight using VTR analysis. The optimization study on the flight trajectory of ski jumping with the aid of optimal control theory was performed by Yoshida et al (1996). However, there are a few studies about flowfield around the athlete and ski surfaces. Then, the purpose of this study is to clarify the characteristic of flowfield about the V style technique in flight phase using Computational Fluid Dynamics.

Method

Two triangular surface meshes were produced on simplified dummy models and skis in the Case 1 (attacking angle of the skis = 0 deg.) and Case 2 (attacking angle of the skis = 20 deg.) using GEOMESH (Fluent Inc.) and PATRAN (MSC Inc.) which were solid modeling and FEM analysis tools. Approximately 20000 tetrahedral cells were used to describe the solution volume which consisted of the dummy model and skis surfaces, a symmetry plane, and a cartesian outer boundary. The incompressible steady-state analysis were performed with RNG k- ϵ turbulent flow model. The initial velocity of inlet boundary was 25 m/sec. and the exit boundary was defined by static pressure. The calculations by Fluent/uns (Fluent Inc.) required 100 iterations to arrive at this solutions with run times on a EWS at about a half hour for each calculation.

Results and Discussion

The high intensity dragging force on the athlete and ski surfaces were observed at the head and feet in both Case 1 (attacking angle of the skis = 0 deg.) and Case 2 (attacking angle of the skis = 20 deg.). The dragging force of Case 1 was 24.06 N and the lifting force of that was 0.33 N. The dragging force of Case 2 was 73.21 N and the lifting force of that was 135.74 N. Accordingly, the Lift/Drag ratio of Case 1 was 0.0137 and that of Case 2 was 1.854 (Fig. 1). The Case 2 generated more lift than the Case 1 and slightly higher drag.

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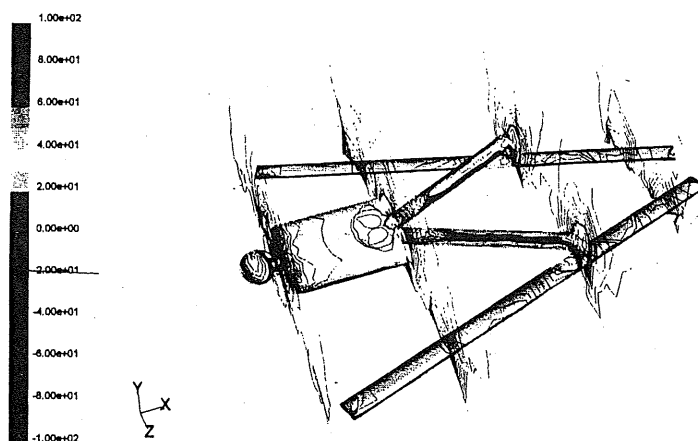


Fig. 1. An example of contour plot of static pressure around the V style Jumper.

COMPARISON OF KINEMATIC GAIT PARAMETERS OF ELDERLY THAI WOMEN WITH YOUNG THAI WOMEN

U. Rattanapraser^{*}, C. Akamanon[†], R. Smith^{*}, K. Chaipackdee[†], S. Pichaiyongwongdee[†]

^{*}Faculty of Health Sciences, The University of Sydney, AUSTRALIA

[†]School of Physiotherapy, Mahidol University, THAILAND

INTRODUCTION

As a basis for objective and quantitative judgements in routine clinical gait evaluation and training of patients with gait abnormalities, a gait analysis database would be a useful tool. Studies of kinematic gait parameters show degenerative changes in elderly women when compared with those of young women¹⁻⁴. Differences in gait characteristics between samples taken from Japanese and American populations have been found¹⁻⁴. The question arises "Are the same differences seen in the Thai population?". Thus this study focuses on the quantitative analysis of the gait characteristics of elderly Thai women compared with young Thai women. Quantitative gait characteristics of elderly Thai Women have not been reported. Temporal-distance variables and the sagittal angular displacement of the ankle, knee, and hip joints at discrete points in the eight phases determined by Perry⁵ of the right gait cycle were investigated.

METHODS

Twenty elderly Thai women aged 60-79 years and twenty young Thai women aged 20-24 years had 6 markers placed on the shoulder, thigh, leg and foot. Each subject was videotaped and footprinted during barefoot walking at self-selected speed on a 10-m walkway for three successful trials. The trajectories of the markers in sagittal plane from recorded video was digitised using the Peak2D Motion Measurement SystemTM. Then the information was transferred to a spreadsheet to calculate the angular displacements following the definitions given by Winter⁶. Temporal-distance variables were obtained from footprints⁷.

RESULTS

There were significant differences between the compared groups in both temporal-distance variables and the sagittal angular displacements. The elderly women demonstrated decreased step and stride lengths, stride length to leg length ratio, stride length to height ratio, walking speed expressed in cm/sec and stature/sec ($p < 0.001$), and cadence ($p \leq 0.01$). They walked with longer stride time ($p \leq 0.05$), stance time ($p \leq 0.001$) and double-support times (left-right and right-left) ($p \leq 0.001$). The elderly group also showed differences in their angular kinematics ($p \leq 0.05$) at some events of the gait cycle including initial contact; at ankle and hip, pre-swing; at knee and hip, initial swing; at ankle and knee, and terminal swing; at knee and hip.

DISCUSSION AND CONCLUSION

The results suggested that the elderly group had a functional decrease in walking, possibly due to degenerative changes. This study produces evidence that the gait characteristics of elderly Thai women were different from those of the young which support the previous studies¹⁻⁴. It indicates that the outcomes of treatment given to the elderly patients with abnormal gait should not be expected to be the same as those of the young group. Some aspects of the results were different from the reports of previous studies which were conducted on samples taken from Japanese and American populations. It confirms that gait characteristics vary with body built, gender, and country of origin.

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This study was partly supported by the University Development Committee Project of the Ministry of University Affairs, Thai Government, Bangkok, THAILAND.

Laryngeal Dynamic Gestures and Muscle Activities in Human Suspensory Propulsion

Morihiko OKADA¹⁾, Hideo OKA²⁾, Misao MIYASHITA¹⁾, Seiji NIIMI³⁾, Tasuku KIMURA³⁾, and Sugio HAYAMA⁴⁾

1)Univ. of Tsukuba, 2)Hyogo University of Education, 3)Univ. of Tokyo, 4) Kansai Med. Univ.

INTRODUCTION

The function of human larynx is claimed to be threefold; to generate primary vocal sounds, to eliminate extrinsic objects, and to provide thoracic stability for upper limb exertions through air-trapping (Hayama, 1996). The third function was assumed by Negus (1949) to have been derived from locomotor adaptation in the ancestral primate stock to arboreal environments. However, actual laryngeal gestures in primates during arboreal locomotion have not been observed due to technical difficulties. At the last ISB congress in Jyväskylä, we presented a preliminary paper reporting direct observation of laryngeal gestures in behaving human subjects by using specially devised endoscopic video-system. Here, we present in-depth analysis of the laryngeal behavior during suspensory propulsion (brachiation) in human subjects, as correlated with the body motion and muscle activities.

METHODS

Nine subjects including 2 world-class gymnasts participated in this study. They were asked to self-suspend from a horizontal ladder with stepping bars at intervals of 0.3 m, and to advance by alternate arm swings. Laryngeal dynamic gestures during the propulsion were directly monitored with the endoscopic video-system. Whole body motions of subjects in the side view were video-recorded and synthesized with the endoscopic images. Surface EMGs were recorded from 8 muscles in the upper limb and back. For nongymnasts, EMGs were also obtained from abdominal and lower limb muscles. EMG and video records were synchronized by using video-frame signals.

RESULTS AND DISCUSSION

In the nongymnastic subjects, laryngeal closures were frequently observed during the propulsion. They occurred around the period when the subjects were about to swing one of the arms forward while self-suspending by the other arm, and continued until the subjects grasped the stepping bar ahead with the swinging arm. On the other hand, the closure never occurred in the gymnasts throughout the cycle of brachiation. Laryngeal constrictions in the nongymnastic subjects corresponded to enhancement in the activities of the anterior portion of the deltoid, costo-sternal portion of the pectoralis major, and trapezius muscle of the holding arm, while these activities were inconspicuous in gymnasts. In the nongymnasts, abdominal muscle activities were also higher in the presence of the laryngeal closure than in its absence.

These results suggest that laryngeal sphincteric actions facilitate explosive exertion of the upper limb as well as forceful control of the floating lower limbs by stabilizing the thoracic and abdominal cavities through air-trapping.

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STANDING UP AND SITTING DOWN AT TWO DIFFERENT SEAT HEIGHTS

RB Shepherd, C Dean, AD Hirschhorn

School of Physiotherapy, The University of Sydney, Australia

INTRODUCTION

Standing up (STS) and sitting down (SIT) are 2 actions in common everyday use and the ability to perform them effectively is critical to independent life. Knowledge of the major spatio-temporal features of these actions provides the basis for training in rehabilitation for individuals with motor impairment. STS has been extensively investigated. However, there are few studies of SIT which is often regarded as analogous to and the reverse of standing up (Rodosky et al. 1989, Kralj et al. 1990). The purpose of the study was to investigate characteristics which we predicted would differ between STS and SIT: the magnitude and timing of peak forces at hip, knee and ankle and momentum of the total centre of body mass (CBM). Since practising STS and SIT from a high seat has been recommended for patients with poor force generating capacity (Carr & Shepherd 1987), a secondary aim was to examine the variables at 2 seat heights.

METHODS

Six female subjects (18-25 yrs) with both feet on a forceplate were videotaped as they stood up and sat down from a seat adjusted to 100% and 130% of shank length. Segmental kinematics and kinetics were calculated in 2 dimensions from synchronised videotape and forceplate data. Joint moments of force were normalised to subject's body mass. The actions were each studied in 2 phases divided by thighs-off (TO): seat contact (SC) and non contact (NC) phase.

RESULTS

Although SIT took significantly longer than STS at each seat height, this difference was a function of the time spent in SC in SIT. The NC phase did not differ between the 2 actions. Overall, moments at hip, knee and ankle were extensor throughout both actions. Peak hip moment did not differ significantly across action or seat height. Peak knee moment, however, was significantly lower in SIT than STS and at 130% seat height compared to 100% and at 130% this decrease was offset by a significant increase in peak ankle moment ($p < 0.01$). Timing of peak moments was very consistent at the hip, occurring at the beginning (STS) and end (SIT) of NC phase at both seat heights. The timing of peak knee moments was variable and intraclass correlations showed this variability to be due to inter-subject variability. Peak horizontal linear momentum of the total body centre of mass occurred prior to TO in STS and at TO in SIT. The path was smooth and continuous in STS but displayed a discontinuity in SIT (Fig. 1).

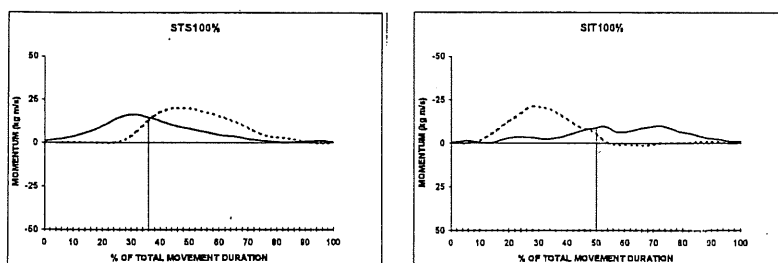


Fig. 1. Representative trials of horizontal (solid) and vertical (broken) linear momentum of CBM in STS (L) and SIT (R). Vertical line= TO.

DISCUSSION

The kinematic and kinetic patterns of movement in both SIT and STS are similar and relatively stable, reflecting that biomechanical characteristics emerge naturally from linked-segment dynamics and task and environmental constraints. There is no evidence that the NC phase of SIT is longer than STS suggested by Kralj et al. to reflect "increased precaution". STS is produced predominantly by concentric muscle work and SIT by eccentric work and the results suggest different control mechanisms. In particular, the discontinuity of horizontal momenta traces in SIT suggests this action is under more feedback control than STS, with corrections necessary due to the braking of the body mass and the need to aim for the seat. The greater consistency in timing of peak knee moment in SIT also suggests that the knee is more constrained in this action than in STS. Control at the knee is complex in closed chain actions, since it is influenced by muscles which do not span the joint but which have attachments to thigh and shank segments.

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MULTI-JOINT LINKS IN APPLICATION TO HAND PROSTHESIS DESIGNING

M.N.Vasiljev, V.P.Tregoubov, V.M.Velikson
St.-Petersburg Social College, St.-Petersburg, Russia

INTRODUCTION

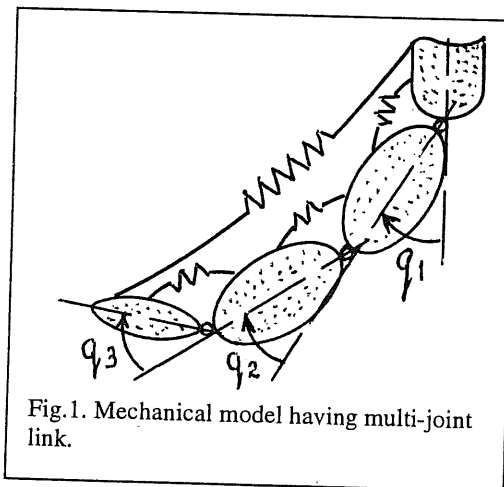
As is well known many skeletal muscles are multi-joint in function. Among these are the finger muscles having the tendons which pass over the radioulnar articulation and several finger joints. Characteristic properties of multi-joint muscle, as well as multi-joint mechanical links at all, are poorly studied (Tregoubov and Grafeev, 1989). In this connection the aim of this work was studying the main properties of multi-joint mechanical links as applied to a designing of hand prostheses.

METHOD

With the above aim in mind the mechanical model consisting of four rigid hinged bodies was proposed (Fig.1). The adjacent bodies are connected by elastic one-joint links (springs). In addition to this two kind of multi-joint links, connecting the first and last bodies, were considered: elastic links and inextensible ones. The comparative mechanical properties of the one-joint links and the multi-joint links have been examined.

RESULTS

Since a force developing by the multi-joint elastic links, as well as a potential energy, is a function of sum of the relative rotation angle changes that is $F\{q_1 + q_2 + q_3\}$, so it



leaves its value unaltered if the angle sum remains constant. It means that such a system possessing of the multi-joint links in absence of external forces can have a neutral equilibrium only. That is not sufficient for a system stabilisation. At the same time the one-joint elastic links can provide a stable equilibrium. In the presence of gravity forces the equilibrium equations can not be satisfied at any values of angles except $q_i = 0$. As to control a system movement, it can be exerted by means of the inextensible multi-joint link displacement (shortening-lengthening) with the necessary availability of one-joint elastic links.

DISCUSSION

The obtained results can be applied to designing the prostheses of hand. For this purpose the elastic one-joint links (spiral springs or special elastic cushions) must be located in each joint and the movement control can be exerted by means of a cable attached to the terminal phalange.

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USING THE CAM DEVICE IN CABLE CONTROLLED PROSTHESES

V.M.Velikson, V.P.Tregoubov, M.N.Vasiljev
St.-Petersburg Social College, St.-Petersburg, Russia

INTRODUCTION

A roller pivoted in the joint rotational axis is one of the devices, which is used in the prostheses having the cable control system. In this case the gear-ratio is held constant during a joint rotation. The roller radius, as well as other parameters of the prostheses construction, can be selected so that movements of the artificial extremity would be on the average similar to movements of the natural one (Velikson et. al., 1989). However, the reproduction of natural movements "on the average" is unsuitable when an invalid has to perform the movements along the specially preset trajectories (occupational operations and sport exercises).

METHOD

To meet the above requirement the roller, having a variable radius similar to cam, was proposed to use. The roller (cam) shape was defined as a dependence of the radius r on the joint rotation angle φ_0 . Based on the condition of the cable inextensibility the following nonlinear differential equation was derived to determine the function $r(\varphi)$:

$$\left(\frac{dr}{d\varphi}\right)^2 + r^2 = \frac{dS}{d\varphi} \frac{d\varphi}{d\psi}.$$

The function $S(\psi)$ defines the cable displacement during a rotation of the controlling (robust) joint. The specific form of this function depends on a shape of the controlling joint and a kind of harness. The known dependence of φ and ψ defines the trajectory of the natural extremity movement to be reproduced by the prostheses. To determine the function $\varphi(\psi)$ the special investigation of 3D upper extremity movements was carried out for the group of normal subjects.

RESULTS

The proposed method was used for the designing of the upper extremity prostheses. The cam was pivoted in the elbow joint axis. The cam shape was calculated by means of the above equation for a set of everyday movements, for three occupational operations and one sport exercise. As result using the cam instead of usual roller provided a closer approximation to the natural movements. It was confirmed during the tests of a pilot specimen.

DISCUSSION

Using the prostheses having the cam devices is particular advantageous for the special-purpose movements, such as occupational and sport ones, where following the preset trajectory is quite important.

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APPLICATION OF VERTICAL JUMP IN EVALUATING ATHLETES' MOTOR RECOVERY

R. RODANO¹, A. MINGRINO¹, R. SQUADRONE^{1,2}¹ Centro di Bioingegneria (Politecnico di Milano - Pro Juventute Fnd.), Milan, Italy² Associazione Scuola Viva, Rome, Italy

INTRODUCTION

Some changes of the mechanical output of the musculo skeletal system are the inevitable consequence of injuries. A partial recovery can lead to relapses or to potentially dangerous compensatory strategies, especially during the practice of sport activities. This justifies the need of specific methods to check the motor recovery of an athlete.

Aim of this paper is to present the results of a study designed to evaluate the lower limbs motor recovery of injured rugby players, by means of the mechanical joint output (net moments and powers) measured during bilateral vertical jumps.

METHOD

Five athletes (height = 1.78 m, SD = 0.03, body mass = 75 kg, SD = 3) who had suffered a unilateral ACL injury (INJ) within the 2-yr. period prior to the study, and eight uninjured (UNI) athletes (height = 1.80 m, SD = 0.05, body mass = 78 kg, SD = 7) were the subjects of this study. All the subjects are professional rugby players, playing in the Italian second league. After 20 min. of standard warm up, each subject performed 10 double-legged countermovement vertical jumps with the right foot on the force platform and 10 jumps with the left foot on the force platform. Kinematics data, concerning the spatial position of ten anatomical landmarks (five per leg), were recorded by means of an optoelectronic system (ELITE, BTS Srl) with a sampling rate of 100 Hz. Simultaneously, ground reaction forces were measured by a Kistler force platform and acquired with a sampling frequency of 1000 Hz. The internal joint centers, such as the corresponding moments and powers of the lower limb acting on the force platform, were estimated by using a special software which inputs were anthropometric, kinematic and kinetic data.

Peak joint moments and powers were used for the analysis together with the individual percentage asymmetry (ASY%) between right/left (for the UNI group) and healthy/ACL (for the INJ group) limbs calculated for each of the kinetic parameters.

RESULTS

Mean jumping height values were 0.421 m. (SD = 0.004) and 0.405 m. (SD = 0.005), respectively for the UNI and INJ group, with no significant group difference. For the three lower limbs joints, means and standard deviations for peak moments and powers, are presented in table I. In addition, are reported the percentage asymmetry between right/left and healthy/ACL limbs for the UNI and INJ groups, respectively.

Examining peak moments and powers values, the only significant group differences were found at the knee, with the ACL limb of the INJ group considerably lower compared to the homologues. With regard to the limb asymmetry, considering moments, the data reveal no significant differences between scores of the two groups. In contrast, examining powers, significant group differences were found at the knee, with the INJ group characterized by higher ASY values.

		UNI			INJ		
		Left	Right	ASY%	Healthy	ACL	ASY%
Moments [Nm]	Hip	104 (18)	114 (35)	13 (8)	112 (71)	129 (30)	9 (6)
	Knee	164 (44)	169 (47)	10 (6)	152 (71)	133 (72)	11 (6)
	Ankle	94 (17)	98 (14)	8 (6)	101 (20)	109 (15)	10 (7)
Powers [W]	Hip	394 (92)	414 (122)	10 (10)	477 (90)	482 (114)	13 (9)
	Knee	765 (70)	792 (112)	6 (5)	701 (90)	477 (124)	22 (7)
	Ankle	535 (176)	585 (185)	13 (7)	558 (108)	640 (101)	18 (9)

Table I

DISCUSSION

In summary, the kinetic analysis during vertical jump seems to provide useful information that can be applied to a clinical evaluation of the athletes during post ACL injury.

Indeed, differently from the results of Kannus, who found no significant limb differences between ACL injured knee and healthy homologues using an isokinetic dynamometer, the subjects of the INJ group had substantial changes from normal patterns and magnitude of the power at the knee joint. This adaptation could be interpreted as a subconscious desire to reduce the net quadriceps force required during jumping.

CLASSIFICATION OF SKI JUMP TAKE-OFF TECHNIQUES BY JOINT POWER

T. SASAKI, K. TSUNODA, E. UCHIDA* and H. HOSHINO**

Hokusei Jr. College, Sapporo, Japan, *Kokugakuin Jr. College, Takikawa City, Japan,

** Ebetu City Gymnasium, Ebetu City, Japan

INTRODUCTION

In ski jumping, take-off action is the most important factor for having ascent force. Ascent force acts on the ground and is required for a jumper to obtain initial velocity and angular momentum. Reaction force gives the body both translational and rotational momentum at take-off. In take-off technique, we ought to aim for optimum movements of the joints and body segments because reaction force is the result of the integrated kinetic parameters of each joint or segment. When jump action produces force, the body, which is affected by ground reaction force. Reaction force makes the jumper revolve. Jump action in laboratory tests was characterized by two techniques of contribution of joint power. The purpose of this study is to clarify the role that joint power and energy play in ski jumping for world class jumpers.

METHODOLOGY

The takeoff action of six jumpers was analyzed from video taken at World Cup competitions at Okura jump hill from 1995 to 1996. Camera speed was 240 frames per second (made by FOR'A Co.). We observed moment force and joint power resulting from actual ski jumps, using seven segment linkage models of seven degrees of freedom. The segments are foot, lower leg, thigh, trunk, head, upper arm and forearm segments. We calculated the torque, energy and power that were produced by each joint. The data from video were collected by computer. An inverse kinematics solution was applied to video analysis. The data analysis system was developed for a Macintosh computer.

RESULTS

Most of the power, from initial action until take-off, is produced by two joints, the hip and the knee joint. The knee joints were the major energy and power producers in all jumpers. Power production in joints was generated in regular order for almost jumpers.

Jump performances are classified into three techniques according to joint power production. The first technique showed the tendency that the hip joint and after some delay the knee joint produced power. The values are positive until maximum power output. This technique is named the "Phase Shift Power Generation Technique". The second technique had the same tendency of power generation as the previous but the delay between maximum knee joint and hip joint power was longer. This technique included a remarkable action which was the forward rotation of the leg. This technique is named the "Phase Shift and Leg Rotation Technique". The last technique had a tendency in which both hip and knee joint powers increased at almost the same time and the values were positive until maximum. This technique is named the "Close Power Generation Technique".

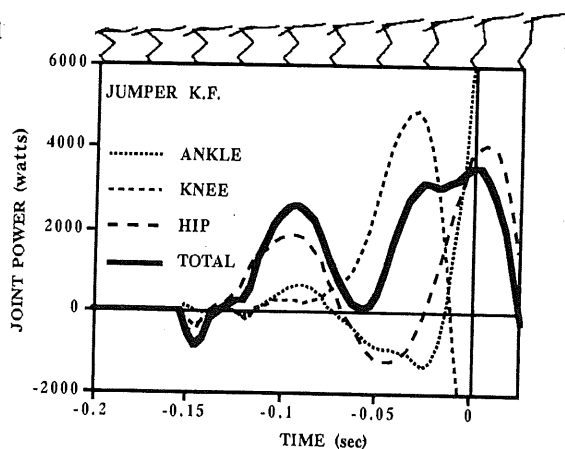


Fig. Phase Shift and Leg Rotation Technique.

HOW TO GAIN SPEED RUNNING OVER WAVES IN ALPINE SKIING

H.-B. Schmiedmayer, P. Lugner

Institut für Mechanik, Technische Universität Wien, Wien, Austria

INTRODUCTION

In downhill events of alpine skiing competitors try to minimize time for a given course. The result depends mainly on the skill of the skier. In this paper it will be shown that this is not only true for turning situations but also when running straight over a wavy surface.

METHOD

In a first step a single mass model is set up, where the ground is represented by a cyclic function $y = f(x) = f(x + \lambda)$ with wavelength λ guiding the reference point G (Fig. 1). The position of the center of mass M of the skier with respect to the ski can be described by the height $h(x)$ and lean angle $\beta(x)$. The moment of inertia $I_M = m \cdot i^2$ is assumed to be constant for the current model. Effects of the frictional drag between ski and snow as well as air resistance are not included.

Using Lagrangian equations of the second kind to the following equation of motion is obtained:

$$\ddot{x} \cdot a(x) + \dot{x}^2 \cdot b(x) = c(x), \quad (1)$$

where $a(x)$, $b(x)$, and $c(x)$ are functions of $h(x)$, $\beta(x)$ and $f(x)$ and their first, second, and third derivatives.

To show the principal effects the following functions were chosen:

$$f(x) = A \cdot \sin\left(x \frac{2\pi}{\lambda}\right), \quad h(x) = h_0 + A_h \cdot \sin\left(x \frac{2\pi}{\lambda} + \varphi_h\right), \quad \beta(x) = A_\beta \cdot \sin\left(x \frac{2\pi}{\lambda} + \varphi_\beta\right). \quad (2)$$

Equation (1) is solved numerically.

RESULTS

All results are compared to the so called 'stiff' skier with $A_h = A_\beta = 0$. The quality of increase in speed and decrease in time for one period depends mainly on the phase-angles φ_h and φ_β whereas the quantity of these effects depends mainly on the amplitudes A_h and A_β . Therefore the following results were calculated with the following parameters:

$$h_0 = 0.9m; \quad \dot{x}_0 = 10m/s. \quad (3)$$

The results for the stiff skier ($A_h = 0$, $A_\beta = 0$) as well as for optimal up-down movement only ($A_h = 0.2m$, $A_\beta = 0$) and for optimal forth-back movement only ($A_h = 0$, $A_\beta = 10^\circ$) are shown in figure 2.

DISCUSSION

The simple model shows that an increase of the speed at the end of a full wave with respect to the speed at the beginning of the wave can be obtained mainly by the up and down movement $h(x)$. On the other hand the back and forth movement $\beta(x)$ results mainly in shorter times for passing a whole wave without considerably increase in speed with respect to a normal run ($h(x) = \text{const}$ and $\beta(x) = 0$). Both motions can be combined to give an optimal motion pattern for overall increase in speed and shorter passing times. Furthermore it is assumed that similar mechanisms influence the speed and time while cornering.

For a more detailed analysis effects frictional drag between ski and snow as well as air resistance should be included in this model to find optimal techniques for running over wavy ground in downhill racing.

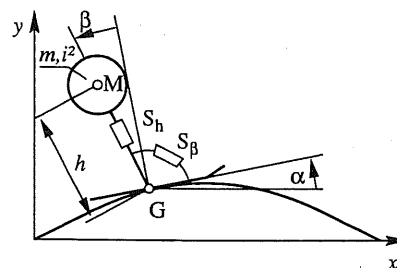


Figure 1: Simple single mass-model for the skier on a wavy ground with two positioning elements S_h and S_β for the height h of the center of mass M and the lean-angle β respectively.

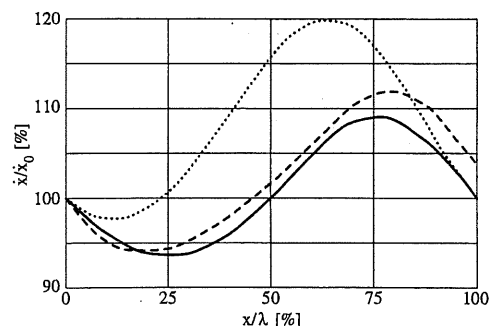


Figure 2: Horizontal speed-factor $\dot{x}/\dot{x}_0$ of point G as a function of x for the stiff skier (solid line) and for the optimal up-down movement only with phase angle $\varphi_h = 0.5\pi$ (dashed line) and for optimal forth-back movement only with phase angle $\varphi_\beta = 0.5\pi$ (dotted line).

SIMULATED AND ACTUAL SKI JUMPING TAKE-OFFS DIFFER IN PLANTAR PRESSURE AND MUSCLE ACTIVATION PATTERNS

M. Virmaavirta, P.V. Komi and J. Perttunen

Department of Biology of Physical Activity, University of Jyväskylä, Finland

INTRODUCTION

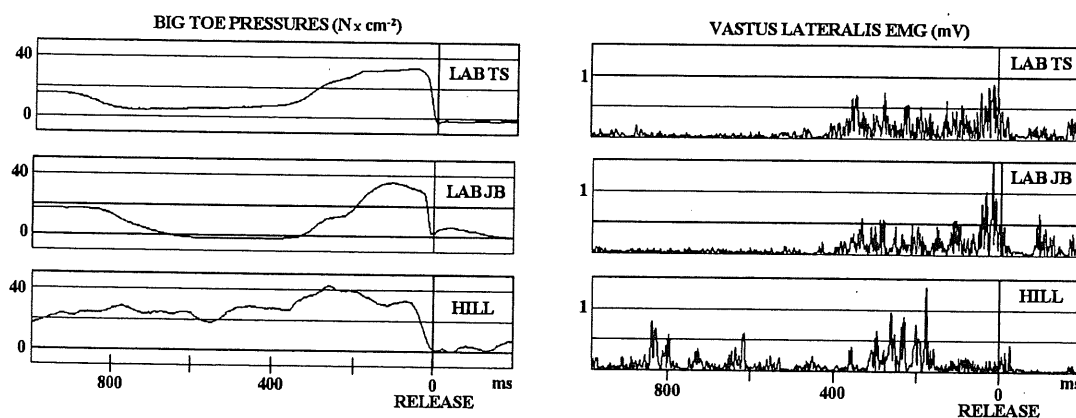
Timing of powerful take-off is important for successful ski jumping performance (Komi & Virmaavirta 1997). Simulated take-offs in laboratory or in training have often been used to mimic actual ski jumping take-off. The present study makes an effort to characterize how similar these take-offs are with regard to plantar pressure and muscle activation patterns.

METHODS

Ski jumping take-offs of world class ski jumper were studied in three different conditions: simulated take-offs in laboratory (LAB)(1) with training shoes (TS) and (2) with jumping boots (JB) as well as in (3) actual jumping hill conditions (HILL). Muscle activities and plantar pressure distribution were recorded by Paromed Datalogger[®] system with the same electrode and insole set-up in all three conditions. Two components (Fz and Fy) of the ground reaction forces were also recorded for registration of the performance level in LAB measurements.

RESULTS AND DISCUSSION

Decrease in the resultant take-off velocity in LAB JB (3.24 ± 0.09) compared to LAB TS (3.36 ± 0.12) can most probably be explained by less effective utilization of plantar flexor muscles with stiff ski boots (revealed by decrease of 10 % in EMG of the gastrocnemius (GA) during the push-off phase). Even less effective use of GA (decrease of 46 % in EMG) during the same phase in HILL conditions emphasize the action of knee extensor muscles in gaining vertical lift during the take-off. An example of the activation patterns of the vastus lateralis (VL) muscle as well as the plantar pressure in different conditions can be seen in the figure. The activation of the VL in HILL conditions is greatest during the early phase of the take-off corresponding to the peak of the measured pressure. The highest pressure during the take-off was measured under the medial side of the fore foot implying that while the jumper's center of mass moves forward the take-off action has a strong lateral component from both feet.



The results of this study demonstrate that in order to perform an effective take-off in actual conditions the jumper must start the preparation early during the inrun curve before the flat part of the apron. In LAB conditions the preparation for the take-off began already approximately 770 ms before the release instant corresponding to 18 meters at the speed used in HILL condition. However, the release instants in LAB and HILL conditions do not match exactly, because at the release instant of the HILL the knee extension is not completed.

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THE DEVELOPMENT OF SPRINT TRAINING MACHINE

Kando KOBAYASHI, Masaaki SUGITA, and Noriko MATSUGAKI
Dept. of Life Sciences (Sports Sciences), Graduate School of Arts and Sciences,
University of Tokyo
3-8-1 Komaba, Meguro-ku, Tokyo 153 Japan

Purpose

Biomechanical research results based on a motion analysis of top sprinters of the world (such as Carl Lewis) confirmed that it is necessary to strengthen the extensor muscle group at the hip joint in order to quickly move thighs around the hip joint as the pivot. Although the importance of training of those extensor muscles has been pointed out, there has not been a conclusive method for strengthening the muscle groups. The authors intended to develop new type of training machine which strengthen the extensor muscle group at the hip joint and improve running performance for many runners.

Concept of Invention

The invention of new training machine "Sprint Training Machine" was related to the concept as follows;

- 1) To effectively strengthen the extensor muscle group at the hip joint (especially the hamstring muscles) are necessary for an ideal running motion through an exercising form which is similar to the actual running motion.
- 2) To make a user learn a leg movement pattern which create such running motion that is considered to be ideal for running fast by having person repeatedly trace a certain locus of movement on the training machine.
- 3) To correct an imbalance between right and left sides of the form which occurs during a natural walking and running, thereby helping the prevention of injury and to provide an exercise stimulation which is effective for improvement of the health. Accordingly, the new machine was developed based on the idea that ideal movement of legs should be constituted by the combining motions which are based on a circular motion like the theory of circular wheel movement and reciprocating movement of the axis of circular wheel.

Composition of "Sprint Training Machine"

A pair of forward and backward movable pedestals respectively provided with a pedal, an arm and a crank and operable independently from each other; a power engine for shifting the movable pedestals backward and forward by a certain distance which is adjustable within a 1 m at a certain reciprocating speed which is also adjustable. A power transmitting portion; A load controlling means for controlling a resistive load during a pedaling exercise such that a resistive load is applied to a user when the pedal is located lower than a horizontal position of the pedal arm while no resistive load is applied to a user when the pedal is located higher than the horizontal position of the pedal arm.

A hand-rail for supporting the operator to balance himself during the pedaling exercise wherein a weight of the operator is supported by the right and left pedals; wherein a sprint training machine being constituted such that a resistive load is applied to the operator when the pedaling exercise is conducted at a position lower than the horizontal position of the pedal arm and that no resistive load is applied to the operator when the pedaling exercise is conducted at a position higher than the horizontal position of the pedal arm, while a rotation axis of the pedal arm is shifted back and forth by a certain distance by means of the power engine.

Effect of training using "Sprint Training Machine"

1. Effect of interactively learning the ideal locus of movement in running motion. Although it is possible to imaginatively understand that the foot portion should move along a circular orbit just like a wheel in a model movement for most efficient, ideal running motion, it has been impossible to actually experience or learn such a movement. However, the present invention has made it possible to learn the ideal model movement in which the foot portion moves along a circular orbit.

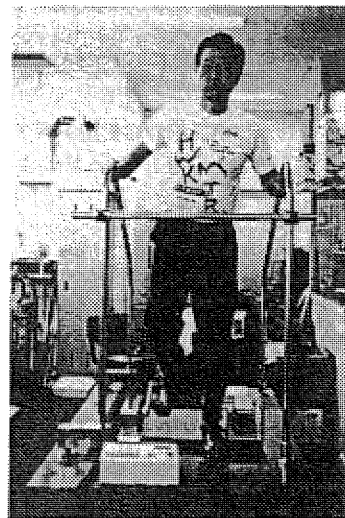
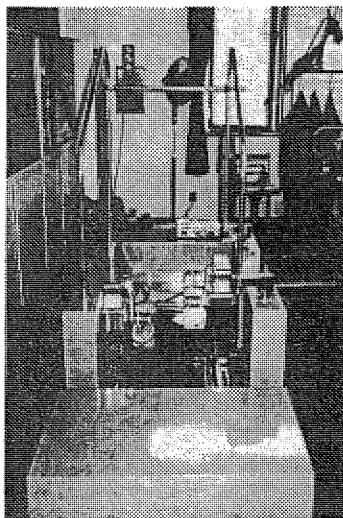
All the athletes (sprinters and marathon runners) who did training with use of the sprint training machine according to the present invention has experienced remarkable correction of respective running forms immediately after the training and their records have also improved.

2. Muscle strength training effect which directly leads to improve of sprint ability.

3. Effect of correcting imbalance between right and left sides of running form.

4. Effect of learning relaxation in running motion.

5. Effect of improving suppleness of body.



AN ADAPTABLE RISK-FREE STRENGTH TRAINING DEVICE WITHOUT PHYSICAL CONSTRAINTS

Yoky Matsuoka¹ and Larry C. Miller²

The Massachusetts Institute of Technology, Cambridge, MA.¹

Boston Biomotion Inc., Wellesley, MA.²

INTRODUCTION

Recently, the importance of strength training is emphasized for increasing athletic performance, for maintaining health, and even for prolonged life. To fit these needs, many linear and rotary resistance devices became commercially available. However, human's athletic movements cannot be matched well on their constraints which often results in inefficient training or even in injury.

In order to overcome these problems, we have constructed a three-dimensional resistance trainer which is adaptable to the exerciser's kinematics and needs.

METHODS

Our machine is a completely passive, spatial manipulator. It has six degrees of freedom in a workspace that is a half sphere approximately 1.5 meters radius. There is a gimbal wrist with three non-actuated degree of freedom and it can be used with a hand or a foot. The three positional degrees of freedom are actuated with brakes and accomplished through serial rotary and linear joints that are cable driven. This machine is controlled by a motion controller with DSP and programmed incorporating interface.

RESULTS

Using our novel strength training device, any kind of strength training is possible. For example, in a curling movement, the bicep muscle can exert about 70% of its potential when the arm is straight (180 degrees), 100% of its strength at 100 degrees and 70% again at 60 degrees. With the machine, the force field can be created to match the kinematics of the user and the strength to achieve the maximum efficiency without injury. Also by keeping track of time, the applied force can be adjusted to be stronger as the training progresses.

At the appearance of pain or fatigue, the user can instantaneously decrease the damping on the fly or stop the motion. Because the resistance is not a weight that can be dropped, the user experiences near zero loading when the motion is stopped. This environment can also be created with a machine which is actuated by motors. However, during an extreme case which the machine fails, it cannot be guaranteed that the motored machine will be stable or harmless to the user.

DISCUSSION

A unique benefit of our machine concept is that the exerciser is freed from the line of action of the force constraint present at the machine/user interface in all current forms of weight or resistance training. Human force production in such activities is highly constrained because of the need to reconcile body position, joint axes, and leverage to the line of action of the force. With our device, the user has complete control over the force direction and magnitude with the end-point force-velocity collinearity. Research has shown that purposeful motion is degraded without force-velocity collinearity and this is easily demonstrated. As an additional benefit, the machine can chart changes in force, position, velocity, acceleration, power, work, and range of motion in real time. The user can have an access to these accurate data as he/she trains.

The device we constructed provides a safe and efficient strength training environment with real-life ranges of motion with appropriate modes and increments of resistance operating under the complete volition of the user. With this machine, we believe that we can take the strength training to the next level.

ACKNOWLEDGEMENT

We would like to thank all Barrett Technology Inc. employees for their contributions.

EVALUATION OF CUSHIONING PROPERTIES OF RUNNING FOOTWEAR

D.G.E. Robertson¹, J. Hamill², D. A. Winter³

¹University of Ottawa, Ottawa, ON, Canada; ²University of Massachusetts, Amherst, MA, USA;

³University of Waterloo, Waterloo, ON, Canada

INTRODUCTION

The cushioning properties of athletic footwear have been evaluated *in vivo* using either ground reaction force or accelerometry data or *in vitro* with impactors. Unfortunately, none of these methods directly measure the amount of cushioning that a runner experiences. An alternative method for the evaluation of cushioning may involve the use of the deformation power in the foot. **Deformation power** (P_{def}) in the foot may be defined as:

$$P_{def} = dE_{foot}/dt - (F_{ankle} \cdot v_{ankle} + M_{ankle} \omega_{foot})$$

where $F_{ankle} \cdot v_{ankle}$ is the joint force power and $M_{ankle} \omega_{foot}$ is the power produced by the ankle moment of force. Previously it was reported that P_{def} (also called power imbalance) was a result of the elastic deformation of the foot (Robertson & Winter, 1980; Seigel, Kepple & Caldwell, 1996; Winter, 1996). This power source, especially during the weight-acceptance phase of support, is due primarily to deformation of the shoe midsole and tissues in the foot. The purpose of this study was to determine if running footwear with differing cushioning properties could be distinguished by the amount of deformation power.

METHODS

Five healthy people served as subjects in this experiment after giving informed consent. Each subject performed five trials at 16 km/h (6 min/mile) in each of two footwear conditions: soft midsole (40-43 Shore C durometer) and hard midsole (70-73 Shore C durometer). In all respects the footwear were identical excepting for their midsole durometer measures. Subjects ran across a force platform from which data were collected at 1 kHz while being videotaped from a sagittal view with a 200-Hz video camera. The rate of change of segmental foot energy (dE_{foot}/dt) was calculated from the video data. Inverse dynamics methods were used to calculate ankle forces, moments and powers. Deformation powers (P_{def}) were then computed using the equation presented above.

RESULTS AND DISCUSSION

The P_{def} was characterized by negative work for approximately the first two thirds of the support period due to energy absorption by the foot or the midsole of the shoe. In the final third of the support period, there was a small positive burst of power that supplied energy to the foot either due to energy return from the midsole or more likely by work done at the metatarsal-phalangeal joint. Table 1 presents the mean peak P_{def} and the negative and positive work done during support for all subjects in both the hard and soft midsole conditions. The P_{def} for the hard midsole shoe produced greater peak powers and absorbed more energy than the soft midsole shoe. Figure 1 presents the five-trial ensemble for both soft and hard footwear conditions for an exemplar subject, however, there were considerable variations among subjects. It appears that the differences in the peak P_{def} may be influenced by body mass. In the lighter subjects the differences in the peak negative P_{def} were not as great as with the heavier subjects. This may be because the heavier subjects deformed the midsole to a greater extent. In conclusion, this technique differentiates between footwear with differing cushioning properties and offers a means of detecting the effects of shoe and foot cushioning on runners.

Table 1: Mean deformation power and work.

	Soft	Hard
Peak $-P_{def}$ (W)	-1211	-1369
Neg. work _{def} (J)	-440	-619
Peak $+P_{def}$ (W)	162	132
Pos. work _{def} (J)	131	64

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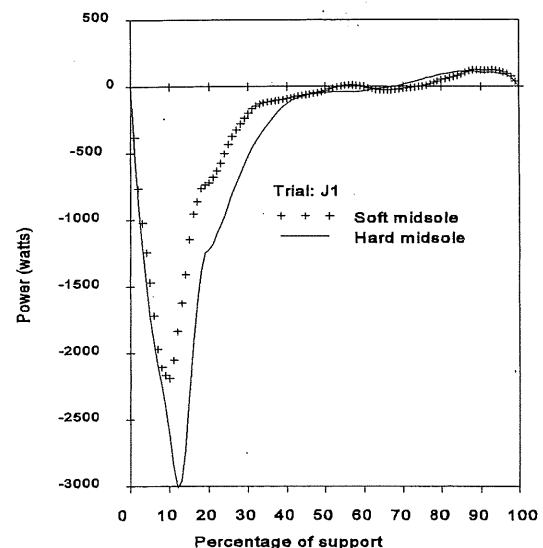


Figure 1. Ensemble averaged deformation power

PROSTHETIC FOOT AS SIMULATOR FOR ATHLETIC SHOE TESTING

M. R. Pitkin

Tufts University School of Medicine, Boston, USA

INTRODUCTION

Simulation of the human foot response to loads within the athletic shoe is needed for variety of testing protocols. It seems reasonable to look at prosthetic feet since they were designed to mimic not only dimensions but the functions of the anatomical foot. A classification of mechanisms of prosthetic feet available was done in this study. Analysis of three typical classes of mechanisms was conducted which suggested the Rolling Joint Foot and Ankle to be an adequate device for the shoe testing.

METHODS

There were many attempts to mimic the performance of the lost foot and ankle by a passive elastic system "charge-release," where "charge" corresponds to dorsiflexion and "release" corresponds to the following plantar flexion. One can say that the level of approximation of the prosthetic foot/ankle performance to those demonstrated by the real foot/ankle, depends on how well a "charge-release" mechanism is designed. The power released and the resistive curve, are two typical characteristics of any "charge-release" mechanism. Both characteristics can be used as targets in a synthesis of a new mechanism. A power released in plantar flexion period in real ankle, was chosen as a target in designing of the most contemporary energy storing (ES) feet (Gitter et al., 1991). While that "release" goal was achieved for plantar flexion, not enough attention has been paid to the "charge" component, namely, to the resistive curve in dorsiflexion period. Indeed, in all ES feet on the market, the charge during dorsiflexion is provided by

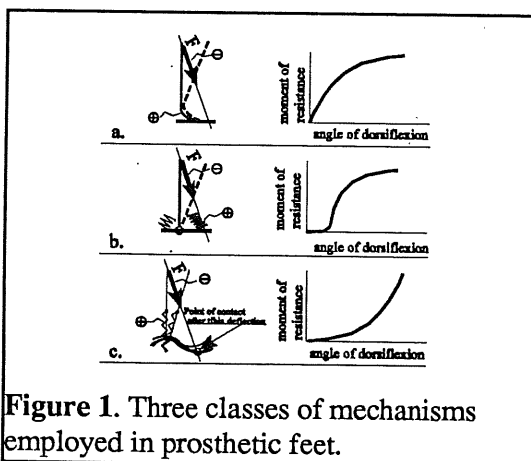


Figure 1. Three classes of mechanisms employed in prosthetic feet.

activation of bending or compression of elastic elements. The analysis shows that these types of mechanisms have non-concave (convex or linear) pattern (Fig.1,a,b), opposite to the concave pattern seen in the real ankle (c). This is also true for structures like Genesis and SACH feet, employing compression and/or extension. A Rolling Joint Foot (RJF) (Pitkin, 1996) was found to be adequate to that request, since as biomechanical study showed, a prosthetic gait synergy has been transferred to a more normal one.

DESIGN

The RJF comprises a higher-pair ankle and metatarsal joints. The joints provides rolling of curved contacting surface along each other with elastic resistance to deflection. Moving components are connected by linear springs of

extension type. Moments of resistance in both joints have a desired concave pattern seen in the anatomical joints (Fig.1,c) provided by the nature of cycloidal motion.

RESULTS

1. Theoretical analysis of the resistance to deflection of variety of prosthetic feet showed that higher compliance of RJF, makes it adequate to the anatomical foot' performance.
2. It is suggested to include RJF in a testing system "shoe - artificial foot."

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ACKNOWLEDGEMENTS

The study was supported by the Ohio Willow Wood Company and NIH Grant 1R43 A R43290-01A1.

A METHOD FOR THE STUDY OF THE DISTANCE BETWEEN ANKLE LIGAMENT INSERTIONS IN VIVO.

A. LUNDBERG, I.G. WINSON, T. HASHIMOTO, C. SARO, P. WESTBLAD

From the Dept of Orthopaedics, Karolinska Institute at Huddinge Hospital, Stockholm, Sweden.

We describe a method of analysing changes in the distances between the attachment points of the ligaments around the ankle during a variety of normal activities while simultaneously assessing joint kinematics.

MATERIALS AND METHODS

Three healthy volunteers had 2.4 mm Hoffman external fixateur pins inserted into the tibia, fibula, talus and the calcaneum using local anaesthetic. Four 10 mm reflective markers were attached securely to each pin. Pin placement was such that no interference was seen between pins, markers or normal anatomical structures. All subjects were pain free during the examination and could walk without a limp during the studies.

Following the insertion of the pins and attachment of the markers two radiographs were undertaken using the roentgen stereophotogrammetric method. The anatomical insertions of the ligaments and the reflective markers were subsequently digitised to provide the spatial relationship between the positions of the ligament insertions and the attached markers. Analysis of motion was then performed during normal walking using an three camera optoelectronic method of analysis (MacReflex) at 50 Hz, and the ligament attachment points were transferred to each frame of the gait file assuming an unaltered spatial relationship between each ligament attachment and the marker set of the corresponding segment. The distance between the attachments of each ligament was then calculated frame by frame.

RESULTS

Fig. 1 shows an example of calculated distances between the attachments of the anterior fibulotalar (FTA) and posterior fibulotalar (FTP) ligaments in one subject during the stance phase of walking. Differences during this movement were small in this example.

CONCLUSION

This technique may prove to be a useful way of analysing the change in ligament characteristics under various life conditions including normal activities in normal individuals and pathological changes caused by injury.

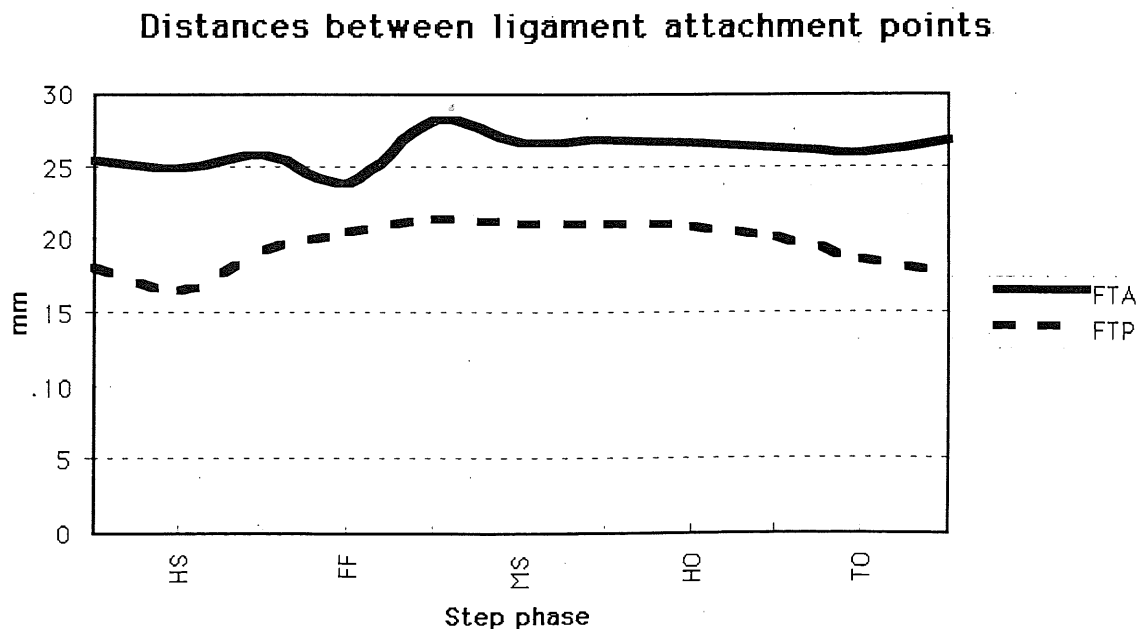


Fig. 1 Distances between ligament attachment points.

APPLICATION OF VERTICAL JUMP IN EVALUATING ATHLETES' MOTOR RECOVERY

R. RODANO¹, A. MINGRINO¹, R. SQUADRONE^{1,2}

¹Centro di Bioingegneria (Politecnico di Milano - Pro Juventute Fnd.), Milan, Italy

²Associazione Scuola Viva, Rome, Italy

INTRODUCTION

Some changes of the mechanical output of the musculo skeletal system are the inevitable consequence of injuries. A partial recovery can lead to relapses or to potentially dangerous compensatory strategies, especially during the practice of sport activities. This justify the need of specific methods to check the motor recovery of an athlete.

Aim of this paper is to present the results of a study designed to evaluate the lower limbs motor recovery of injured rugby players, by means of the mechanical joint output (net moments and powers) measured during bilateral vertical jumps.

METHOD

Five athletes (height = 1.78 m, SD = 0.03, body mass = 75 kg, SD = 3) who had suffered a unilateral ACL injury (INJ) within the 2-yr. period prior to the study, and eight uninjured (UNI) athletes (height = 1.80 m, SD = 0.05, body mass = 78 kg, SD = 7) were the subjects of this study. All the subjects are professional rugby players, playing in the Italian second league. After 20 min. of standard warm up, each subject performed 10 double-legged countermovement vertical jumps with the right foot on the force platform and 10 jumps with the left foot on the force platform. Kinematics data, concerning the spatial position of ten anatomical landmarks (five per leg), were recorded by means of an optoelectronic system (ELITE, BTS SrL) with a sampling rate of 100 Hz. Simultaneously, ground reaction forces were measured by a Kistler force platform and acquired with a sampling frequency of 1000 Hz. The internal joint centers, such as the corresponding moments and powers of the lower limb acting on the force platform, were estimated by using a special software which inputs were anthropometric, kinematic and kinetic data.

Peak joint moments and powers were used for the analysis together with the individual percentage asymmetry (ASY%) between right/left (for the UNI group) and healthy/ACL (for the INJ group) limbs calculated for each of the kinetic parameters.

RESULTS

Mean jumping height values were 0.421 m. (SD = 0.004) and 0.405 m. (SD = 0.005), respectively for the UNI and INJ group, with no significant group difference. For the three lower limbs joints, means and standard deviations for peak moments and powers, are presented in table I. In addition, are reported the percentage asymmetry between right/left and healthy/ACL limbs for the UNI and INJ groups, respectively.

Examining peak moments and powers values, the only significant group differences were found at the knee, with the ACL limb of the INJ group considerably lower compared to the homologues. With regard to the limb asymmetry, considering moments, the data reveal no significant differences between scores of the two groups. In contrast, examining powers, significant group differences were found at the knee, with the INJ group characterized by higher ASY values.

		UNI			INJ		
		Left	Right	ASY%	Healthy	ACL	ASY%
Moments [Nm]	Hip	104 (18)	114 (35)	13 (8)	112 (71)	129 (30)	9 (6)
	Knee	164 (44)	169 (47)	10 (6)	152 (71)	133 (72)	11 (6)
	Ankle	94 (17)	98 (14)	8 (6)	101 (20)	109 (15)	10 (7)
Powers [W]	Hip	394 (92)	414 (122)	10 (10)	477 (90)	482 (114)	13 (9)
	Knee	765 (70)	792 (112)	6 (5)	701 (90)	477 (124)	22 (7)
	Ankle	535 (176)	585 (185)	13 (7)	558 (108)	640 (101)	18 (9)

Table I

DISCUSSION

In summary, the kinetic analysis during vertical jump seems to provide useful information that can be applied to a clinical evaluation of the athletes during post ACL injury.

Indeed, differently from the results of Kannus, who found no significant limb differences between ACL injured knee and healthy homologues using an isokinetic dynamometer, the subjects of the INJ group had substantial changes from normal patterns and magnitude of the power at the knee joint. This adaptation could be interpreted as a subconscious desire to reduce the net quadriceps force required during jumping.

CLASSIFICATION OF SKI JUMP TAKE-OFF TECHNIQUES BY JOINT POWER

T. SASAKI, K. TSUNODA, E. UCHIDA* and H. HOSHINO**

Hokusei Jr. College, Sapporo, Japan, *Kokugakuin Jr. College, Takikawa City, Japan,

** Ebetu City Gymnasium, Ebetu City, Japan

INTRODUCTION

In ski jumping, take-off action is the most important factor for having ascent force. Ascent force acts on the ground and is required for a jumper to obtain initial velocity and angular momentum. Reaction force gives the body both translational and rotational momentum at take-off. In take-off technique, we ought to aim for optimum movements of the joints and body segments because reaction force is the result of the integrated kinetic parameters of each joint or segment. When jump action produces force, the body, which is affected by ground reaction force. Reaction force makes the jumper revolve. Jump action in laboratory tests was characterized by two techniques of contribution of joint power. The purpose of this study is to clarify the role that joint power and energy play in ski jumping for world class jumpers.

METHODOLOGY

The takeoff action of six jumpers was analyzed from video taken at World Cup competitions at Okura jump hill from 1995 to 1996. Camera speed was 240 frames per second (made by FOR'A Co.). We observed moment force and joint power resulting from actual ski jumps, using seven segment linkage models of seven degrees of freedom. The segments are foot, lower leg, thigh, trunk, head, upper arm and forearm segments. We calculated the torque, energy and power that were produced by each joint. The data from video were collected by computer. An inverse kinematics solution was applied to video analysis. The data analysis system was developed for a Macintosh computer.

RESULTS

Most of the power, from initial action until take-off, is produced by two joints, the hip and the knee joint. The knee joints were the major energy and power producers in all jumpers. Power production in joints was generated in regular order for almost jumpers.

Jump performances are classified into three techniques according to joint power production. The first technique showed the tendency that the hip joint and after some delay the knee joint produced power. The values are positive until maximum power output. This technique is named the "Phase Shift Power Generation Technique". The second technique had the same tendency of power generation as the previous but the delay between maximum knee joint and hip joint power was longer. This technique included a remarkable action which was the forward rotation of the leg. This technique is named the "Phase Shift and Leg Rotation Technique". The last technique had a tendency in which both hip and knee joint powers increased at almost the same time and the values were positive until maximum. This technique is named the "Close Power Generation Technique".

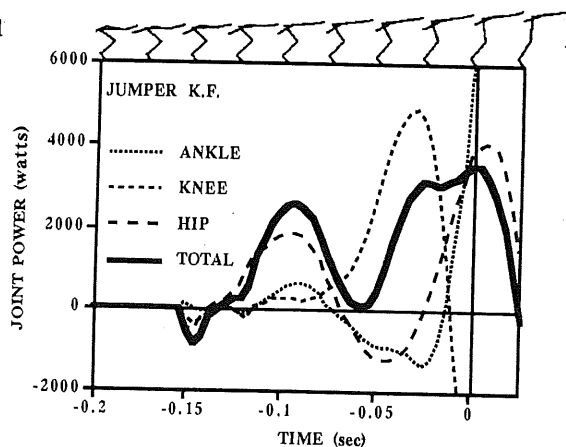


Fig. Phase Shift and Leg Rotation Technique.

HOW TO GAIN SPEED RUNNING OVER WAVES IN ALPINE SKIING

H.-B. Schmiedmayer, P. Lugner

Institut für Mechanik, Technische Universität Wien, Wien, Austria

INTRODUCTION

In downhill events of alpine skiing competitors try to minimize time for a given course. The result depends mainly on the skill of the skier. In this paper it will be shown that this is not only true for turning situations but also when running straight over a wavy surface.

METHOD

In a first step a single mass model is set up, where the ground is represented by a cyclic function $y = f(x) = f(x + \lambda)$ with wavelength λ guiding the reference point G (Fig. 1). The position of the center of mass M of the skier with respect to the ski can be described by the height $h(x)$ and lean angle $\beta(x)$. The moment of inertia $I_M = m \cdot i^2$ is assumed to be constant for the current model. Effects of the frictional drag between ski and snow as well as air resistance are not included.

Using Lagrangian equations of the second kind to the following equation of motion is obtained:

$$\ddot{x} \cdot a(x) + \dot{x}^2 \cdot b(x) = c(x), \quad (1)$$

where $a(x)$, $b(x)$, and $c(x)$ are functions of $h(x)$, $\beta(x)$ and $f(x)$ and their first, second, and third derivatives.

To show the principal effects the following functions were chosen:

$$f(x) = A \cdot \sin\left(x \frac{2\pi}{\lambda}\right), \quad h(x) = h_0 + A_h \cdot \sin\left(x \frac{2\pi}{\lambda} + \varphi_h\right), \quad \beta(x) = A_\beta \cdot \sin\left(x \frac{2\pi}{\lambda} + \varphi_\beta\right). \quad (2)$$

Equation (1) is solved numerically.

RESULTS

All results are compared to the so called 'stiff' skier with $A_h = A_\beta = 0$. The quality of increase in speed and decrease in time for one period depends mainly on the phase-angles φ_h and φ_β whereas the quantity of these effects depends mainly on the amplitudes A_h and A_β . Therefore the following results were calculated with the following parameters:

$$h_0 = 0.9\text{m}; \quad \dot{x}_0 = 10\text{m/s}. \quad (3)$$

The results for the stiff skier ($A_h = 0$, $A_\beta = 0$) as well as for optimal up-down movement only ($A_h = 0.2\text{m}$, $A_\beta = 0$) and for optimal forth-back movement only ($A_h = 0$, $A_\beta = 10^\circ$) are shown in figure 2.

DISCUSSION

The simple model shows that an increase of the speed at the end of a full wave with respect to the speed at the beginning of the wave can be obtained mainly by the up and down movement $h(x)$. On the other hand the back and forth movement $\beta(x)$ results mainly in shorter times for passing a whole wave without considerably increase in speed with respect to a normal run ($h(x) = \text{const}$ and $\beta(x) = 0$). Both motions can be combined to give an optimal motion pattern for overall increase in speed and shorter passing times. Furthermore it is assumed that similar mechanisms influence the speed and time while cornering.

For a more detailed analysis effects frictional drag between ski and snow as well as air resistance should be included in this model to find optimal techniques for running over wavy ground in downhill racing.

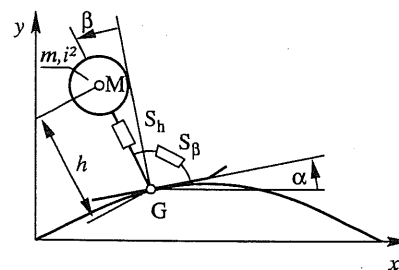


Figure 1: Simple single mass-model for the skier on a wavy ground with two positioning elements S_h and S_β for the height h of the center of mass M and the lean-angle β respectively.

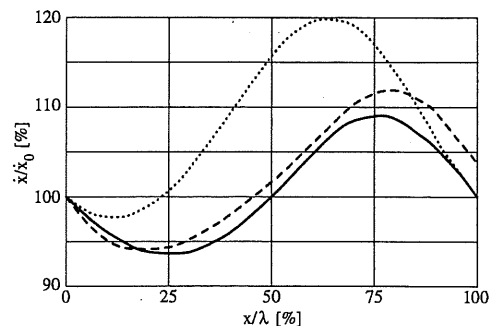


Figure 2: Horizontal speed-factor $\dot{x}/\dot{x}_0$ of point G as a function of x for the stiff skier (solid line) and for the optimal up-down movement only with phase angle $\varphi_h = 0.5\pi$ (dashed line) and for optimal forth-back movement only with phase angle $\varphi_\beta = 0.5\pi$ (dotted line).

SIMULATED AND ACTUAL SKI JUMPING TAKE-OFFS DIFFER IN PLANTAR PRESSURE AND MUSCLE ACTIVATION PATTERNS

M. Virmaavirta, P.V. Komi and J. Perttunen

Department of Biology of Physical Activity, University of Jyväskylä, Finland

INTRODUCTION

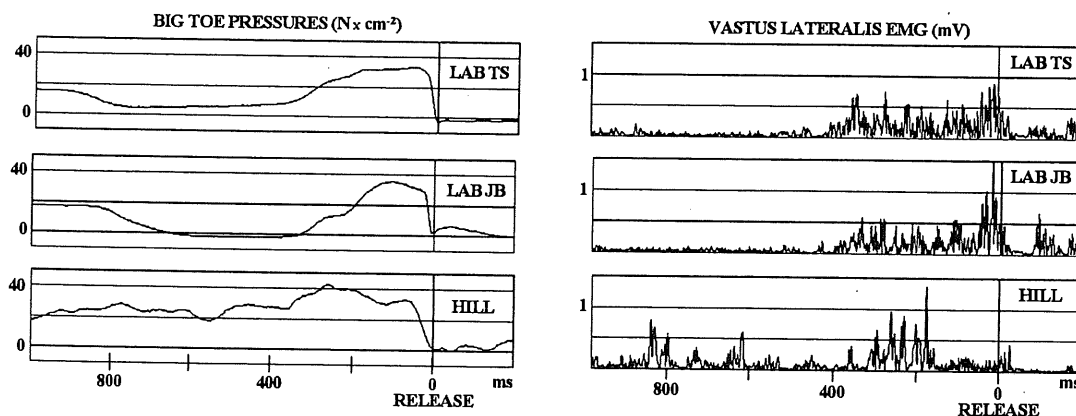
Timing of powerful take-off is important for successful ski jumping performance (Komi & Virmaavirta 1997). Simulated take-offs in laboratory or in training have often been used to mimic actual ski jumping take-off. The present study makes an effort to characterize how similar these take-offs are with regard to plantar pressure and muscle activation patterns.

METHODS

Ski jumping take-offs of world class ski jumper were studied in three different conditions: simulated take-offs in laboratory (LAB)(1) with training shoes (TS) and (2) with jumping boots (JB) as well as in (3) actual jumping hill conditions (HILL). Muscle activities and plantar pressure distribution were recorded by Paromed Datalogger[®] system with the same electrode and insole set-up in all three conditions. Two components (Fz and Fy) of the ground reaction forces were also recorded for registration of the performance level in LAB measurements.

RESULTS AND DISCUSSION

Decrease in the resultant take-off velocity in LAB JB (3.24 ± 0.09) compared to LAB TS (3.36 ± 0.12) can most probably be explained by less effective utilization of plantar flexor muscles with stiff ski boots (revealed by decrease of 10 % in EMG of the gastrocnemius (GA) during the push-off phase). Even less effective use of GA (decrease of 46 % in EMG) during the same phase in HILL conditions emphasize the action of knee extensor muscles in gaining vertical lift during the take-off. An example of the activation patterns of the vastus lateralis (VL) muscle as well as the plantar pressure in different conditions can be seen in the figure. The activation of the VL in HILL conditions is greatest during the early phase of the take-off corresponding to the peak of the measured pressure. The highest pressure during the take-off was measured under the medial side of the fore foot implying that while the jumper's center of mass moves forward the take-off action has a strong lateral component from both feet.



The results of this study demonstrate that in order to perform an effective take-off in actual conditions the jumper must start the preparation early during the inrun curve before the flat part of the apron. In LAB conditions the preparation for the take-off began already approximately 770 ms before the release instant corresponding to 18 meters at the speed used in HILL condition. However, the release instants in LAB and HILL conditions do not match exactly, because at the release instant of the HILL the knee extension is not completed.

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THE DEVELOPMENT OF SPRINT TRAINING MACHINE

Kando KOBAYASHI, Masaaki SUGITA, and Noriko MATSUGAKI
Dept. of Life Sciences (Sports Sciences), Graduate School of Arts and Sciences,
University of Tokyo
3-8-1 Komaba, Meguro-ku, Tokyo 153 Japan

Purpose

Biomechanical research results based on a motion analysis of top sprinters of the world (such as Carl Lewis) confirmed that it is necessary to strengthen the extensor muscle group at the hip joint in order to quickly move thighs around the hip joint as the pivot. Although the importance of training of those extensor muscles has been pointed out, there has not been a conclusive method for strengthening the muscle groups. The authors intended to develop new type of training machine which strengthen the extensor muscle group at the hip joint and improve running performance for many runners.

Concept of Invention

The invention of new training machine "Sprint Training Machine" was related to the concept as follows;

- 1) To effectively strengthen the extensor muscle group at the hip joint (especially the hamstring muscles) are necessary for an ideal running motion through an exercising form which is similar to the actual running motion.
- 2) To make a user learn a leg movement pattern which create such running motion that is considered to be ideal for running fast by having person repeatedly trace a certain locus of movement on the training machine.
- 3) To correct an imbalance between right and left sides of the form which occurs during a natural walking and running, thereby helping the prevention of injury and to provide an exercise stimulation which is effective for improvement of the health. Accordingly, the new machine was developed based on the idea that ideal movement of legs should be constituted by the combining motions which are based on a circular motion like the theory of circular wheel movement and reciprocating movement of the axis of circular wheel.

Composition of "Sprint Training Machine"

A pair of forward and backward movable pedestals respectively provided with a pedal, an arm and a crank and operable independently from each other; a power engine for shifting the movable pedestals backward and forward by a certain distance which is adjustable within a 1 m at a certain reciprocating speed which is also adjustable. A power transmitting portion; A load controlling means for controlling a resistive load during a pedaling exercise such that a resistive load is applied to a user when the pedal is located lower than a horizontal position of the pedal arm while no resistive load is applied to a user when the pedal is located higher than the horizontal position of the pedal arm.

A hand-rail for supporting the operator to balance themselves during the pedaling exercise wherein a weight of the operator is supported by the right and left pedals; wherein a sprint training machine being constituted such that a resistive load is applied to the operator when the pedaling exercise is conducted at a position lower than the horizontal position of the pedal arm and that no resistive load is applied to the operator when the pedaling exercise is conducted at a position higher than the horizontal position of the pedal arm, while a rotation axis of the pedal arm is shifted back and forth by a certain distance by means of the power engine.

Effect of training using "Sprint Training Machine"

1. Effect of interactively learning the ideal locus of movement in running motion. Although it is possible to imaginatively understand that the foot portion should move along a circular orbit just like a wheel in a model movement for most efficient, ideal running motion, it has been impossible to actually experience or learn such a movement. However, the present invention has made it possible to learn the ideal model movement in which the foot portion moves along a circular orbit.

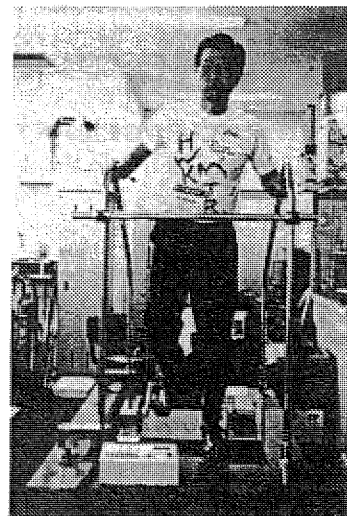
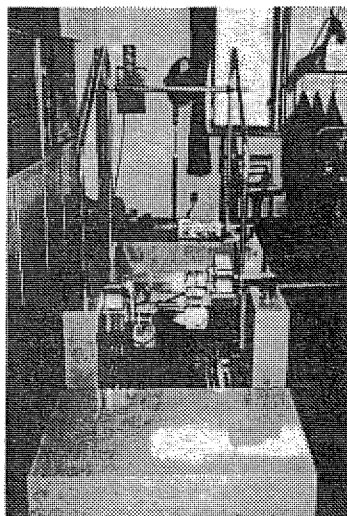
All the athletes (sprinters and marathon runners) who did training with use of the sprint training machine according to the present invention has experienced remarkable correction of respective running forms immediately after the training and their records have also improved.

2. Muscle strength training effect which directly leads to improve of sprint ability.

3. Effect of correcting imbalance between right and left sides of running form.

4. Effect of learning relaxation in running motion.

5. Effect of improving suppleness of body.



AN ADAPTABLE RISK-FREE STRENGTH TRAINING DEVICE WITHOUT PHYSICAL CONSTRAINTS

Yoky Matsuoka¹ and Larry C. Miller²

The Massachusetts Institute of Technology, Cambridge, MA.¹

Boston Biomotion Inc., Wellesley, MA.²

INTRODUCTION

Recently, the importance of strength training is emphasized for increasing athletic performance, for maintaining health, and even for prolonged life. To fit these needs, many linear and rotary resistance devices became commercially available. However, human's athletic movements cannot be matched well on their constraints which often results in inefficient training or even in injury.

In order to overcome these problems, we have constructed a three-dimensional resistance trainer which is adaptable to the exerciser's kinematics and needs.

METHODS

Our machine is a completely passive, spatial manipulator. It has six degrees of freedom in a workspace that is a half sphere approximately 1.5 meters radius. There is a gimbal wrist with three non-actuated degree of freedom and it can be used with a hand or a foot. The three positional degrees of freedom are actuated with brakes and accomplished through serial rotary and linear joints that are cable driven. This machine is controlled by a motion controller with DSP and programmed incorporating interface.

RESULTS

Using our novel strength training device, any kind of strength training is possible. For example, in a curling movement, the bicep muscle can exert about 70% of its potential when the arm is straight (180 degrees), 100% of its strength at 100 degrees and 70% again at 60 degrees. With the machine, the force field can be created to match the kinematics of the user and the strength to achieve the maximum efficiency without injury. Also by keeping track of time, the applied force can be adjusted to be stronger as the training progresses.

At the appearance of pain or fatigue, the user can instantaneously decrease the damping on the fly or stop the motion. Because the resistance is not a weight that can be dropped, the user experiences near zero loading when the motion is stopped. This environment can also be created with a machine which is actuated by motors. However, during an extreme case which the machine fails, it cannot be guaranteed that the motored machine will be stable or harmless to the user.

DISCUSSION

A unique benefit of our machine concept is that the exerciser is freed from the line of action of the force constraint present at the machine/user interface in all current forms of weight or resistance training. Human force production in such activities is highly constrained because of the need to reconcile body position, joint axes, and leverage to the line of action of the force. With our device, the user has complete control over the force direction and magnitude with the end-point force-velocity collinearity. Research has shown that purposeful motion is degraded without force-velocity collinearity and this is easily demonstrated. As an additional benefit, the machine can chart changes in force, position, velocity, acceleration, power, work, and range of motion in real time. The user can have an access to these accurate data as he/she trains.

The device we constructed provides a safe and efficient strength training environment with real-life ranges of motion with appropriate modes and increments of resistance operating under the complete volition of the user. With this machine, we believe that we can take the strength training to the next level.

ACKNOWLEDGEMENT

We would like to thank all Barrett Technology Inc. employees for their contributions.

EVALUATION OF CUSHIONING PROPERTIES OF RUNNING FOOTWEAR

D.G.E. Robertson¹, J. Hamill², D. A. Winter³

¹University of Ottawa, Ottawa, ON, Canada; ²University of Massachusetts, Amherst, MA, USA;

³University of Waterloo, Waterloo, ON, Canada

INTRODUCTION

The cushioning properties of athletic footwear have been evaluated *in vivo* using either ground reaction force or accelerometry data or *in vitro* with impactors. Unfortunately, none of these methods directly measure the amount of cushioning that a runner experiences. An alternative method for the evaluation of cushioning may involve the use of the deformation power in the foot. **Deformation power** (P_{def}) in the foot may be defined as:

$$P_{def} = dE_{foot}/dt - (F_{ankle} \cdot v_{ankle} + M_{ankle} \omega_{foot})$$

where $F_{ankle} \cdot v_{ankle}$ is the joint force power and $M_{ankle} \omega_{foot}$ is the power produced by the ankle moment of force.

Previously it was reported that P_{def} (also called power imbalance) was a result of the elastic deformation of the foot (Robertson & Winter, 1980; Seigel, Kepple & Caldwell, 1996; Winter, 1996). This power source, especially during the weight-acceptance phase of support, is due primarily to deformation of the shoe midsole and tissues in the foot. The purpose of this study was to determine if running footwear with differing cushioning properties could be distinguished by the amount of deformation power.

METHODS

Five healthy people served as subjects in this experiment after giving informed consent. Each subject performed five trials at 16 km/h (6 min/mile) in each of two footwear conditions: soft midsole (40-43 Shore C durometer) and hard midsole (70-73 Shore C durometer). In all respects the footwear were identical excepting for their midsole durometer measures. Subjects ran across a force platform from which data were collected at 1 kHz while being videotaped from a sagittal view with a 200-Hz video camera. The rate of change of segmental foot energy (dE_{foot}/dt) was calculated from the video data. Inverse dynamics methods were used to calculate ankle forces, moments and powers. Deformation powers (P_{def}) were then computed using the equation presented above.

RESULTS AND DISCUSSION

The P_{def} was characterized by negative work for approximately the first two thirds of the support period due to energy absorption by the foot or the midsole of the shoe. In the final third of the support period, there was a small positive burst of power that supplied energy to the foot either due to energy return from the midsole or more likely by work done at the metatarsal-phalangeal joint. Table 1 presents the mean peak P_{def} and the negative and positive work done during support for all subjects in both the hard and soft midsole conditions. The P_{def} for the hard midsole shoe produced greater peak powers and absorbed more energy than the soft midsole shoe. Figure 1 presents the five-trial ensemble for both soft and hard footwear conditions for an exemplar subject, however, there were considerable variations among subjects. It appears that the differences in the peak P_{def} may be influenced by body mass. In the lighter subjects the differences in the peak negative P_{def} were not as great as with the heavier subjects. This may be because the heavier subjects deformed the midsole to a greater extent. In conclusion, this technique differentiates between footwear with differing cushioning properties and offers a means of detecting the effects of shoe and foot cushioning on runners.

Table 1: Mean deformation power and work.

	Soft	Hard
Peak $-P_{def}$ (W)	-1211	-1369
Neg. work _{def} (J)	-440	-619
Peak $+P_{def}$ (W)	162	132
Pos. work _{def} (J)	131	64

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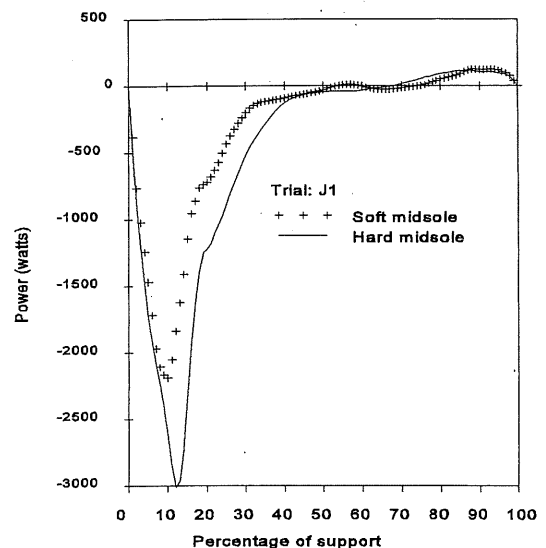


Figure 1. Ensemble averaged deformation power

PROSTHETIC FOOT AS SIMULATOR FOR ATHLETIC SHOE TESTING

M. R. Pitkin

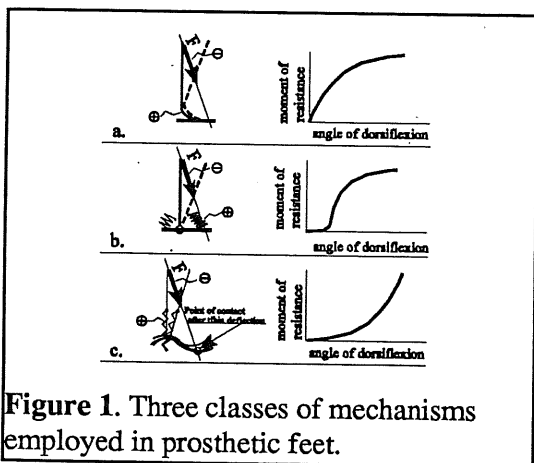
Tufts University School of Medicine, Boston, USA

INTRODUCTION

Simulation of the human foot response to loads within the athletic shoe is needed for variety of testing protocols. It seems reasonable to look at prosthetic feet since they were designed to mimic not only dimensions but the functions of the anatomical foot. A classification of mechanisms of prosthetic feet available was done in this study. Analysis of three typical classes of mechanisms was conducted which suggested the Rolling Joint Foot and Ankle to be an adequate device for the shoe testing.

METHODS

There were many attempts to mimic the performance of the lost foot and ankle by a passive elastic system "charge-release," where "charge" corresponds to dorsiflexion and "release" corresponds to the following plantar flexion. One can say that the level of approximation of the prosthetic foot/ankle performance to those demonstrated by the real foot/ankle, depends on how well a "charge-release" mechanism is designed. The power released and the resistive curve, are two typical characteristics of any "charge-release" mechanism. Both characteristics can be used as targets in a synthesis of a new mechanism. A power released in plantar flexion period in real ankle, was chosen as a target in designing of the most contemporary energy storing (ES) feet (Gitter et al., 1991). While that "release" goal was achieved for plantar flexion, not enough attention has been paid to the "charge" component, namely, to the resistive curve in dorsiflexion period. Indeed, in all ES feet on the market, the charge during dorsiflexion is provided by



activation of bending or compression of elastic elements. The analysis shows that these types of mechanisms have non-concave (convex or linear) pattern (Fig.1,a,b), opposite to the concave pattern seen in the real ankle (c). This is also true for structures like Genesis and SACH feet, employing compression and/or extension. A Rolling Joint Foot (RJF) (Pitkin, 1996) was found to be adequate to that request, since as biomechanical study showed, a prosthetic gait synergy has been transferred to a more normal one.

DESIGN

The RJF comprises a higher-pair ankle and metatarsal joints. The joints provides rolling of curved contacting surface along each other with elastic resistance to deflection. Moving components are connected by linear springs of

extension type. Moments of resistance in both joints have a desired concave pattern seen in the anatomical joints (Fig.1,c) provided by the nature of cycloidal motion.

RESULTS

1. Theoretical analysis of the resistance to deflection of variety of prosthetic feet showed that higher compliance of RJF, makes it adequate to the anatomical foot performance.
2. It is suggested to include RJF in a testing system "shoe - artificial foot."

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ACKNOWLEDGEMENTS

The study was supported by the Ohio Willow Wood Company and NIH Grant 1R43 A R43290-01A1.

A METHOD FOR THE STUDY OF THE DISTANCE BETWEEN ANKLE LIGAMENT INSERTIONS IN VIVO.

A. LUNDBERG, I.G. WINSON, T. HASHIMOTO, C. SARO, P. WESTBLAD

From the Dept of Orthopaedics, Karolinska Institute at Huddinge Hospital, Stockholm, Sweden.

We describe a method of analysing changes in the distances between the attachment points of the ligaments around the ankle during a variety of normal activities while simultaneously assessing joint kinematics.

MATERIALS AND METHODS

Three healthy volunteers had 2.4mm Hoffman external fixateur pins inserted into the tibia, fibula, talus and the calcaneum using local anaesthetic. Four 10mm reflective markers were attached securely to each pin. Pin placement was such that no interference was seen between pins, markers or normal anatomical structures. All subjects were pain free during the examination and could walk without a limp during the studies.

Following the insertion of the pins and attachment of the markers two radiographs were undertaken using the roentogen stereophotogrammetric method. The anatomical insertions of the ligaments and the reflective markers were subsequently digitised to provide the spatial relationship between the positions of the ligament insertions and the attached markers. Analysis of motion was then performed during normal walking using an three camera optoelectronic method of analysis (MacReflex) at 50 Hz, and the ligament attachment points were transferred to each frame of the gait file assuming an unaltered spatial relationship between each ligament attachment and the marker set of the corresponding segment. The distance between the attachments of each ligament was then calculated frame by frame.

RESULTS

Fig. 1 shows an example of calculated distances between the attachments of the anterior fibulotalar (FTA) and posterior fibulotalar (FTP) ligaments in one subject during the stance phase of walking. Differences during this movement were small in this example.

CONCLUSION

This technique may prove to be a useful way of analysing the change in ligament characteristics under various life conditions including normal activities in normal individuals and pathological changes caused by injury.

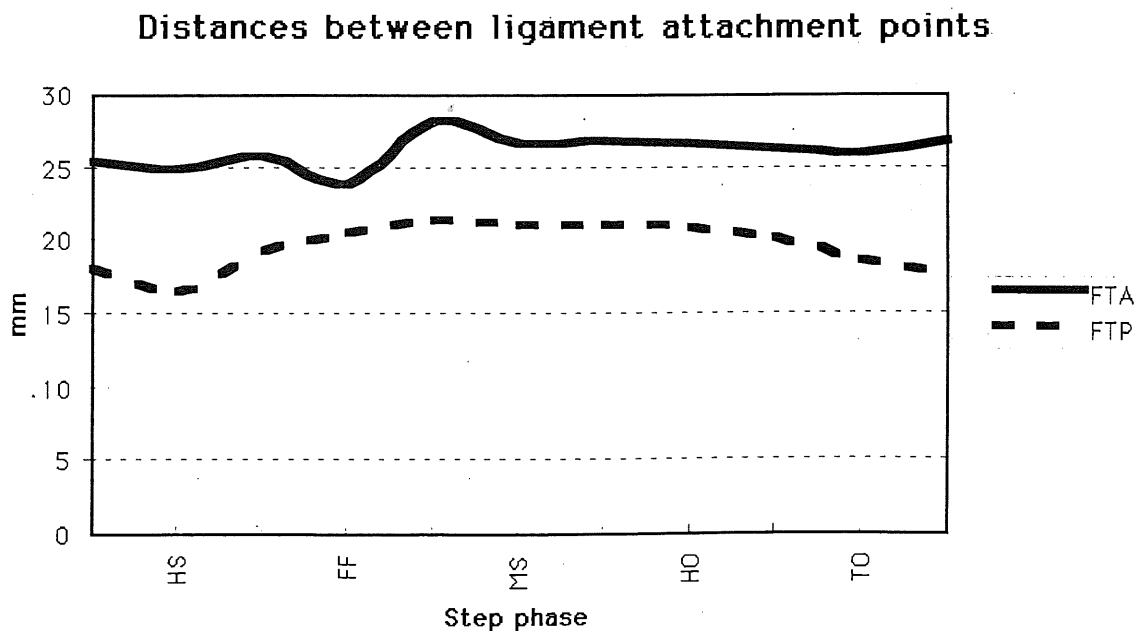


Fig. 1 Distances between ligament attachment points.

MEASURING SYSTEM FOR IN VIVO RECORDING OF FORCE SYSTEMS IN ORTHODONTIC TREATMENT - TECHNICAL IMPLEMENTATION AND EXPERIMENTAL TESTS

D. Friedrich¹, G. Rau¹, P. Diedrich²

¹ Helmholtz Institute for Biomedical Engineering, University of Technology, Aachen, Germany

² Department of Orthodontics, Faculty of Medicine, University of Technology, Aachen, Germany

INTRODUCTION

The cause of tooth movement in orthodontic treatment with fixed appliances is the force system, which is generated by an elastic archwire and transmitted by a *bracket* to the respective tooth. Since such an appliance is statically a multiply undetermined system, determination of the orthodontic forces and torques and, thus, application of a non-critical force level, is very problematic. In practice, force application is predominantly empirical-intuitive on the basis of recommendations developed in vitro (Bourauel *et al.*, 1992) and in theory (Koenig and Burstone, 1989; Miyakawa *et al.*, 1985). Aim of our approach is in vivo measurement of the forces and torques therapeutically applied to the individual tooth. The recommendations used to date for the appliance of forces have to be examined in clinical studies.

METHODS

The force systems generated by the archwire and acting on the teeth are isolated from the respective tooth by means of divisible, purpose-designed brackets and diverted into a force-torque sensor, which is fixed in this position relative to the patient's dental arch by a specially designed fixing device (Rosarius *et al.*, 1996). A precision coupling element integrated into the bracket permits the archwire and tooth to be completely detached during the measuring operation. The sensor is connected to this purpose-designed bracket by means of a gripping appliance. Accurate spatial fixation of the archwire in the therapeutical position is crucial for the measuring precision of the system, since smallest movements affect the force system to be measured. They are caused on the one hand by elasticities of the measuring system, and on the other hand by tensioning after the connection of the sensor, i. e. by the deviation of the neutral location of the system from the therapeutical position. The specially designed device ensures precise fixation of the sensor. The patient's head is positioned relative to the system by means of a special appliance. This comprises a glabella device as well as a forehead and chin support. Electrical measurement of the mechanical quantities is carried out by a 6-axis force-torque sensor with semiconductor strain gauge elements, an electronical evaluator and a mobile measuring computer (PC).

RESULTS

Extensive calibration reproducibly showed that the error due to electrical measurement of the mechanical quantities is less than 2%. The movements of the in vivo system caused by elasticities and tensioning were recorded in three dimensions by means of a laser distance sensor (Fig. 1). The abscissa-segment of the characteristic lines represents the movement of the neutral location due to tensions; the gradient corresponds to the stiffness. In addition, the results of experimental in vitro quantification of measuring accuracy by means of a purpose-designed dental arch model are shown. Based on these results, initial in vivo studies are discussed and assessed.

DISCUSSION

The in vitro tests showed the suitability of the measuring system for the intended application since high reproducibility and only a small error could be demonstrated. Movements of the measuring system up to 0.06 mm result from a therapeutical force of 1.5 N, an expected maximum in a worst case situation. The resulting error is analyzed as a function of the relevant parameters.

Acknowledgements - This research is supported by the *Deutsche Forschungsgemeinschaft* (ref.: Di 595/1-2).

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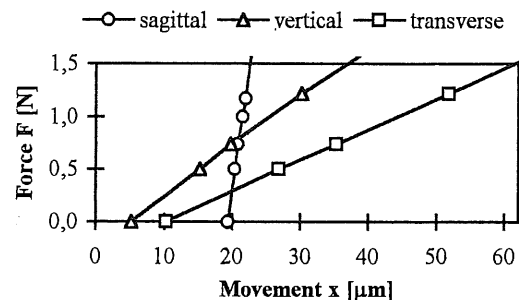


Fig. 1: Movements of the measuring system due to tensions and elasticities. The worst case with cumulative effects is shown.

INTERSEGMENT MOMENTS OF THE LOWER EXTREMITY JOINTS IN RUNNING - INFLUENCE OF INERTIA

B. Krabbe, R. Farkas, W. Baumann
Institute of Biomechanics, German Sport University Cologne, Germany

INTRODUCTION

The input parameters for the determination of muscle and joint forces are the intersegment moments of the joints of the lower extremity. From the mechanical point of view, the resultant intersegment moments consist of different contributions. During the contact phases of gait or running the intersegment moments of the joints of the lower extremity are often calculated only as the moments dependent on the ground reaction force (GRF). Even during touch down and take off of a dynamic motion like running, the importance of inertia is not well understood in all joints of the lower extremity, especially if a three-dimensional approach is employed.

THEORY

The interesting moment M_G of all muscles and other biological structures (ligaments, joint area forces etc.) of the mentioned joint has to compensate the intersegment moment $M_{\bar{G}}$ (see Hof, 1992) that can be defined as:

$$M_{\bar{G}} = -M_G = M_B + M_R + \sum_{i=1}^{No.Seg.} M_{m_i} - \sum_{i=1}^{No.Seg.} M_{T_i} - \sum_{i=1}^{No.Seg.} M_{C_i} \quad (1)$$

where M_B is the moment of the ground reaction force, M_R the free frictional moment in the transversal plane, M_{m_i} the moment depending on the weight of a segment, M_{T_i} the moment of inertia force (with respect to the proximal joint), M_{C_i} the rate of change of angular momentum.

EXPERIMENTAL METHODS

The experimental data were taken from a complex biomechanical analysis with 10 elite Marathon runners; mean aged 33,4 (SD=10,3 years), of national class (Marathon time 2:30, SD=0:07 h). With the help of an opto-electronic Selspot II-system (Selcom AB, Partille, Sweden), the position of interesting joint points was calculated from skin markers. Information about the Ground Reaction Forces and their point of application were sampled with a Kistler force plate (type Z 4852/b, Kistler Instruments, Winterthur, Swiss) invisibly installed inside the running plane of 20 m length. The synchronised measuring frequency of both systems was 500 Hz.

RESULTS AND DISCUSSION

The special results illustrated in this section are based on one typical runner (heel striker) of the above mentioned group. The inertia dependent parts of the intersegment moments are dependent on the acceleration of the segments centre of gravity as well as on the angular velocity of the segments. The accelerations of the centres of gravity, calculated from noisy landmarks (SELSPOT Leds), are the 2nd time derivatives of the centres of gravity. Both, the course, as well as the maximum of the acceleration depend on the way of smoothing the raw data from the numerical point of view (Woltring, 1986). In this investigation, a cubic spline of 3rd order was investigated.

The amount of the rate of change of angular momentum with respect to the mentioned joint axis (this is the sum of the moments of inertia force and the rate of change of angular momentum with respect to the segment's centre of gravity) is neglectable for the ankle, 40 Nm for the knee and 170 Nm for the hip joint. If the intersegment moments are the interesting results of a biomechanical analysis it seems to be necessary to calculate the inertial parts of the intersegment moments for the hip. The calculation of the intersegment moments of the knee needs this contribution, if the touch down part is of special interest. For the ankle, a quasistatic calculation of the intersegment moment is sufficient.

It can be stated that it is not necessary to calculate the rate of change of angular momentum (with respect to the segment's centre of gravity) during the stance phase of motions up to a horizontal velocity of 5 m/sec because of the small contribution to the intersegment moment as well as the complex algorithm from the numerical point of view.

CONCLUSIONS

It can be said that the calculation of the moment of inertia force is necessary for the calculation of the intersegment moments of the hip and - if the touch down phase is of special interest - for the knee in opposite to the ankle. The rate of change of angular momentum with respect to the segment's centre of gravity can be neglected for all joints of the lower extremity.

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STRATEGY OF MUSCLE COORDINATION IN A MULTI-JOINT DYNAMIC TASK

B. I. Prilutsky^{1,3}, T. Isaka², A. Albrecht¹, M. M. Ryan¹ and R. J. Gregor¹

¹Department of Health and Performance Sciences, Georgia Institute of Technology, Atlanta GA, USA

²College of Science and Engineering, Ritsumeikan University, Kusatsu, Shiga, Japan

³Biomechanics Laboratory, Central Institute of Physical Culture, Moscow, Russia

INTRODUCTION

It has been demonstrated that in static and dynamic, multi-joint contact tasks performed by the legs, two relationships with a strong correlation exist (i) between the difference in electromyographic activity (EMG) of rectus femoris (RF) and hamstrings (HA) and the difference in knee and hip resultant moments, and (ii) between gastrocnemius (GA) EMG and the difference in ankle and knee moments (Jacobs and van Ingen Schenau, 1992; Toussaint et al., 1992). The fact that these relationships occur in all subjects is striking because the desired joint moments can be theoretically produced by an infinite number of possible strategies of muscle activation. The reasons why the observed strategy of muscle coordination is chosen by the nervous system are not understood. In the simulation study of static, contact tasks, we recently showed that the above EMG/moment relationships may be a consequence of a strategy of muscle coordination that minimizes muscle fatigue (Prilutsky and Gregor, 1997). The purpose of this study was to examine if the strategy of muscle coordination during a dynamic task (load lifting) is consistent with the strategy that minimizes muscle fatigue.

METHODS

Subjects lifted barbell weights using a back lift technique (Toussaint et al., 1992). The task was performed at three speeds -- preferred or normal (movement time was about 1 s), slow (about 1.8 s), and fast (about 0.8 s), and two lifting loads -- about 10 and 20 kg. Recorded ground reaction forces and coordinates of the main leg joints were used to calculate the resultant joint moments. Surface EMG of major leg muscles including RF, HA, and GA were also recorded. The EMG was rectified and low-pass filtered, and then normalized with respect to the peak filtered EMG recorded in an 85%-maximum snatch. The normalized EMG was shifted in time to account for an electromechanical delay between the EMG and joint moments. A 2D, musculoskeletal model of the leg (four segments and 9 muscles), the optimization criterion of Crowninshield & Brand (1981), and the optimization toolbox of MATLAB (The MathWorks Inc.) were used to estimate muscle forces.

RESULTS AND DISCUSSION

The obtained joint moments, patterns of EMG, and estimated muscle forces were consistent among the subjects. In particular, all subjects demonstrated a strong linear relationship between the difference in EMG/force of RF and HA and the difference in knee and hip moments, and between GA EMG/force and the difference in ankle and knee moments (Fig. 1). There were no significant differences in these relationships (in terms of correlation or regression coefficients) between the task conditions (load, speed). The estimated magnitude and rate of muscle length changes during the task allowed us to assume that their influence on the relationship between EMG and muscle force was small. With this assumption, we speculate that the observed strategy of muscle activation (Fig. 1, A, C) corresponds to the strategy of muscle coordination (Fig. 1, B, D) that minimizes muscle fatigue (Crowninshield and Brand, 1981). This strategy of muscle coordination is characterized by allocation of relatively more force to muscles with a long moment arm and large physiological cross-sectional area, and by a synergistic muscle action. Based on the results of this and our previous studies, we propose that an important role of two-joint muscles in multi-joint tasks is to reduce the total muscle force, stress, and fatigue by acting as agonists at both joints they cross.

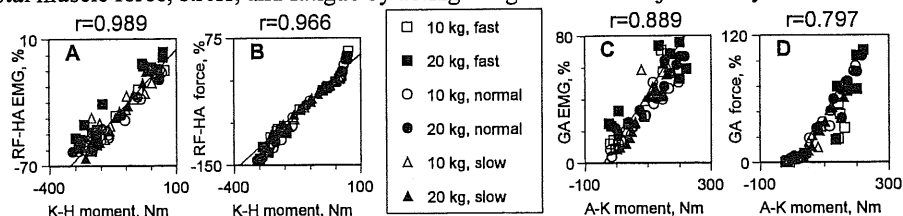


Figure 1. Difference in normalized EMG (A) and force (B) of RF and HA as a function of difference in knee and hip moment, and difference in normalized EMG (C) and force (D) of GA as a function of difference in ankle and knee moment. Data of one subject and different task conditions. For each task condition, 10 data points taken each 10% of movement time are presented.

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EFFECTS OF MUSCLE CONTRACTIONS ON BEAT-BY-BEAT MEASURED HEMODYNAMIC PARAMETERS AND CARDIAC-LOCOMOTOR COUPLING DURING CYCLING

T. MORITANI, Y. NISHIJIMA and H. UE

Laboratory of Applied Physiology, Graduate School of Human and Environmental studies, Kyoto University, Japan, 606

INTRODUCTION

The pressure within exercising skeletal muscle rises and falls rhythmically during normal human locomotion. Recent findings regarding the entrainment of cardiac and locomotor rhythms have suggested that synchronization of these rhythms might be related to certain physiological functions, such as optimization of blood flow supply to exercising muscles and minimization of the energy costs of cardiac work (Kirby et al., 1989). We speculated that if the peak intra-arterial pressure due to cardiac contraction occurs at the lowest phase of the intramuscular pressure cycle, cardiac load would be minimized and diastolic perfusion of subendocardial muscle would be improved. Also venous return to the heart might be enhanced if the push from the skeletal muscle pump were temporally coordinated with the atrial repolarization with a maximal utilization of venous return.

METHODS

To test the hypothesis, precise phase relationship between the intramuscular pressure of the leg muscles as estimated by the pedaling force and beat-by-beat calculated hemodynamic parameters were obtained by means of computer-aided on-line analysis system (Moritani, 1994). In the first experiment (FREE cycling), 20 healthy male students cycled ergometer at their preferred pace for 10 min, at the intensity of 80-90% of their anaerobic threshold. The time lag between the peak of pedaling force (PF) and the ECG R wave (R) was calculated on the beat-by-beat basis. To further verify the interaction between these hemodynamic parameters and the timing of muscle contractions, in the second experiment, subjects were instructed to pedal at prescribed phases of the ECG RR intervals.

RESULTS

Total data points of 10,910 for beat-by-beat measured hemodynamic parameters were grouped into 10 different time-normalized phases. The results showed that when the peak force signal was in coincidence with the atrial contraction phase of the cardiac cycle, stroke volume of the following beat decreased ($p < 0.01$), indicating less effective venous return by the muscle pump acting against the already closed atria. Stroke volume showed a marked decrease when the muscle contraction was applied during the phases of the heart systole ($p < 0.01$). Similarly, systolic blood pressure and TTI (tension time index, as a measure of myocardial oxygen requirement) increased when the muscle contraction was applied shortly after the ventricular systolic phase ($p < 0.01$). When the muscle contraction was applied within the 0-10% of the RR-interval (ventricular ejection phase), stroke volume significantly decreased, and systolic blood pressure ($p < 0.05$) and TTI ($p < 0.01$) increased compared with that of FREE cycling ($p < 0.05$).

DISCUSSION

Our findings suggested that the alterations in venous return produced by the muscle pump and the changes in total peripheral resistance produced by internal muscle pressure cause a significant change of hemodynamic parameters. Particularly, when the peak force signal was in coincidence with the atrial systolic phase of the cardiac cycle, stroke volume of the following beat decreased, indicating a lower efficiency of the muscle pump.

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COORDINATION IN CYCLING AT INCREASING PEDALLING RATES IS NOT COMPATIBLE WITH INVARIANT RELATIVE TIMING MODELS.

T.G. Welter, W.M.M. Dorssers, A. Beelen, G.J. van Ingen Schenau.

Institute for Fundamental and Clinical Human Movement Sciences, Amsterdam, The Netherlands

INTRODUCTION

In the search for the principles used by the CNS to simplify the inverse dynamics problem in multi-joint movements, various theories have been proposed. For the case of increased movement speed it has been hypothesized that muscle activation patterns were executed faster but with invariant relative timing between muscle activity bursts (Schmidt, 1985). Modifications could be initiated by temporarily suppressed invariance. The meaning of these modifications could be to account for inertial effects that become larger at higher movement speeds. In many tasks muscles do not only account for movement kinematics but also for forces exerted on the environment. Previous studies revealed a special function of bi-articular muscles in the control of the direction of pedal force. This function was however never related to inertial effects. Cycling at increasing pedalling rates was used to investigate movement coordination with respect to these two principles.

METHODS

Six well trained cyclist performed three 3 bouts of 5 minutes cycling at 430 W with pedalling rates of 50, 80 and 110 revolutions per minute (RPM) after proper warming up. EMG (8 leg muscles), pedal forces and filmdata were sampled. Inverse dynamic analyses yielded mean curves per cycle for the three conditions. In the analyses kinematic and quasistatic moments according to Hull and Jorge (1985) were used. An electro-mechanical delay was incorporated in muscle EMG to indicate true muscle force delivering periods.

RESULTS

In the downstroke all muscle activation curves showed a shift to app. 30° later in the cycle at increasing pedalling frequency which corresponded with a shift of force direction on the pedal. Bi-articular muscles were active outside their shortening periods, mono-articular muscles were not. Just after TDC higher inertia caused a larger knee extending effect. Rectus femoris that extends the knee, lowered its activity accordingly to preserve the required pedal force vector. An opposite effect was found for hamstrings and gastrocnemius who increased their activity as inertia did not contribute to the necessary knee flexor moment at bottom dead center (BDC). In the lowest pedalling frequency an extra hamstring burst was present in the upstroke, related to smaller inertial effects which was absent in higher pedalling rates. Muscles showed non-uniform timing changes in rising and descending slopes of the activity curves.

DISCUSSION

We found that the phase and amplitude changes in total torque curves and the activity especially of bi-articular muscles were not compatible with the notion of the invariant timing in muscle activity patterns. Muscle activity changes appeared specifically aimed at exploiting inertial effects. It is obvious that movements are regulated by the continuously changing flow of proprioceptive information. However, time constraints make it very unlikely that feedback is used in triggering and shaping activity on an ongoing basis in all movements. Nerve conduction delays and time constants in the excitation dynamics of muscles, represent a delay that is longer than the duration of most fast movements. In cycling this timelag would represent a considerable part of the pedal cycle and it would cause timing problems every time the speed is changed. Moreover, EMG data without EMD incorporated show that at higher pedalling speeds EMG in many muscles starts earlier in the cycle. This anticipation would not be expected if sensory triggering depends on for example the rate of stretch in one muscle occurring at a specific point in the cycle.

It seems more likely that muscle activity patterns are the result of a, possibly learned, central pattern generator whose output is adapted on the basis of sensory information in a feedforward fashion. The 'skill' of cycling would be acquired in a process of learning, leading to a rather detailed representation through gradual adaptations of neural networks. The adaptation in the cycling central pattern generator would then consist of changes that are optimized in the learning process, to incorporate the inertial effects into the task of generating force efficiently on the pedal.

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THE EFFECT OF PEDALING RATE ON COORDINATION IN CYCLING

R. R. Neptune[§], M. L. Hull[†] and S. A. Kautz[‡]

[§]Human Performance Laboratory, University of Calgary, AB, Canada

[†]Department of Mechanical Engineering, University of California, Davis, CA, USA

[‡]Rehabilitation R&D Center (153), Veterans Affairs Medical Center, Palo Alto, CA, USA

INTRODUCTION

The objective of this study was to examine the effect of pedaling rate on coordination defined by EMG burst onset/offset timing, and iEMG and to interpret any apparent changes in the context of muscle function.

METHODS

Electromyograms (EMG) from eight lower extremity muscles and crank angle data were collected from ten subjects at 250 watts across pedaling rates ranging from 45 to 120 RPM. The muscles examined were the soleus (SOL), medial gastrocnemius (GAS), tibialis anterior (TA), vastus medialis (VAS), rectus femoris (RF), medial hamstring (SM), biceps femoris (BF) and gluteus maximus (GMAX). Individual EMG burst onset and offset (expressed in crank angle) and integrated EMG (iEMG) were computed. In addition, a phase-controlled functional group (PCFG) analysis was performed to interpret observed changes in the EMG patterns in the context of muscle function. Functional regions were defined based on the work of Raasch et al. (1996) as either top (T), bottom (B), extensor (E) or flexor (F). Integrated muscle activation was quantified within each region to identify the primary muscle function as the region with the most activity. Then, the union of muscle activity within adjacent regions was examined and muscles were reclassified as bifunctional if this summed activity exceeded the maximum of the primary region activity by 20%.

RESULTS

The average EMG burst patterns indicated changes in both timing and magnitude across pedaling rates. A significant linear trend in the burst onset was identified for all muscles except TA and GAS. All significant trends shifted the burst onset earlier in the crank cycle for increasing pedaling rates except SOL which exhibited a small but significant linear trend towards a later shift. The burst offset analysis revealed significant linear trends to shift later in the crank cycle for all muscles except RF and TA.

Analysis of the iEMG revealed that the muscles responded differently to increased pedaling rate. The GAS, BF, and SM increased their excitation dramatically when pedaling rate increased with a significant linear association, as did VAS. GMAX and SOL were found to have quadratic associations with pedaling rate, with minimums occurring at 90

RPM. TA and RF had no significant correlation with pedaling rate.

Muscle function characterized by the PCFG analysis defined all muscles with single functions except RF and SOL which were defined as top-extensor (T-E) and extensor-bottom (E-B), respectively. GMAX and VAS were characterized as extensor (E) muscles while the hamstrings, BF and SM, and GAS were characterized as bottom (B) muscles. No muscle was classified as a flexor (F) muscle, and TA was the only top (T) muscle. As pedaling rate was increased to 120 RPM, the primary function of each muscle remained unchanged.

DISCUSSION

Except for GAS and SOL, individual muscles had a similar phase advance in EMG onset when pedaling rate increased. These phase advances were consistent with typical activation time constants. Thus, activation dynamics appear to be a primary factor influencing muscle excitation as pedaling rate increases.

The iEMG of the hip and knee extensors (GMAX and VAS) used to accelerate the crank during the E region was not as sensitive to pedaling rate as was the iEMG of the bottom transition muscles GAS, BF and SM. Previous simulation research has shown that TA and RF muscles are particularly well-suited to propel the crank through the top transition region, while GAS, BF, SM and SOL are well-suited for the bottom transition region (e.g. Raasch et al., 1996). Changes in EMG data as cadence increases provide further evidence for the importance of transition muscle contribution to pedaling through these regions (where limb movement switches between flexion and extension). The large increases observed in GAS, BF and SM EMG activity represent a pedaling rate sensitivity that is possibly related to coping with the increasing magnitude of velocity-dependent interaction forces arising either between individual limb segments, or at the crank (Kautz and Hull, 1993).

ACKNOWLEDGMENTS

This research was supported by Shimano Corporation of Osaka, Japan.

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THREE-DIMENSIONAL KINEMATICS OF THE FOREFOOT, REARFOOT, AND LEG WITHOUT TIBIALIS POSTERIOR FUNCTION DURING STANCE PHASE OF WALKING: A CASE STUDY

U. Rattanaprasert*, R. Smith*, M. Sullivan*, W. Gilleard*

*Faculty of Health Sciences, The University of Sydney, * St. Vincent's Clinic, Sydney, AUSTRALIA

INTRODUCTION

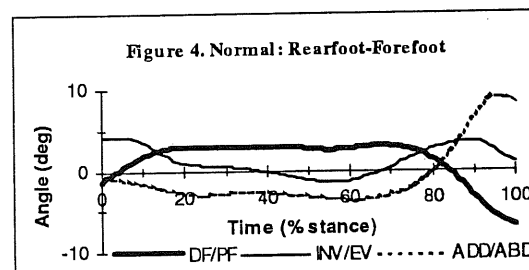
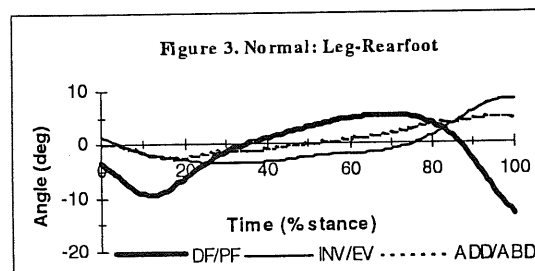
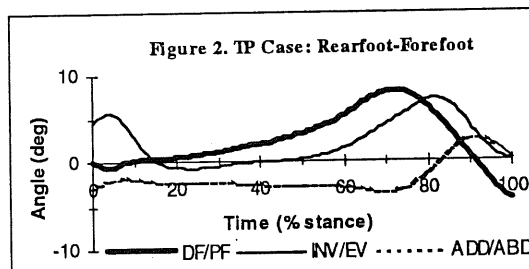
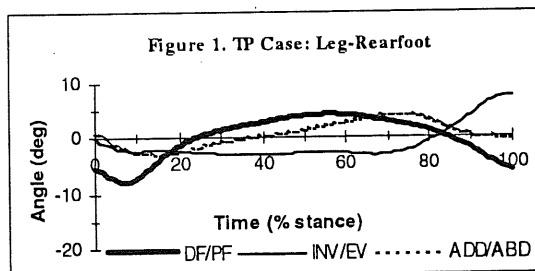
Tibialis posterior (TP) is a strong invertor of the foot and has a major role in controlling forefoot motion in walking and running. Its insertion, as many tendinous expansions, provides dynamic support along the plantar aspect of the foot and maintains the medial longitudinal arch¹. It also helps to maintain the balance of the tibia on the foot. Many studies have shown that lack of the functional activity of this muscle leads to changes in the longitudinal arch and affects the motion of the foot^{2,3}. This study examined the three-dimensional (3D) motion of the leg, rearfoot and forefoot in one case without the function of tibialis posterior in comparison with the ensemble average of 10 normals⁴.

METHODS

A 35-year-old female subject, with right tibialis posterior tendon laceration with a complete tear at the age of 6 which was not repaired surgically, participated in this study. The MRI of the foot revealed no abnormal development of the bones. The subject had eleven retroreflective markers placed on the right forefoot, rearfoot, and leg. Expert Vision Advanced (EVA) Motion Analysis System™ with four cameras was used to collect data at 60 Hz. Five trials of walking at a self selected speed were recorded. A synchronised force platform record was used to determine the frame of first foot contact and toe-off. The 3D trajectories of the markers were measured and calculated by the EVA Motion Analysis System™. The relative angles between the adjacent segments were then computed according to the International Society of Biomechanics conventions. The angular displacement data was time-normalised and averaged.

RESULTS AND DISCUSSION

The 3D motion of the rearfoot relative to the leg and the forefoot relative to the rearfoot without the function of TP showed different patterns from the same motion for normals⁴ as presented sequentially in Figure 1 & 2 and Figure 3 & 4. After heel strike, the rearfoot started to dorsiflex (DF) and adduct (ADD) earlier. At about 60% stance it slowly plantarflexed (PF) but failed to maintain ADD while preparing for toe-off. Inversion/Eversion (INV/EV) had a smaller range of motion than in normals. The increased range of DF and INV of the forefoot during foot flat and prior to toe-off was consistent with the fact that there was no functional activity of TP to maintain the arch of the foot under the stress.



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A SELFORGANIZED NEURAL NETWORK FOR GAIT ANALYSIS

Schmid O.*, Schöllhorn W.***, Bauer H. U.***

* Department of Orthopedics, Friedrich-Alexander University of Erlangen-Nuremberg, Germany

** Institute for Sports Sciences, Johann-Wolfgang-Goethe University of Frankfurt/Main, Germany

*** MPI für Strömungsforschung, Göttingen, Germany

INTRODUCTION

The purpose of this study was to obtain additional information to the standard gait parameters about the pre-operative patients with hip or knee osteoarthritis (each 10 patients). In order to compare the findings with normal gait patterns we added an investigation of 16 healthy persons (8 female, 8 male). Kinematic data, especially joint-angles and angular velocities were taken as input in a non-supervised artificial neural network (ANN). Other investigators (e.g. Holzreiter and Kohle, 1993) used supervised neural networks to determine 'normal' from 'pathological' gait or to estimate locomotion parameters (Sepulveda et al., 1993). By using a non-supervised ANN the ability to distinguish between pathological and healthy movement patterns is already implicit. 'Pseudo-unsupervised' ANNs were used by Hassoun et al. (1994) for classification of EMG-waveform signals. Feature vectors of the different findings were compared with known feature vectors. In the literature no investigation with kinematic data training a non supervised ANNs was found. This study provides first impressions about usage of this approach in clinical environment on decision making as well as on rehabilitation context.

METHODS

Our laboratory setup includes capture of kinematic data. Kinematic data were obtained using computer aided video based 3D-Motion Package (albavision). The patients' motion on a walkway was filmed by four synchronized CCD-video-cameras with a frequency of 50 Hz in order to capture every point of interest at least by two cameras. 15 body landmarks were marked with colored markers on well palpable anatomical structures in a modified Helen Hayes marker placement version (Kadaba et al., 1990). At minimum 3 trials of each subject were recorded on the walkway. After 3D reconstruction using a similar approach to Tsai (1985) calculation of 6 degrees of freedom joint motion were executed according to anatomically relevant joint axes (Grood and Suntay, 1983). Every highdimensional gait pattern was mapped on a low (four) dimensional trajectory in a virtual space by a selforganized Kohonen map (SOM). The Kohonen map was trained by means of the time courses of the following kinematic variables: Angle and angular velocities and angular acceleration of the left and right hip, knee and ankle joint in frontal and sagittal planes. In order to structure the mapped trajectories the euclidian distance between all trajectories were determined and cluster analysed using single linkage method. Additionally, the ranges (difference of maximal and minimal value) and initial values of every variables' time course was determined and cluster analysed.

RESULTS AND DISCUSSION

Our findings of the time and distance parameters were within the ranges of others investigators' data from osteoarthritis patients (e.g. Berman et al., 1987). The cluster analysis of the time course variables revealed different clusters than those of the initial and range values. Whereas the analysis of the variables time courses by means of the SOM was able to separate 'normal' from 'pathological' gait patterns no criteria could be found for the time discrete clustering. The analysis of different groups of variables led to the best assessment, the more variables we included. The exclusive consideration of the angular time courses provided the worst result. Comparing to principal component analysis we could improve the percentage of recognition up to 70% by means of the SOM. Considering single variables, different contributions to the overall differences within the clusters could be diagnosed. In comparison to other movement patterns recognition methods the SOM does not need any knowledge in advance. A principal advantage of the SOMs is the unveiling of the latent structures of a flow of data whereas the supervised artificial neural networks require preknowledge of the system. According to the principles of visual perception the SOM provides the possibility to analyse a highdimensional data set as a whole as well as in parts. Transferring to movement analysis the presented combination of SOM and clusteranalysis provides an effective supplementary tool for the diagnosis in clinical areas. Additionally, the high recognition-rate suggests to use this approach for using the feature vectors for the optimization of the rehabilitation process. Because of the integration of a patient via the time course of feature vectors into different groups we will be able to develop more individualized rehabilitation programs in order to shorten this process.

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INFLUENCE OF WALKER HEIGHT ON UPPER EXTREMITY JOINT MOMENTS DURING WALKER-ASSISTED GAIT

G. G. Simoneau, G. W. Hambrook, R. A. Bachschmidt, G. F. Harris

Physical Therapy and Biomedical Engineering Depts., Marquette University, Milwaukee, WI USA

INTRODUCTION

To date, minimal research has been done to describe upper extremity kinematics and kinetics during walker ambulation. This lack of information is in part due to the lack of appropriate instrumentation to measure the "walker reaction forces" applied to the user's hands. We hypothesized that the magnitude of the loads applied to the upper extremities will be partially dictated by walker settings such as walker height. The purpose of this study was to determine the effects of walker height on the joint moments at the elbow and shoulder during walker ambulation.

METHODS

A walker dynamometer, consisting of a standard tubular aluminum walker instrumented with 24 double-pattern resistive foil strain gages, was developed to measure the magnitude and direction of the forces applied to each handle of the walker when ambulating (Bachschmidt et al., 1995). Bilateral 3-D joint kinematic data of the trunk and upper extremities were acquired with a six-camera Vicon (Oxford, U.K.) motion analysis system. The inverse dynamic approach was used to combine walker reaction forces with kinematic data to determine 3-D wrist, elbow and shoulder joint moments bilaterally.

Data were collected on a healthy 21-year-old female. Three walker heights were used: 1) walker height set at the level of the ulnar styloid process of the subject while standing (this setting is the clinically accepted standard walker height), 2) walker height set at 5.08 cm (2 inches) above the standard height and 3) walker height set at 5.08 cm below the standard height. Four full gait cycles were acquired for each of the three conditions. Ambulation was performed with the left leg being nonweightbearing. A standard step length of 0.40 meter was used.

RESULTS

While 3-D data are available for each joint, the data presented below are limited to the flexion-extension moments of the right elbow and shoulder for the three conditions. Each trace in the graphs represents an ensemble average of four full gait cycles from walker contact to subsequent walker contact. The joint moments are expressed as a ratio of the peak moment obtained for the highest walker height setting.

Figure 1. Shoulder flexion/extension moments. WO and WC correspond to the walker being lifted off the ground and the walker being repositioned on the ground, respectively. Loading on the upper extremities, causing a shoulder flexion moment, takes place from WC to WO. During this phase of the gait cycle, the weight of the body is progressively transferred to the upper extremities in order to advance the right foot forward. While the general features of the three moment curves are similar among the three conditions, a difference of approximately 10% is noted in the peak torque values between the highest and lowest settings.

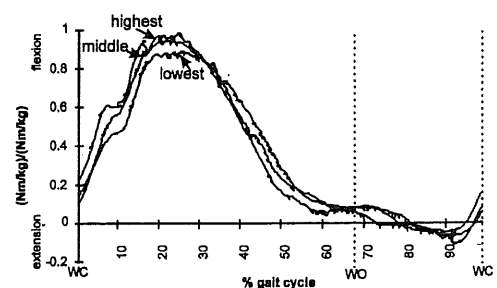
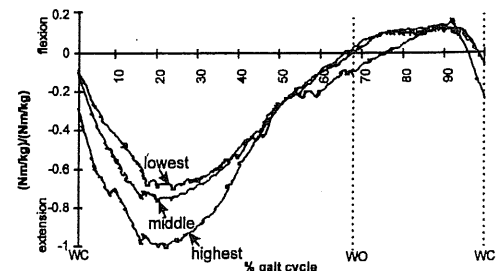


Figure 2. Elbow flexion/extension moments. An internal elbow extension moment is created when loading the upper extremities in order to move the right foot forward. While the general appearance of the moment curves is similar among the three conditions, a difference of approximately 30% is noted in the peak torque values between the highest and lowest settings. A small flexion moment is noted when the walker is lifted (WO to WC).



DISCUSSION

We have successfully instrumented a walker that, in conjunction with kinematic data, allows us to describe upper extremity joint moments during walker ambulation. Data collected to date indicate that walker height affects the magnitude of the joint moments at the shoulder and elbow, with the magnitude of the moments increasing with the higher walker height settings. These findings have direct and significant implications in the clinical management of patients who require the use of a walker. We continue to collect data on young control subjects and older individuals in order to further describe upper extremity joint kinetics during walker ambulation.

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ACKNOWLEDGMENTS

The authors would like to thank the Foundation for Physical Therapy and the NIH (1T32HD07437-01A1) for partially sponsoring this study.

THE INFLUENCE OF ENERGY TRANSFER BETWEEN THE HUMAN BODY AND THE HORIZONTAL BAR OR UPPER ASYMMETRIC BAR UPON THE GYMNASTIC PERFORMANCE

A. Arampatzis and G.-P. Brüggemann

Institute for Athletics and Gymnastics, German Sport University Cologne, Germany

INTRODUCTION

The execution of difficult flight elements and dismounts is a prerequisite for internationally acceptable performance. The gymnast requires a high initial energy to successfully perform these elements and this initial energy is derived from the preceding movement element. On the horizontal bar and the asymmetric bars this is usually the giant swing.

The major aims of this study were:

1. To understand the influence of muscular energy supply upon the utilisation of the bar's elasticity.
2. To compare two different execution techniques of the giant swing prior to dismounts on the horizontal bar and the giant swing on the asymmetric bars.

METHODS

Experimental data were collected during the finals of the World artistic gymnastic championships in 1994. The experimental setup consisted of two video cameras (recording frequency: 50 Hz) and two bars instrumented with force sensors (sampling frequency: 500 Hz). One camera recorded the complete body movement while the other synchronously recorded only the movement of the bar. The instrumented bars permitted the recording of two orthogonal forces (vertical and horizontal). The data analysed were 28 giant swings prior to dismounts from the horizontal bar (19 executed with the power technique and 9 with the conventional technique) and 20 giant swings prior to dismounts from the asymmetric bars. The human body was modeled by a 15 segment system. The 15 segments were connected by 14 rotational and 4 translational joints.

RESULTS

The total energy of the body, the angular momentum of the body and the horizontal velocity of the centre of mass were significantly higher ($p < 0.05$) at the conclusion of the power technique. The mechanical power in the shoulder and hip joints demonstrated that muscular energy was only produced in the upward phase of the conventional giant swing technique on the horizontal bar and asymmetric bars. However, in the power technique energy was also produced in the downward phase.

CONCLUSION

There are two characteristic major phases in the giant swing prior to dismounts from the horizontal bar and asymmetric bars. During the first phase energy flows from the gymnast's body to the bar. In this phase the athlete can utilise the elasticity of the bar if the energy transferred to the bar is greater than the decrease in the total body energy. The elasticity of the bar was only utilised in those giant swings executed with the power technique (criterion 1, fig. 1).

The energy transferred to the bar returns to the body during the second phase. In this phase the athletes can utilise their muscular capacity if the increase in total body energy is greater than the energy returned from the bar. There was no significant ($p < 0.05$) difference in the utilisation of muscular capacity between the two techniques on the horizontal bar. The gain in energy was least on the asymmetric bars (criterion 2, fig. 1). The power technique can be used on the asymmetric bars if the gymnast hyperextends the upper body in order to pass the lower bar.

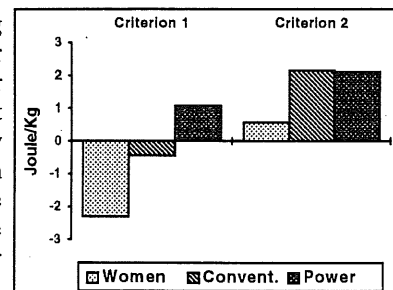


Fig.1: Criteria 1 and 2

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ANALYSIS OF THE DOUBLE BACK SALTO DISMOUNT FROM THE PARALLEL BARS

PIERRE GERVAIS

Faculty of Physical Education and Recreation, University of Alberta, Edmonton, Alberta

INTRODUCTION

The dismount is the last element of a routine and in international competition is easily the most discriminating factor between performances. Coaches have identified Parallel Bars as one of the poorest events with respect to controlled landings. The predominant optional dismount used by male gymnasts from Parallel Bars is the double back salto. Takei et al. (1992) proposed that mastering of the fundamental skill can prepare the gymnast for safe and efficient learning of more complex and higher risk dismounts. The thorough learning of this skill can also lay the foundation for what has been described as the preprogrammed strategies to control the landings from these dismounts thus minimizing point deductions while maximizing safety. The purpose of this study was to evaluate the mechanics of the double back salto dismount from Parallel Bars, performed in the tucked position.

METHODS

Data for this study was collected at the 1995 Canadian National Artistic Gymnastics Championships. Nine subjects, two novice, two junior and five senior level gymnasts, all familiar with the dismount, volunteered to participate in the study. The subjects were video taped from the side and data was sampled at 60 Hz using the APAS system by Ariel Dynamics, Inc.. The subjects' angular momentum was determined as per the methods of Gervais and Tally (1993). Transverse axes through the shoulder joint, C of G and ankle closest to the camera were chosen as the axes of rotation while the gymnasts were on the bars, were airborne and were landing, respectively. Each dismount was judged independently by an experienced judge. Each dismount was judged out of 5 points each for the dismount and landing components of the skill. Product-moment correlation between the measured mechanical factors and the judge's score were determined.

RESULTS AND DISCUSSION

A total of 29 performances were recorded. The swing and take-off phase started with initial movement from the handstand. The vertical velocity at take-off was significantly correlated to that of the total points awarded ($r=.39$, $p<.05$). The angle of take-off which ranged from 67.2° to 89.9° , although not statistically significant, did indicate a predominantly vertical take-off trajectory for the centre of gravity. Statistically, the greater the shoulder extension and the higher the hips at release the greater the score awarded by the judge. The average take-off angular momentum was $50.6 \text{ kg.m}^2/\text{s}$. On average the angular momentum reached its maximum value at a trunk angle with respect to vertical of 24.1° prior to the bottom of the swing. Flight time was positively correlated to the points awarded for the dismount score. How tightly the gymnast tucks determines the angular speed with which the gymnast will rotate given the quantity of angular motion he has from take-off. The hip angle was significant for the landing score ($r=.39$) and total score ($r=.46$) at the .05 and .02 level of significant respectively. As was suggested by Takei et al (1992) in their study of the double back salto from horizontal bar and which was revealed here, the tighter the tuck and the earlier the kickout is executed the more controlled the ensuing landing and the higher the points awarded. It was found that greater success was achieved when the gymnast utilized a more rapid and complete hip extension during the course of the kickout in preparation for the landing. This provided for a significantly better landing position which when combined with a controlled lowering through knee flexion, as opposed to primarily hip flexion, minimized point deductions or losses. The landing phase was defined from the point of contact with the mat to the position in which the centre of gravity was at its lowest height. At contact with the mat, the greater the lean the smaller the score received or alternatively the more extended the hips at landing the higher the points awarded. When comparing the knee angle to hip angle bend at the lowest point (end) of the landing, the average landing bias (knee/hip angle) ratio of 1.76 indicated a greater hip bias or on average a greater utilization of hip flexion in the attenuation of the landing momentums. However, this was found to be negatively correlated to the judge's score ($r=-.42$, $p<.05$). This would suggest that the smaller the ratio or the greater the knee flexion during the landing the smaller the point deductions during the landing phase of the skill. Since the knee angle at the lowest point and its range of motion were also found to be significantly related to the landing score it can be argued that a better landing score is achieved with greater knee extensor involvement through a greater range of motion during the landing as opposed to an attempt at controlling the landing through hip flexion.

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Acknowledgement: This study was supported in part by an Applied Sport Research Program, Sport Canada Grant.

3D ANGULAR MOMENTUM ANALYSIS OF DISMOUNT LANDING IN GYMNASTICS

Yu Liu

Chinese Culture University, Taipei, Taiwan

INTRODUCTION

Landing is one of the important techniques to achieving high performance in gymnastics. Dismount landing mechanics is most complicated and therefore relatively little research has been devoted to this issue (Bruggemann, 1989). In previous studies, landing mechanics were often discussed on horizontal velocity of body center of mass (c.m.), force diagram and human balance theory (Bruggemann, 1989; Liu Rong-Zeng, 1984). The purpose of present study is to investigate the biomechanical contributors to dismount landing technique with regard to angular momentum (AM) during flight and landing.

METHODS

Four elite gymnasts of national level were videotaped 3-dimensionally at a rate of 60Hz during performance of handspring and salto fwd tucked vault in gymnastics using Peak Performance Motion Analysis System. According to the score given by judges, trials of each subject were selected and divided into two groups (n=8): one for successful landing and one for failed landing. Meanwhile, 3D AM about the c.m. of the whole body, AM of the main body parts and their percentage to the whole body for second flight and landing phases were calculated. A comparison between successful landing and failed landing was conducted using paired T-Test in order to find biomechanical contributors to the landing technique.

RESULTS

Fig. shows the time curves of the AMs of the whole body and body parts during a successful landing and a failed landing. The standard error of standardized AM ($AM/mass \cdot height^2$) of whole body is about 10% in present study. This is similar to the results from Yeadon (1990) and Dapena (1978). From present calculation, there is no significant difference between the values of AM for successful landing and failed landing. Nevertheless, a significant difference is given between these two groups regarding the lower limb's AM and its ratio to the whole body ($T=-3.58, p<.05$).

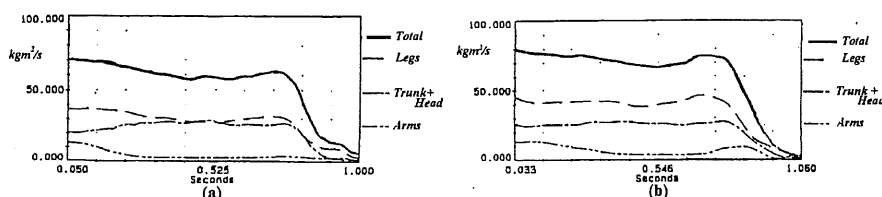


Fig. Time curves of the angular momentum
(a) Successful landing ; (b) Failed landing

DISCUSSION

According to Liu Rong-Zeng (1984), gymnastic landing is actually a deceleration during landing of a projectile motion. It involves flight, landing, deceleration and stop. The movement during flight - to control and regulate the AM - before landing is a major influence factor on the quality of landing. From our results, there is no difference between successful and failed landing with regard to whole body AM. The movement, especially the relative movement of body parts about the c.m. during the flight contribute to the quality of landing. Lower limb's AM and its ratio to the whole body's have a significant bearing on the relative movement about the c.m. and the timing of this movement. It is therefore an influence factor on the landing quality.

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ANALYSIS OF MOTION ON SPEED SKATING

Kousuke Okamoto, Tetsu Nishimura, Naoki Watanabe, Koga Yuzo, Sakata Takuya, Fukagawa Yuzo.
Dept. Mech. Eng., Col. Sci. & Tech., NIHON University, Tokyo, Japan

INTRODUCTION

In this study, we analyze the speed skating motion dynamically from the view of mechanics to promote the record of the Japanese skater. The speed skating is performed on the oval rink, so skating motion can be divided into three typical motions as following phases. That is, a starting motion, a skating motion on straight course and curved course. Here we only discuss a straight course skating motion. The subjects for this study are the sprint skaters.

METHOD

For analysis of a human body motion, a human body is replaced by a mechanical links system. This links system is composed of link members as body elements and nodal points as human joints. The mechanical analysis of human motion demands to obtain the actual displacement data of nodal points. Thus, we take video picture images on straight course skating, and transform these images into digital data on computer display. Finally, we obtain 3-dementional data by converting 2-dimensional data with D.L.T.method.

We adopted right handed 3-dimentional coordinate system, as X-axis is placed to the normal direction from inner course to outer course and Y-axis directs to the direction along the straight course.

RESULT

As everyone knows, skating motion is very complicated. So, we only discuss the motion of legs. The motion of the legs can be divided into typical two motion. First one, the gliding and kicking motion on the ice. And Second, the drawing motion after kicking motion. Each subject have various features on the drawing motions. Here, the two typical motions ,that is, the motion of subject A and B, are compared.

Fig.1 shows the relative motion of the ankle to the opposite ankle on the gliding blade. In this phase, the right blade is gliding on the ice, and the left blade is free from the ice and is being drawn. From this figure, the difference between subject A and B is apparent. When the blade catches the ice, the moving direction of ankle of subject B differs from that of the Center of Gravity (abbreviate C.G.). On the other hands, for subject A, the moving direction of ankle is parallel to that of the C.G. It is imagined that the blade motion on catching the ice for the subject B may cause deceleration larger rather than the subject A.

Fig.2 shows the motion of the C.G. and ankles on straight course skating. The solid line expresses the motion of the ankle at the time the blade is gliding. And, dotted line expresses the drawing motion of the legs. The movement of the C.G. is expressed by chain line. For subject A, just after the blade catching the ice, the moving direction of the blade and the C.G. proceeds parallel each other for a while. We designate this section as Ls.

Fig.3 shows the relation between length of Ls and mean velocity on 4 subjects. For subject A, in contrast to subject B, length of Ls is the longest and mean velocity is also the highest.

DISCUSSION

As a result, shown in Fig.3, it seems that subject B should correct his drawing motion shown in Fig. 1. And also it is imagined that longer Ls cause the higher mean velocity.

But, the blade in the Ls phase can not produce the large acceleration toward to the proceeding direction. Hence, distance of Ls should have optimum value.

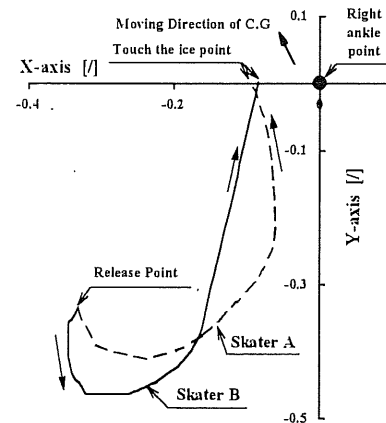


Fig.1 Relative displacement of ankle

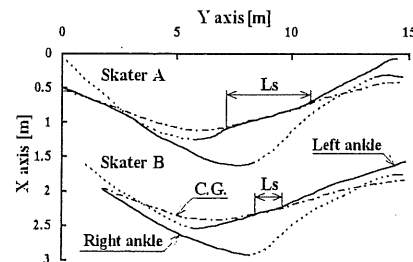


Fig.2 Ankles and C.G.

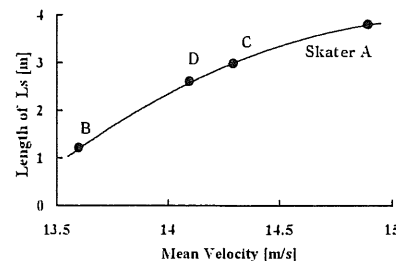


Fig.3 Ls - Mean velocity

Biomechanical Analysis for the Races of Short Track Skate

Kiyomi Ueya (Yamanashi University, Yamanashi, Japan)

Akinori Yoshihara (Graduate school of Yamanashi University, Yamanashi, Japan)

Kazunori Asaba (Tsuru Bunka University, Yamanashi, Japan)

Introduction

Short track skate is one of the event of the Winter Olympic Games like Speed Skate, but its popularity is not so high. An original country of short track skate was England, and the first competition was held in 1938. Nowadays, there are races of 500m, 1000m, 1500m, 3000m for men & women, 3000m relay for women, 5000m relay for men.

A biggest feature of the race of short track skate is an "order system", is not a "time system". Consequently, an order of a goal is important than a time in each race. In order to win a preliminary race to the quarterfinals (semi finals, finals), the skater have to goal within second place in each race.

A distance of one round of short track rink is 111.12m and a curve is a circle with a radius of 8.00m (8.25m) and a length of a straight line is 57.71m (56.14m). How to skate at each phase, especially at a curve, is very important in the race of short track skate.

Consequently, short track skaters are required to skate using a special techniques based on speed, balance, agility, coordination and the good judgements based on the anticipations for a race.

Purpose of This Study

The purpose of this study was to analyze the races of short track skate and to find the biomechanical features of Japanese Olympic skaters. This study was done as one of the consolidation for Japanese Short Track Skate Team of Nagano Winter Olympic Games. That means the consolidation is expected for Japanese skaters to get the medals in Nagano Winter Olympic Games. In that means, this study was done as a contribution and an introduction of biomechanics to the competition sports activities.

Method of This Study

1) Analytical Competitions

(1) 1996, 1997 All Japan Championships, (2) 1996, 1997 World Championships (Hague; Netherland, Nagano ; Japan), and (3) Friendship Races of Canada and Japan (1996).

2) Analytical Method

Eight video cameras were operated to film the races of the skaters at each competition.

3) Analytical Items of skating at the Races

- ① Lap time of each phase of the rounds, ② Start time/speed, ③ Cornering distance/time/speed,
- ④ Last spurt distance/time/speed, ⑤ Distance/time of cornering by left/right foot support,
- ⑥ Centrifugal force at a curve, ⑦ Skating forms at each phase of the race.

Results and Conclusions

In the races of short track skate, not only sprint ability, speed endurance but also the tactics for winning became the important factor. Japanese short track skaters have the good techniques for skating, especially at a curve, but they don't skate based on an explosive speed, muscular power when they need to change speed and speed up freely at any phases of a race.

Biomechanical advices gaining from this study for improving the techniques of skating brought the good results to Japanese Olympic Short Track Skaters.

THE EFFECTS OF A SHORT-TERM POWER TRAINING OF THE ANKLE JOINT ON THE STARTING TECHNIQUE OF KOREAN FEMALE NATIONAL SHORT-TRACK SPEED SKATERS

Young-Hoo Kwon & Myung-Kyu Jun*

Korea Sport Science Institute, Seoul, KOREA

*Korea Skating Union, Seoul, KOREA

INTRODUCTION

In recent years, Korea has been one of the dominant forces in short-track speed skating since the introduction of this sport to Korea in early 80s. Korean female skaters have been strong in the 1,500-m and 3,000-m events, but revealed weakness in the 500-m and 1,000-m events due to low starting ability. Unfortunately, more emphasis is on the 500-m and 1,000-m events in the World Team Championships (no 1,500-m event) and only 500-m and 1,000-m events are included as individual events in the Winter Olympics.

The purpose of this study was to biomechanically evaluate the effects of a two-week power training of the ankle joints on the starting technique of the Korean female national skaters in an effort to enhance the starting ability for successful performance at the 1996 World Team Short-Track Speed Skating Championships.

METHODS

Four female Korean national skaters participated in this study as subject. The mean age, mass and height were 16 years, 51.3 kg, (SD: 2.63 kg), and 1.61 m (SD: 0.06 m), respectively. A 2-week power training program for the ankle joint was administered using a Cybex-6000. A pre-test was performed before the training program started with ankle angular velocity being 30 and 120 deg/s, and the training protocol was selected based on the test results: 3 sets at 150 deg/s, followed by 4 sets at 120 deg/s, and 5 sets at 90 deg/s; 10 reps per set; sets separated by 1- or 3-min rest. A post-test was administered at the end of the training period to evaluate the power improvement of the ankle.

The sprint starts of the subjects were videotaped before the training program started and right after they returned from the Team Championships for three-dimensional motion analysis. Four S-VHS video cameras were employed, 2 of which covered the first half of the 14-m start interval while the others covered the second half. Continuity in time and space among the video views was secured through gen-locking the cameras and using range poles for 3-D DLT calibration. Start interval time, position and velocity of the vertex of the head, step length and rate, and the body lean angle were computed for data analysis.

RESULTS

The start interval (14 m) time decreased from 3.20 s (SD: 0.09) to 3.11 s (SD: 0.04) by 0.09 s on the average. All subjects scored decreased start interval time in the post-experiment. The difference in the horizontal position of the vertex of the head between the pre- and post-experiment increased as time elapsed throughout the start interval. Both step length and rate of the first two steps increased from 0.70 m (SD: 0.15 m) to 0.74 m (SD: 0.10 m) and from 2.32 steps/s (SD: 0.21 steps/s) to 2.45 steps/s (SD: 0.15 steps/s), respectively. All subjects scored increased step length and rate. The subsequent steps revealed no consistent change.

DISCUSSION

The post-experiment trials were generally characterized by lower initial starting position, increased step length and rate in the first two steps, and, as a result, increased starting speed. Although the starting speed was generally improved throughout the start interval, the initial acceleration ability was specifically enhanced due to the increase in step rate and length in the early phase of the start. It was observed that the starting posture (either flat or planting style) affected the position and velocity pattern of the vertex of the head, resulting minor differences in the patterns of these variables and in the initial body lean angle.

From the data analysis, it was concluded that the starting ability was indeed enhanced through a 2-week intensive power training of the ankle joint mainly through the improvement of the ability to accelerate in the early phase of the start. The Korean female skaters won all four 1,000-m matches they competed, two of four 500-m matches, and eventually the team championship at the 1996 World Team Championships. (In the Team Championships, four skaters from each country individually compete in 4 of 7 matches, one in each, in the 500-m and 1,000-m events.)

THE MAXIMAL POWER OUTPUT AND ISOMETRIC STRENGTH OF LOWER LIMBS IN JUNIOR TENNIS PLAYERS

Z. Trzaskoma, P. Lipiński, and J. Janiak
Institute of Sport, Warsaw, Poland

INTRODUCTION

Studies on tennis players representing various levels of achievement have increased in number in recent years and include such aspects as anthropometry, physiological characteristics of a tennis match, fitness development, influence of the equipment characteristics on the outcome or a complex evaluation of that sport (Bergeron et al., 1991, Groppel and Roetert, 1992). Some authors made an attempt at establishing normal values for selected physical fitness or tennis-specific skills parameters recorded in young tennis players. Experimental studies on the strength and power output developed by junior players, as well as relationships between these factors of player's physical potential, are scarce. The aim of this study was to establish the values of maximal mechanical power output and isometric strength of lower limbs in junior Polish elite tennis players and to determine relationships between power and strength indices.

METHODS

The study was conducted on 17 boys and 17 girls aged 13 to 16 years. Their mean ($\pm$ SD) age was 14.8 ± 1.1 and 14.3 ± 1.2 years, body height - 176.7 ± 5.5 and 169.9 ± 7.2 cm, body mass - 63.0 ± 4.6 and 58.5 ± 8.5 kg, respectively. Maximal torques of lower limbs muscles (MTLL - flexors and extensors of knee and hip joints) under static conditions were measured on the special stand designed and constructed locally. Values of maximal power output were recorded in three short-lasting laboratory tests: counter-movement jump (CMJ, 6 single jumps separated by 3 min intermissions, Kistler piezoelectric platform interfaced with microcomputer which used the local made computer programme to collect data), legs exercise on the pendulum device (PEN, 3 sets of 10 repetitions pendulum rebound swings, a pendulum device designed and constructed locally) and legs flexion/extension under isokinetic conditions (ISO, 3 sets of 6 repetitions, with the preset angular velocities: 400, 300 and 200 $\text{deg} \cdot \text{s}^{-1}$, CES Ariel 5000 device). The study was approved by the local Commission of Ethics.

RESULTS AND DISCUSSION

Highest values of maximal power output were attained in the CMJ test (boys - 2174 ± 367 W, girls - 1653 ± 307 W) and were significantly higher from those recorded in either ISO or PEN (boys - 1342 ± 265 and 1402 ± 313 W, girls - 1057 ± 186 and 1229 ± 346 W, respectively). Values of maximal muscle torques (the sum of both legs - MTLL) attained by boys and girls were as follows: knee extensors (KE) - 572 ± 91 and 481 ± 79 , knee flexors (KF) - 271 ± 52 and 206 ± 44 , hip extensors (HE) - 1028 ± 174 and 884 ± 150 , hip flexors (HF) - 150 ± 28 and 129 ± 18 N·m, respectively. Boys attained significantly higher values of maximal power and isometric strength than girls in all tests (except PEN). Highest values of maximal power output attained in three tests (CMJ, PEN and ISO) by junior tennis players in this study were lower by about 21% (boys) to 36% (girls) from those reported by us earlier for advanced male tennis players (Trzaskoma, 1997).

Maximal power	MTLL	KE	KF	HE	HF
CMJ boys	0.213	0.107	0.451	0.099	0.227
girls	0.636**	0.723**	0.188	0.483*	0.530*
PEN boys	0.141	0.100	- 0.011	0.151	0.100
girls	0.245	0.498*	- 0.469	0.194	0.506*
ISO boys	0.504*	0.380	0.353	0.496*	0.164
girls	0.545*	0.697**	0.118	0.393	0.387

Table 2. Person's correlation coefficients between the values of maximal power output and maximal isometric strength; * $p < 0.05$, ** $p < 0.01$.

Significant correlations between maximal power in the ISO test and MTLL were found for boys and girls while in CMJ test - for girls only. The lack of correlations between maximal values of power output, attained by young tennis players in most studied tests, and values of maximal muscle torques support the view that relationships between power and strength are changeable.

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TIMING AT THE LOWER EXTREMITY JOINTS IN VERTICAL LANDING

M. Ayalon, D. Ben-Sira, A. Ayalon

The Zinman College of Physical Education & Sport Sciences at the Wingate Institute, Israel

INTRODUCTION

A typical landing vertical ground reaction force (VGRF) curve is characterized by four phases: 1. ground contact to the first peak; 2. from the first peak to a local minimum; 3. from the local through to the second peak; and 4. from the second peak until the force stabilizes at body weight. These phases are related to the movements at the lower limb joints (Ayalon et. al. 1995). The focus of the current report is on the time between initial contact with the ground and the second peak force. During this very short time period (40-70 ms) the joints perform a timed coordination in order to minimize the forces which act on the body. The discussion of timing problems should not be limited only to the duration of joint action. Better understanding requires comprehension of the "quality" of the work done by the joints. The purpose of the study was to investigate the timing coordination of the lower extremity joints during vertical landing (VL) with special emphasis on comparisons between "good" (GL) and "bad" (BL) performances as classified on the basis of peak VGRF.

METHODS

Subjects were ten physical education students (age 23-27). Each subject performed ten vertical landings from a specially designed platform to ensure verticality of the landings, set at a height of 0.60 meters. Intervals of at least three minutes were provided between trials. The VGRFs were measured with a Kistler force plate (model 9281b). The joint angles were measured by Mie electrogoniometers set at the ankle, knee and hip joints. All data were transferred through a Metro-Byte A-D converter at a sampling rate of 2.5 KHz, into a PC computer for analysis. The high sampling rate was selected because of the very fast nature of the activity under investigation, and in order to be able to identify rapid changes in the coordination chain.

RESULTS

Differences between BL and GL were found in the magnitude of the angular velocities and angular accelerations at all joints within each of the subjects. In BL values were as much as 40% higher. In these comparisons, no significant differences were found in either the initial joint position at ground contact, or in the range of motion. However, when comparing performances across subjects, kinematic variables in BL of one subject were not necessarily different from those of GL in another subject.

In GL the temporal sequence between the ankle and the knee was found to be as follows: 1. ankle acceleration, 2. knee acceleration/ankle deceleration - starting simultaneously 3. knee deceleration. While this pattern appeared in all GLs, among subjects, no specific pattern was observed in BLs. Nevertheless, a temporal offset between the initiation of ankle deceleration and of knee acceleration was clearly observed.

DISCUSSION

The results demonstrate that a specific kinematic sequence characterizes GL. On the other hand "bad" performance can not be characterized by any specific sequence of events. They can only be negatively identified by their deviation from the GL model.

Absolute kinematic values are found relevant for distinguishing between inter-individual "good" and "bad" landings. However, these criteria are not valid for the explanation of intra-individual variation.

Successful performance with regard to the criterion of minimizing peak VGRF, is associated with a unique sequencing of kinematic parameters of the lower extremity. When evaluating performing quality, higher priority can be attributed to kinematic variables such as joint angular velocity and angular acceleration over position and range of motion.

The complexity of VL has significant implications for the learning of this skill. Skill improvement can be based on qualitative manipulation of the timing variables during skill acquisition. Previous investigations demonstrate that feedback of biomechanical variables may contribute to the learning of motor skills (Hatze, 1976; Ayalon & Ben-Sira, 1988). Further investigation is required in order to determine whether feedback regarding the precise timing of the kinematic chain may positively affect performance quality in VL.

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IS JOINT LOAD IN SQUAT LIFTING RELATED TO INTERJOINT COORDINATION?

Lars Lindbeck and Katarina Kjellberg
National Institute for Working Life, S-171 84 SOLNA, Sweden

INTRODUCTION

Work technique is a somewhat unclear concept but the term is often used as well in occupational biomechanics as in sport biomechanics. The term has an obvious connection to performance and movement coordination. It has been observed that the interjoint coordination changes systematically with increasing load during squat lifting (Scholz 1993; Burgess-Limerick et al. 1995). Moreover, gender differences have been observed in the performance of an incremental lifting machine test in terms of timing, displacement, velocity, acceleration, force and power (Stevenson et al. 1990; Stevenson et al. 1996). The purpose of the present study was to investigate whether the degree of synchronization of hip-knee coordination would be related to the net joint reaction moments and if there are any gender differences in this respect.

METHODS

The study was carried out by means of two series of squat lift experiments on twelve female subjects and on ten male subjects, respectively. The subjects, standing on a Kistler force plate, performed squat lifts of a box with handles using two velocities; a fast lift of approximately 1 s and a slow lift of 2 s. Five trials of each lift type were measured. Only the third was analyzed. The box was lifted from the level of the force plate to a table adjusted to navel height. The weight of the box was for the men 12.8 kg and for the women 8.7 kg. The difference in load was assumed to correspond to differences in physical capacity.

Markers were attached to the ankle, knee, hip, shoulder, elbow, and wrist joints, respectively. Marker positions were registered by means of optoelectronic motion analysis systems (Selspot II and Mac Reflex). Force plate and movement data were sampled simultaneously at 50 Hz. Angular motions and net joint moments in the knee and the lower back were then calculated from the recorded data. The interjoint coordination was quantified as a relative phase angle between the knee joint and the hip joint, respectively, as suggested by Burgess-Limerick et al. (1993). Relative phase angles, i.e. the knee joint phase angle subtracted from the hip joint angle, were determined and max and min values of the relative phase angles were calculated for all subjects. The largest deviation from perfectly synchronized hip-knee coordination, i.e. when the relative phase angle equals zero, was used as a measure of coordination for the subjects. Net reaction moments in the knee joint and the L5/S1 joint, respectively, were calculated by means of a biomechanical 2-dimensional dynamic model.

RESULTS

The interjoint coordination was better synchronized for women than for men. The max and min values of the relative phase angles during the lift movement were on average +17 deg. and -33 deg., respectively, for women. The corresponding values for men were +25 deg. and -81 deg., respectively. The women's coordination as illustrated by means of angle-angle diagrams was in general similar for the slow lifts and for the fast lifts. For the men, however, the coordination appeared to depend whether the lift was a slow or a fast one. Neither for women nor for men was found any correlation between the degree of coordination and net moments in the knee or the lower back.

DISCUSSION

The analyses revealed that the hip-knee coordination was better in-phase for the women than for the men. The reason for that is difficult to explain. It has been reported that increasing load mass increases the deviation from perfectly in-phase-coordination (Scholz 1993; Burgess-Limerick et al. 1995). The men in our study lifted a heavier weight and it might be that it did not correspond to a greater physical capacity. The absence of correlation between coordination and joint load indicate that the degree of interjoint coordination is of minor importance for joint loads in squat lifting. It has been suggested that a decreased interjoint coordination might in some cases even decrease the required muscular effort (Burgess-Limerick et al. 1995).

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MODIFICATIONS OF KINEMATIC STRATEGIES DURING LEG RAISING IN BELOW KNEE AMPUTEES

M.Cincera^{1,2}, L.Mouchnino³, M.L.Mille³, A.Pedotti^{1,2}, J.Massion⁴

1 - Dept. of Bioengineering, Politecnico di Milano, Milano, ITALY

2 - Centro di Bioingegneria, Fondazione Pro Juventute 'Don C.Gnocchi', Milano, ITALY

3 - Faculté des Sciences du sport, Université de la Méditerranée, Marseille, FRANCE

4 - Laboratory of Neurobiology and Movements, CNRS, Marseille, FRANCE

INTRODUCTION

In motor tasks which involve the maintenance of equilibrium during standing activities, the total body center of mass (COM) position is a reference value controlled during movement execution [1]. The displacement of a given body segment is associated with a compensatory displacement of other body segments with the aim of guaranteeing the biomechanical equilibrium.

The control of COM projection onto the ground raises the question of how this regulation may be achieved in healthy subject and in presence of modification of any sensory systems (i.e. in presence of below-knee amputation). In the present study, analysis of kinematic synergies was carried out using the Principal Components Analysis (PCA) [2], which allows for identification of both the number and the structure of kinematic degrees of freedom of movement performance.

METHODS

Five male control subjects (age range, 26 to 62) and five male below-knee amputees (BKA) (age range, 24 to 57) participated to the study. In response to a light indicating the side of movement, each subject was instructed to raise as fast as possible the leg in the frontal plane to an angle of 45°. Each subject performed twenty trial (ten for each side) in a randomized order.

The movement kinematics were recorded using the ELITE system (100 Hz). COM displacement was computed in the frontal plan, together with three body angles and their relative range of motion (ROM): α , formed by the supporting leg and the vertical axis; β , formed by the trunk and the supporting leg; ϕ , formed by the raising leg and the vertical axis. PCA was applied to the covariance matrix formed by the temporal series associated with α , β e ϕ .

RESULTS

Two phases of the motor act can be defined: 'preparation phase' (PP), corresponding to postural adjustment preceding the onset of the leg raising; 'movement phase' (MP), corresponding to the leg raising. Regardless of the raising leg, BKA showed a lateral displacement of the COM equally distributed in the two phases (PP=58±10mm; MP=75±18mm), while control group demonstrated a higher COM displacement in MP (63±20mm) with respect to PP (25±12mm). CM shift was mainly due to α or β angles, which showed a higher ROM in PP with respect to control subjects for which body angles variations were mainly observed during MP.

In control subjects, PCA showed that at least 92% of the movement variance could be explained by the first component and that the structure of the first component (on average, α =11%; β =11%; ϕ =98%) was repeatable within each subject. By contrast, when BKA raised the sound leg, a second component were required to explain more than 90% of the total variance. For three of the five subject, the structure of the second component was repeatable and demonstrated the influence within this component of postural angles (on average, α =21%; β =89%).

DISCUSSION

PCA analysis showed that the leg raising performance can be considered a one-degree of freedom movement and that the motor synergy adopted by each subject is repeatable and include in a single motor command the control of voluntary movement and the associated anticipatory postural adjustments.

By contrast, BKA showed a dissociation of the kinematics (COM and body angles) of the motor act in the two phases of the movement (PP and MP). PCA showed a partial uncoupling between postural adjustments and leg movement, with a prevalence of preparatory postural adjustment. The preparatory postural adjustments are responsible for the presence of a second significant component. This uncoupling strategy may reflect the amputee's level of postural adaptation to their severe constraints.

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THE MECHANICS OF THE DOUBLE LEG CIRCLES AT THE POMMEL HORSE: A TWO-SEGMENT COMPOUND CONICAL PENDULUM

P. de Leva

Istituto Superiore Statale di Educazione Fisica, Rome, Italy

The double leg circles at the pommel horse can be performed with different techniques. Typically, the gymnast is in a support position on his hands (Fig. 1), and rotates about a vertical axis, sweeping a conical surface with his body longitudinal axis. In this complex condition of controlled disequilibrium, the gymnast's main goal is to obtain and maintain throughout the performance the largest possible body tilt angle (θ) and center of mass (CM) height. The purpose of the study was to explore the mechanics of these exercises, and define the role of shoulder muscle torques in the achievement of the gymnast's goal.

When the gymnast's front is constantly facing the horse throughout a revolution, all of the gymnast's segments can be assumed to have the same *vertical* angular velocity (ω). From a mechanical standpoint this is the simplest technique, and was selected for a mathematical simulation. The gymnast was modeled as a system composed of two rigid subsystems, joined at the shoulders: "body" (B), and upper limbs (UL; Fig. 2). Both hands were on one handle, that was assumed to be rotating together with the gymnast. The model was assigned the mean inertial parameters of the 1996 Olympic finalists for pommel horse, estimated by scaling mean proportions (de Leva, 1996). B was free to rotate about the shoulders, and UL about the handle. The model was allowed to exploit pure torques of known magnitude about medio-lateral axes: (a) two internal torques by the two subsystems on each other at the shoulders (T_{SB} , T_{SUL}), and (b) one external torque by the horse (Th_{UL}). Three constraints were imposed: the tilt angles (θ , β) of B and UL, and their common ω were kept constant throughout the performance. Accordingly, two free body diagrams were built (Fig. 2). Under the above constraints, the angular momenta (H_B , H_{UL}) of the two subsystems about their CMs depend only on their inertial parameters and the values of θ , β , and ω . Typically the H_B and H_{UL} vectors are not vertical, and sweep predictable conical paths with the same ω as the gymnast. This implies non-null derivatives ($\dot{H}_B$, $\dot{H}_{UL}$), computable for any given combination of values $\theta=\theta_1$, $\beta=\beta_1$, $\omega=\omega_1$. In sum, these derivatives are *needed* to meet the constraints. According to Newton's second law, their known values equal the net torques *needed* by the subsystems to meet the constraints. With $\theta=\theta_1$, $\beta=\beta_1$, $\omega=\omega_1$, the *actual* net torques T_B and T_{UL} are the sum of the given pure torques plus the torques by the forces shown in Fig. 2, obtained by inverse linear dynamics. Only when both $T_B = H_B$ and $T_{UL} = H_{UL}$ the constraints are met. Based on the laws of mechanics, both equations were expressed as nonlinear functions of the kinematic variables (θ , β , ω). If one variable is given a value, a system containing the two functions yields, for the two unknowns, all of the possible pairs of values *meeting the three constraints*. Given either θ or β , the solved system (8th degree, 2nd for ω , 4th for the other unknown) had a maximum of eight pairs of real roots.

When θ was given a value of 66.5° (obtained from film data), Th_{UL} kept null, and β limited to 0° < β < 90°, the results indicated in Fig. 1 were obtained. Condition 4 was obtained with no pure torque. Condition 3 by applying a relatively low shoulder torque (T_{SB} , on body). Conditions 1 and 2 are static positions requiring intense T_{SB} 's. It was concluded that for $T_{SB} \geq 0$ (counterclockwise), and for any given θ , the highest CM position (i.e. the smallest β) is obtained without shoulder torques. But the gymnast's vertical angular momentum (H_Y) had its peak in condition 4. Hence condition 4 is the easiest to maintain ($T_{SB} = 0$), but the hardest to obtain when starting from a static condition ($H_Y = 0$). Surprisingly, with $T_{SB} < 0$ (clockwise) the CM height increased further; however, an even higher H_Y was necessary.

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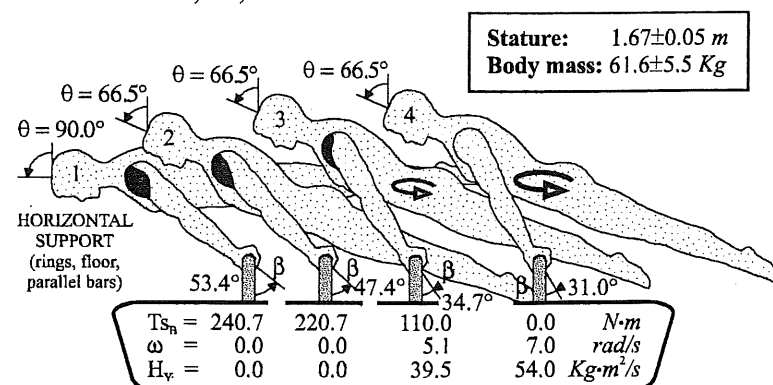


Fig. 1. Effect of shoulder torques (T_{SB} , $T_{SUL} = -T_{SB}$) on system mechanics.

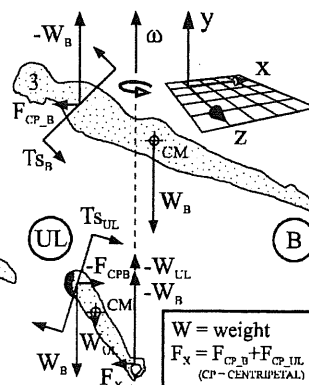


Fig. 2. Free body diagrams.

LIFTING TECHNIQUES AND SOME BIOMECHANICAL BACK VARIABLES - A STUDY ON WOMEN

Katarina Kjellberg, Lars Lindbeck, Mats Hagberg
National Institute for Working Life, Solna, Sweden

INTRODUCTION

Among factors reported to be associated with work related back injuries, one major factor is improper handling technique, but the relationship is not completely elucidated (Bigos et al. 1986). Biomechanical studies of manual handling often deal with handling techniques. Still, there is a lack of proper measuring methods that take account of different aspects of work technique, in particular dynamic aspects and individual performances. Also, only a limited number of studies on women have been reported. The aim of the present study was to investigate whether some selected biomechanical back variables can be used for distinguishing between different lift methods and different performances in lifting tasks. It was decided to use female subjects because of the large number of women working in occupations with high risks for back injuries at handling tasks, e.g. nursing occupations.

METHODS

Twelve female subjects lifted a box with two lift methods, leg lift and back lift, and two lift velocities, a fast lift of approximately 1 s and a slow of 2 s. Their ages were 22-60 years (mean=39 years). Nobody did lifting tasks in their daily work and nobody had any ongoing musculoskeletal symptoms. Body movements, ground reaction forces and activity of the erector spinae muscle were registered by means of an optoelectronic three dimensional motion analysis system (the MacReflex system), a force plate and electromyography. The lifts were analysed with a dynamic biomechanical model. The dependent variables selected from the measurements are shown in Table 1. Analyses of variance with repeated measures were performed to test for differences due to lift methods and lift velocities for all dependent variables.

RESULTS

The range of trunk flexion was naturally greater during back than leg lifts ($p<0.0001$) (Table 1). For the leg lifts a slightly larger trunk flexion was obtained during fast lifts in comparison with slow lifts ($p=0.029$). The peak angular velocity was larger during back lifts than during leg lifts ($p<0.0001$), and larger during fast lifts than during slow lifts ($p<0.0001$). The peak trunk acceleration was about the same during leg lifts as during back lifts, but of a higher magnitude during fast lifts than during slow lifts ($p<0.0001$). The peak of the L5/S1 net moment was higher for the fast back lifts than for the fast leg lifts. No general effect of lift method was obtained though. The moment was higher at fast lifts compared with at slow lifts ($p<0.0001$). No method effect was found, neither for the mean EMG amplitude, nor for the peak EMG amplitude. Both EMG variables showed higher values during fast lifts than during slow lifts ($p=0.025$ and $p=0.0068$).

Table 1. Values for selected dependent variables for the different lift methods (back and leg) and different lifting velocities (fast and slow). Mean values and standard deviations (in parenthesis) are shown.

Variables	Back lifts				Leg lifts			
	Fast		Slow		Fast		Slow	
Trunk angular displacement (degrees)	91	(4.5)	91	(4.2)	59	(7.3)	56	(6.7)
Peak trunk angular velocity (rad/s)	3.5	(0.43)	2.3	(0.40)	2.8	(0.58)	1.8	(0.23)
Peak trunk angular acceleration (rad/s ²)	15.7	(3.0)	7.7	(1.8)	15.1	(4.0)	7.5	(1.7)
Peak L5/S1 net moment (Nm)	166	(22)	134	(16)	160	(22)	136	(19)
Mean EMG erector spinae (%RVE*)	242	(147)	207	(75)	197	(101)	183	(82)
Peak EMG erector spinae (%RVE)	486	(337)	369	(145)	423	(272)	346	(204)

*%RVE = percentage of Reference Voluntary Electrical activation

DISCUSSION

The trunk angular displacement is an obvious variable to describe lift methods with different back movement patterns. The peak trunk angular velocity seemed to be the most valuable variable for distinguishing between lift methods. To distinguish between the two lift velocities, the peak trunk angular velocity and acceleration, the peak L5/S1 moment and the mean and peak EMG amplitude were useful. The results imply that the dynamic factors had larger influence on the lower back load than the squat and stoop posture. Both trunk angle and trunk velocity have been associated with risks for low back disorders (Marras et al. 1993). Regarding the L5/S1 moment and the EMG measures, several studies have applied these variables on comparisons of the squat and stoop method with varying results. Different study designs and difficulties to control for variations in EMG measurements during large trunk movements could explain the disagreements. In conclusion, trunk angular velocity seemed to be the most powerful variable for assessing work technique.

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KNEE JOINT KINEMATICS DURING SIDESTEPPING - GENDER COMPARISONS

S.G. McLean[†], R.J. Neal[†], P. Myers[†] and M. MacDonald^{*}

[†]Dept. of Human Movement Studies, The University of Queensland, QLD, AUSTRALIA

^{*}Dept. of Human Movement Studies, Queensland University of Technology, QLD, AUSTRALIA

INTRODUCTION

A disproportionately higher incidence of sports related knee, and in particular anterior cruciate ligament (ACL) injuries is found to occur in females compared to males. Sex differences such as those pertaining to joint structure, ligament laxity and joint biomechanics during specific movements have been proposed as reasons for this imbalance (Hutchinson and Ireland, 1995). The majority of these factors (e.g., intercondylar notch width) and their possible link to injury have been examined (Laprade and Burnett, 1994). There is however, a paucity of data describing sex specific knee joint kinematics during "high risk" movements. Sidestep cutting, an evasive manoeuvre involving an athlete's rapid lateral deviation from a previous direction of motion, is proposed as a possible mechanism non-contact ACL injury. While knee joint kinematics during the sidestep cut have been quantified previously for males (McLean et al., 1997), similar data on female athletes do not exist. The purpose of the current study was to compare knee joint kinematics of male and female athletes during typical sidestep cuts in the hope of providing insight into why females display a higher incidence of non-contact ACL injury.

METHOD

Thirty participants (male=16, female=14) proficient in the sidestep cut, had knee joint kinematics recorded during the stance phase of two gait conditions (straight run, cut at 30°- 50° to horiz.) at speeds between 5 m.s⁻¹ - 6.5 m.s⁻¹. Motion of precisely attached external markers (lower limb) was digitised at 200 Hz using a four-camera Motion Analysis system. Video data were captured bi-laterally over six trials for each condition and the 3D coordinates were generated. These data were applied to custom software based on Grood and Suntay's (1983) joint coordinate system that allowed calculation of the three clinical knee joint rotations (flex/ext, ABd/ADd, ext/int rotation). Mean $\pm$ SD maximum and minimum angular displacements and subsequent ranges of motion were determined for these rotations for males and females. Dependent measures were subsequently analysed via a two-way (sex X gait) repeated measures ANOVA, setting $\alpha = .05$.

RESULTS AND DISCUSSION

Figure 1 illustrates male and female mean ranges of motion for the three clinical rotations during the stance phase of sidestep cutting. Maximum, minimum and range of motion were found to be similar across gender for flexion/extension and external/internal rotation. Female subjects however, displayed significantly ($P<.05$) larger maximum and minimum knee abduction angles when compared to males. Larger Q angles present in females, causing increased genu valgum on foot contact, may explain this difference.

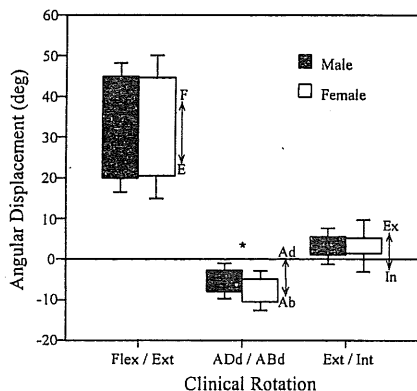


Figure 1. Mean ranges of three knee joint rotations during sidestepping.

The increased ABd/ADd observed for females, appears to have little clinical significance, with all three rotations being well within a "safe" range of motion. Of concern however, was the fact that females displayed significantly greater inter and intra subject variability for external/internal rotation. A lack of consistency between trials, may offer the potential for injury to structures such as the ACL, with the performance of an abnormal and possibly hazardous cut being more likely. Increased kinematic variability has been suggested to result from low levels of conditioning and experience (Hutchinson and Ireland, 1995). While females in the current study were elite, they had participated in their respective sports 4.7 $\pm$ 0.4 years less than their male counterparts and underwent less stringent training regimens. Further analyses led to the conclusion that level of experience, rather than gender, was the major contributing factor to the degree of inconsistency in knee kinematics displayed by individuals during sidestep cutting.

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COMPARISON OF THE MECHANICAL LOAD DURING PUSHING AND PULLING OF TWO-WHEELED CONTAINERS

B. Schibye¹, K. Sogaard¹ & K. Klausen²

¹ National Institute of Occupational Health, Copenhagen, Denmark, ² August Krogh Institute, University of Copenhagen, Copenhagen, Denmark.

INTRODUCTION

Lifting and carrying tasks are now often replaced by pushing and pulling tasks. The compression of the spine during pushing in front of the body has been found to be smaller compared to pulling in front of the body. However, in many situations the worker is pulling e.g. a container behind the body while he is walking forward. The aim of the present study was therefore to compare this situation with the pushing situation.

METHODS

Seven waste collectors were pushing a two wheeled container in front of the body or pulling the container behind the body with two hands. Forces were measured at the hands with a 3-dimensional force transducer (AMTI). The simultaneous working postures were analyzed from video recordings by digitizing the joint centres using Peak Performance. The compression forces at the L4/L5 level and the torques at the shoulder joints were calculated using a 2-dimensional sagittal static link segment model (Watbak 5.1). The calculations were performed when the push/pull force was highest in the period with constant velocity. EMG from the back and the abdominal muscles were obtained with surface electrodes.

RESULTS

Table 1: Mechanical load in pushing/pulling containers of different weight. Results are presented as mean $\pm$ SD. Torque I: Total external torque at L4/L5. Torque II: Torque produced by the body mass above L4/L5. Torque III: Torque produced by the push/pull force. Positive value: flexion torque, negative value: extension torque.

	Container weight kg	Pull/push force N	Compression L4/L5 N	Torque I Nm	Torque II Nm	Torque III Nm
push	40	98 $\pm$ 19	542 $\pm$ 167	-4(n=1) 15 $\pm$ 10(n=6)	44 $\pm$ 12	-32 $\pm$ 10
	65	113 $\pm$ 29	611 $\pm$ 261	19 $\pm$ 14	57 $\pm$ 17	-38 $\pm$ 17
pull	40	111 $\pm$ 31	938 $\pm$ 192	26 $\pm$ 12	30 $\pm$ 11	-4 $\pm$ 2
	65	142 $\pm$ 24	1089 $\pm$ 168	34 $\pm$ 11	37 $\pm$ 8	-3 $\pm$ 6

The compression force of the spine at L4/L5 level was low in all situations (tab.1), and significantly higher during pulling than during pushing. In both situations, the calculated total external torque at L4/L5 was found to act in flexion, and the back muscles were active in both situations. This flexion torque was significantly lower during pushing than during pulling. The total torque is the result of the flexion torque produced by the mass of the arms and the forward inclined upper body, and the torque produced by the push/pull forces. The flexion torque caused by the mass of the upper body and arms is highest during pushing, so the explanation for the lower total flexion torque in this situation is the higher extension torque produced by the pushing force. Both the pushing and the pulling force produce an extension torque at L4/L5, but since the lever arm for the force is shortest during pulling, the produced torque is small. The lower compression during pushing is further explained by the fact that the pushing force per se produces a traction of the spine since the direction of the pushing force is partly downwards, whereas the pulling force produces a compression. The flexion/extension torque at the shoulder joints was small (below 9 Nm) in all situations and no difference was found between pushing and pulling.

CONCLUSION

During pulling the compression is higher than during pushing. In both situations the back muscles are active, but since the external flexion torque is highest during pulling, the muscle activity results in the highest compression force in this situation.

Acknowledgement

The present study is part of the 1993-98 research programme Waste Collection and Recycling, which is supported jointly by the Danish Ministry of the Environment and the Ministry of Labour.

KINETICS AND ENERGETICS DURING LOAD CARRYING

G. Sjøgaard¹, B. Laursen¹, D. Ekner¹, E. Simonsen² & M. Voigt³

¹ National Institute of Occupational Health, Copenhagen, Denmark, ² The Panum Institute, Copenhagen, Denmark, ³ The University of Aalborg, Aalborg, Denmark.

INTRODUCTION

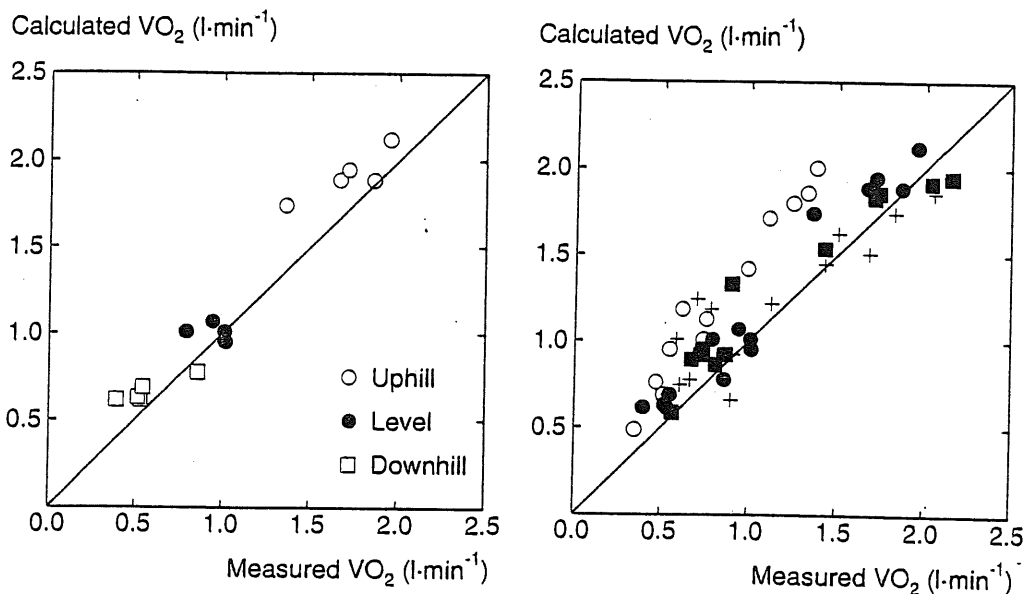
During physically heavy work tasks the musculoskeletal tissues are exposed to mechanical as well as metabolic loadings. The metabolic load is due to the active development of mechanical energy by conversion of chemically bound energy in the muscle tissues under the consumption of oxygen. This implies a causal relationship between kinetics and energy turnover in terms of oxygen uptake. The aim of the present study was to validate a biomechanical model for prediction of whole body energy turnover from kinematic and anthropometric data during load carrying.

METHODS

Loads of 0, 5, 10 kg were carried symmetrically in both hands (S) comprising total loads of 0, 10, and 20 kg, respectively and additionally, 10 and 20 kg were carried asymmetrically in one hand only (A). While carrying these loads the subjects were walking on a treadmill (approx. 5 km/h) at horizontal level, uphill, or downhill the slopes being approx 5 degrees. Each of the 6 subjects performed the 5x3 different exercise bouts once while oxygen uptake was measured and once while film recordings were obtained. These were converted to video recordings and analysed by the Peak Performance system. Based on a link segment model (Laursen 1996) and assumptions on efficiencies during concentric and eccentric exercise, respectively (Norman 1989), the total energy turnover rate was calculated in terms of oxygen uptake assuming an "energy content" of 20 kJ per 1 liter of oxygen uptake. A resting oxygen uptake of 0.25 l·min⁻¹ was added to each calculation. These data were compared to the corresponding values of oxygen uptake measured directly.

RESULTS

Individual data for one subject are given in the figure to the left with different signatures for the 3 slopes. The highest value was found for 20 kg carried asymmetrically uphill and the lowest for walking downhill with no load. Data for all subjects are given in the figure to the right together with the line of identity.



CONCLUSION

The close relationship between energy turnover rate measured directly and estimated based on a biomechanical model justifies to assess the metabolic load from kinematic data. Mechanical as well as metabolic loads represent risk factors for the development of occupational musculoskeletal and/or cardiovascular disorders. In combination these risk factors may be evaluated from video recordings.

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ANALYSIS OF THE POWER BALANCE DURING WHEELCHAIR PROPULSION USING A 3-D MUSCULOSKELETAL SHOULDER AND ELBOW MODEL.

Frans C.T. van der Helm¹, H.E.J. (DirkJan) Veeger², Leonard A. Rozendaal².

¹ Dept. of Mechanical Engineering, Delft University of Technology, The Netherlands; ² Dept. of Human Movement Sciences, Vrije Universiteit Amsterdam, The Netherlands.

INTRODUCTION - The energetic efficiency of manual wheelchair propulsion is very low (external power/-consumed power is about 6-10%; Veeger et al., 1991) compared to e.g. running or cycling (about 20-25%). Woude (1989) suggested that this could be due to the discontinuity of force application, relatively large recovery phase, synchronous force application and/or trunk movements. Individual muscles have a maximal efficiency of about 36% at 30% of their maximal contraction velocity. Obviously not all muscles can contract at their optimal velocity during the whole propulsion cycle. An additional complicating factor is that in the upper extremity the kinematic redundant Degrees-of-Freedom must be stabilized, i.e. the moment must be balanced. This requires additional muscle effort, sometimes at zero (isometric) or negative (eccentric contraction) efficiency. The power exertion or absorption of individual muscles and the power exchange with kinematic energy of the moving segmental masses and with energy release for propulsion is a start for the study of the low propulsion efficiency. Goal of this paper is to analyse the power balance during a wheelchair propulsion cycle using a large-scale musculoskeletal model of the shoulder and the elbow.

METHOD - At a stationary wheelchair ergometer, two subjects were measured at four different speeds (0.56 - 1.69 m/s) and four different load levels (10%, 20%, 30%, 40% Vo_{2max}). Motions were recorded using a 3-D camera system (VICON). Trunk, humeral and forearm motions were measured using marker trees attached to these segments. Scapular and clavicular motions were estimated using regression equations previously derived from static measurements for each subject. External forces and moments at the hand could be measured due to the special design of the ergometer. Recorded motions (position, velocity and acceleration) and external forces at the hand were input variables to the musculoskeletal model.

The musculoskeletal model consist of the thorax, clavicle, scapula, humerus, ulna, radius and hand, and all 31 muscles and muscle parts which are crossing these joints. The shoulder part of the model has been published before (Van der Helm, 1994; Van der Helm & Veeger, 1996). The elbow part has been added recently based on data obtained by Veeger et al. (1997). The model contains 9 kinematic DOF (thorax position and rotation, wrist rotation) and 8 dynamic DOF (3 DOF at the shoulder girdle, and the glenohumeral, humero-ulnar and ulno-radial joints). Muscle forces were calculated by optimization of the sum of squared muscle activations, weighed by the muscle volume. This criterion allowed for a reasonable approximation of the energy consumed during dynamic as well as static muscle contraction. Muscle dynamics were accounted for using an inverted muscle model.

RESULTS - At every 12 msec during one complete propulsion cycle the motion equations were evaluated and muscle forces optimized. Since all mechanic variables in the model are known, a power balance at each time instant could be made. Sources of power in the system are the external power applied at the wheelchair rim (propulsion force times hand velocity, hand moment times hand angular velocity), gravitational power (moving masses with respect to the gravity field), kinematic power (accelerating/decelerating masses and moments of inertia), power exerted by the kinematic degrees-of-freedom like the thorax and wrist motion (generalized forces times velocity), and power generated by the muscles (muscle force times contraction velocity). The major power generators are the m. pectoralis major and m. serratus anterior in the shoulder. At the elbow less power is generated. The trunk contributes significantly to the power. During the forward motion of the trunk kinetic power is released, during the backward motion of the trunk the kinematic DOF generate power.

DISCUSSION - During hand contact, the propulsion power is the major negative source of power (taking energy out of the system). There is very little change in kinetic power of the segments, and the muscles are the major power generators. Almost no negative muscle power is generated, indicating that almost no eccentric contraction occurs. Certainly a number of muscles are below their optimal contraction velocity ($\sim 0.3 V_{max}$), which is one reason for the low efficiency. During the recovery phase muscles are dissipating and generating kinetic energy (decelerating and accelerating the segments. Especially much energy is 'stored' in this phase in the potential energy of the thorax (moving upwards). In the future, it will be attempted to estimate the power consumed by the muscles in order to analyse the efficiency of individual muscles.

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Three-Dimensional Head-Neck Model for Impact and High-Acceleration Studies

Shyh-Chour Huang

Department of Mechanical Engineering,
National Kaohsiung Institute of Technology,
Kaohsiung, Taiwan 807, R.O.C.

Introduction

In this paper, a flexible and inexpensive technique that employs multibody dynamics analysis is used to model and simulate head-neck system. A user-oriented computer software simulator, which can run on a personal computer, was also developed. This software simulator is intended to be more "user friendly" and reliable than the earlier simulators.

Methods

This model can have as many as 54 degrees of freedom (3 for the translations and 3 for the orientations of the bodies relative to their adjacent bodies). Each joint has an internal force and moment to constrain its motion. These forces and moments may be modeled as two-parameter, viscoelastic solids. The force-displacement and moment-rotation relation is taken as:

$$F = -(k_f x + c_f \dot{x}) \quad (1) \quad M = -(k_m \theta + c_m \dot{\theta}) \quad (2)$$

The Kane method [1] state that the sum of the generalized active and passive forces is zero for each coordinate x_b .

$$F_b + F_b^* = 0 \quad (3)$$

$$a_{bp} x_p = f_b \quad b = 1, \dots, 54 \quad (4)$$

where the a_{bp} and f_b are

$$a_{bp} = m_k v_{kbn} v_{kpn} + I_{knn} \omega_{kbn} \omega_{kpn}$$

$$f_b = F_b - (m_k v_{kbn} \dot{v}_{kpn} \dot{x}_p + I_{knn} \omega_{kbn} \dot{\omega}_{kpn} \dot{x}_p + e_{rsn} I_{ksn} \omega_{kbn} \omega_{kqr} \omega_{kpn} \dot{x}_q \dot{x}_r) \quad (5)$$

Equations (4) then represents a system 54 simultaneous, second-order ordinary differential equations. Then may be integrated numerically using a standard integration algorithms. Such algorithms have been constructed and compiled into a Fortran based computer program.

Results

The paper presents the modeling of the head-neck and its simulation. Figure 4 show that there is a good comparison between the experimental results and those predicted by the computer model. It suggests that the model represents one of the most sophisticated head-neck models available. However, more testing and improving needs to be done. Specifically, the three-dimensional features of the model need to be further checked with experimental data. Also, better experimental values for the soft tissue biomechanical properties need to be obtained. An principal application of the modeling presented herein is with whole-body simulator. The head-neck model which simulates gross-motion head-neck dynamics can be used to determine dynamic forces on the upper torso during crashes. Furthermore, criteria for injury need to be established.

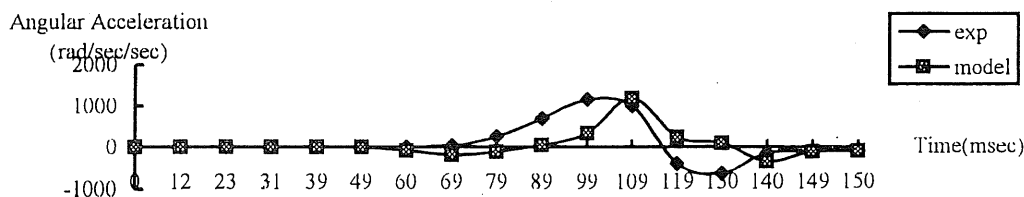


Figure 4. Comparison of Analytical Model and Experimental Angular Acceleration of the Head.

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A THREE-DIMENSIONAL DYNAMIC MODEL OF THE HUMAN NECK

N.Bertholon ^{ox}, S.Robin ^x, J.Y.Le Coz ^o, W.Skalli ^x, F.Lavaste ^x

^oLaboratoire d'Accidentologie et de Biomécanique, PSA Peugeot Citroën - RENAULT, Nanterre, France

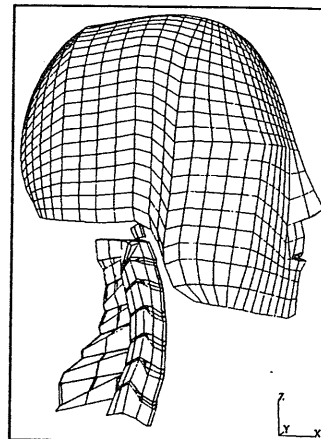
^xLaboratoire de BioMécanique, ENSAM, Paris, France

Introduction :

There are two aspects to the biomechanics of the neck in automotive crashes. On the one hand, the neck governs the movement of the head and so the understanding of its behaviour is important in the evaluation of head injury. On the other hand, mechanisms causing injuries to the neck itself are not fully understood and both experimental and numerical studies may improve the comprehension of these mechanisms. This study describes the development and validation of a numerical dynamic model of the neck based on the work of Dauvilliers¹. This model is part of a human body model which helps to better understand and to prevent car occupant injuries.

Method :

The three-dimensional human neck model was developed using the Radioss[®] explicit code. Due to the fact that it is a part of a global model the time step has to be kept to a minimum. Thus each vertebra from T1 to C2, and the head, were considered as rigid bodies. Only the geometry of the main mechanical elements of the vertebrae was taken into account (i.e. vertebral bodies, spiny apophysis and articular apophysis). Between the vertebrae, intervertebral discs are modeled as 8-nodes elements, ligaments and muscles are combined as spring-damper elements and articular facets as contact interfaces. The links between the axis, the atlas and the head are very complex and therefore are modeled by two spherical joints. The first of these joints, between C1 and C2, enables vertebra C1 to realize rotate around the axis of the odontoid and allows flexion-extension movement. The second, between head and C1, is located above the middle of the occipital condyles. Existing properties were introduced into the model where possible and the other parameters were assessed using a parameter identification process. Mass and inertia characteristics of the head and the neck are taken from the literature.



The model response to frontal, lateral and oblique impacts was verified using the head-neck responses of human volunteers subjected to sled acceleration impacts performed at the Naval Biodynamics Laboratory². The velocity of vertebra T1 was used as an input to the model to simulate the impacts. The model responses are compared with volunteer response corridors. The corridors used are for linear accelerations of the anatomical center of the head, angular accelerations of the head, head rotations (Bryant angles) and displacements of the anatomical center of the head relative to the T1-vertebral body.

Results :

A good correlation is obtained between model responses and experimental corridors for all impact directions and throughout the duration of the impact (loading and rebound phases up to 400 ms). The model reproduces the aspects and maximum values of linear and angular head acceleration curves for all impact conditions. The main rotations of the head about the X, Y and Z axes agreed with corridors in all impact directions too (less than 4% error on maximum values). The displacements of the head's anatomical center about all axes are satisfactory in terms of their aspects. However, some maximum values are higher or lower than experimental ones, about the X axis in frontal direction (+12%), about the Y axis in oblique and lateral directions (-9 and -12 %) and about the Z axis in frontal, oblique and lateral directions (-24, -32 and -12% respectively). In spite the lack of experimental results in rear-end impact, initial simulations give good agreement in terms of maximum values in rotation of the head. Although active muscle response is not taken into account, the rebound phase of the head is simulated well by the model with very high damping coefficients.

Discussion :

The model response to frontal, lateral and oblique impact agreed with volunteer responses, and simulations in rear end impact are encouraging. Thus, kinematic response of the head relative to the torso is well simulated. However, the dynamic motion of each vertebra is not given by existing experiments. Intervertebral displacements at impact are needed in order to simulate neck kinematics and thus be able to evaluate neck injuries. As the model does not predict injuries, a refined version will be developed by considering the muscles independently from the ligaments as in the de Jagger model³. Further research work will be focused on experimental testing on cadavers in rear-end impact with the recording of vertebra kinematics and modeling of injury mechanisms.

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METHOD FOR OPTIMAL CALCULATION IN SWING PHASE MOTION OF TRANS-FEMORAL PROSTHESIS

Y. Suzuki, E. Genda, T. Kasahara, and Y. Tanaka
Rosai Rehabilitation Engineering Center, Nagoya, Japan

INTRODUCTION

Although trans-femoral prosthesis motion simulation has been carried out in the past for the purposes of design and evaluation, the emphasis has been on the functional characteristics of the prosthesis, and not on how the amputee controls the prosthesis with respect to its functions. There is a need to know how the amputee controls the prosthesis in order to optimize prosthetic design. We therefore constructed a sagittal-plane musculoskeletal model of a trans-femoral residual limb using amputee MR Images and hip-joint muscular force measurements, and combined it with a computer model of a trans-femoral prosthesis so that functional abilities of trans-femoral amputees could be evaluated through computer simulation. Using this model, we performed swing-phase motion optimization, and verified the method of optimization by solving for minimum swing phase time, maximum walking velocity.

METHODS

A sagittal-plane musculoskeletal model of the trans-femoral residual limb was constructed. The iliopsoas, rectus-femoris, gluteus-maximus, and hamstring muscles were selected to be included in the model as major muscles that contribute to hip-joint flexion and extension. Points of origin and insertion for these muscles were estimated using MR Images of the subject, and muscles lengths were determined through hip-joint angle calculations. Model parameters such as the maximum force of the contractile element for each muscle were obtained from isometric force measurements of the subject.

The residual limb and socket was considered one rigid body, and below the knee was modeled as one segment because there was assumed to be no foot motion relative to the shank during swing phase. Accordingly, the prosthesis model was a rigid body composed of two links, with the knee joint mechanism placed between the two links. The LAPOC-M730 (Imasen Engineering, Japan) knee joint was the one used in the model, characterized by an extension-assist spring and a pneumatic cylinder to smooth knee flexion and extension movements.

The joint angles and accelerations of the lower limb for toe-off and heel-strike were obtained from gait analysis. In order to minimize terminal impact which occurs at maximum knee extension during swing phase, an inequality state variable constraint was used to keep the knee-extension angular velocity within a reasonable range. In order to be able to solve as a minimum time problem, swing phase time was considered the performance index.

RESULTS

Figure 1 shows both the results of motion analysis and simulation calculations for a trans-femoral amputee during swing phase, with the angle in degrees from vertical of the hip and knee joints. The actual measured minimum time for swing phase was 0.42 seconds, but the calculated minimum time was 0.44 seconds. The knee-extension angular velocity boundary condition which had been used as a state variable constraint was at a maximum at the time of maximum extension. Although neural excitation, a control input, was present in hip flexors at early swing phase and in hip extensors at late swing phase, neural excitation was also present in hip flexors immediately prior to maximum knee extension.

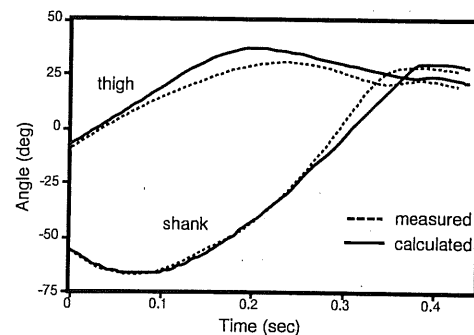


Fig. 1 Angle of lower extremity in swing phase (measurement v.s. calculation)

DISCUSSION

It can be concluded that the calculation methods used are reasonable from the proximity of calculated data to the measured data. As a swing phase minimum time problem, a simple bang-bang type control was predicted with hip flexors active during early swing phase and hip extensors active during late swing phase, but the results indicated that the hip flexors were also active immediately prior to maximum knee extension. This was seen as an effort to reduce terminal impact, which in the simulation was controlled by using a state variable constraint for the knee-extension angular velocity. The swing phase actually gets shorter when calculations are made without the boundary condition. From this, it can be concluded that the air damping mechanism in the prosthesis is not functioning fully for this subject.

Although only the swing phase time was reported here, other unknown parameters such as the diameter of the air cylinder orifice can be found, allowing optimal prosthetic design for this subject. If, for example, the objective function was minimizing energy consumption, optimal values for more comfortable walking could be found.

THREE DIMENSIONAL FINITE ELEMENT MODEL OF THE CEMENTED REPLACED FEMUR. EFFECTS OF THE METAPHYSEAL FILLING ON THE FEMUR MECHANICAL BEHAVIOR.

Astoin,E. ; Simondi,M. ; Lavaste,F.

Laboratoire de Biomécanique, ENSAM Paris, France.

INTRODUCTION. In cemented fixation, a better metaphyseal filling has been proved to decrease the magnitude of shear stresses transmitted to the cement mantle¹. Based on this concept, a new monoblock femoral prosthesis was developed, using an anatomical study to provide adequate filling in the metaphyseal zone for the great majority of currently implanted femurs. A numerical simulation based on the finite element method was developed to point out the effect of this metaphyseal filling on the rotational stability and the mechanical behavior of the femur.

METHOD. A morphometric analysis was realized on 200 femurs. Measurements were performed with a digitizing table on frontal roentgenograms. A statistical study pointed out the main parameters to define a set of femoral stems optimizing the contact with the internal metaphyseal cortical bone. Beside, a finite element model was developed, using the ANSYS® package. Frontal and sagittal roentgenograms of a femur were used to determine the diaphyseal and epiphyseal shapes of the cortical bone. A standard cortical metaphyseal shape was deformed with a krigage algorithm² to be adapted to the global shape. A prosthesis of the adequate size, called reference prosthesis, was inserted inside the femur, and the cement was defined in the space between the prosthesis and the bone. Two models were there realized. First, one with exact prosthesis/bone contact on the internal metaphyseal cortical bone (figure 1-a), representing a theoretical case. Second, replacing the exact contact by a very thin cement mantle. The bone/cement interface was modeled linearly, while the prosthesis/cement interface was modeled with contact elements. In order to validate the influence of the metaphyseal exact-fit in cemented femoral fixation, we realized three other numerical models, reducing the stem in its frontal metaphyseal width of respectively 10%, 20% (figure 1-b) and 30% from the reference. The tests were realized by fixing the distal diaphyseal cortical nodes, and applying 3D physiologic efforts both on the prosthesis head and the greater trochanter. Then we compared the stem rotations by measuring displacement of the prosthesis head in the sagittal plane. The stress-shielding effect was evaluated by measuring the Von Mises Stresses in the cortical bone.

RESULTS. The movement of the prosthesis head are equivalent for the two tests with the reference prosthesis. But for the 10%, 20% and 30% reduced stem it is increased from the movement of the reference stem head by respectively 6%, 9% and 16%. Regarding to the internal cortical bone, the reference prosthesis with an exact contact have the higher Von Mises stresses. For the 4 tests without direct prosthesis/bone contact, the Von Mises stresses are more than 60% less important: 66% inferior for the reference prosthesis, and respectively 76%, 79% and 81% for the 10%, 20% and 30% reduced stems.

DISCUSSION AND CONCLUSION. The metaphyseal filling for cemented prosthesis is a new concept comparing to Charnley's experience. Then, the evaluation of its effects needs to be pointed out, and the results presented below are a first step: The best results for torsional stability are interesting, as it is a major problem for cemented stems. Besides, although the rigidity of the stem alone is increased, a stem with larger frontal metaphyseal width and closer contact with the bone increases the stresses transmitted to the bone, reducing the stress-shielding threat. This is closely dependent of the type of interface modelling. Actually, a numerical model can not be the single element to validate a prosthesis, but it allows a better understanding of the behavior of a cemented replaced femur. Improvement are then possible.

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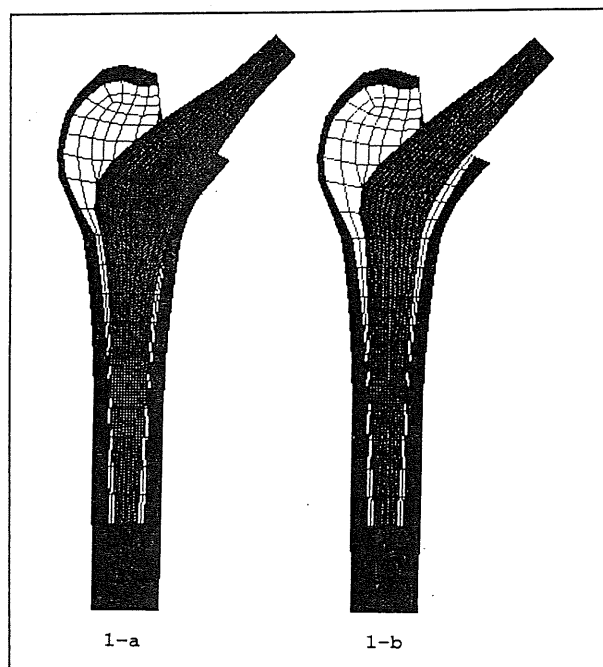


figure 1: representation of the model with the reference prosthesis and exact bone/prosthesis contact (a) and of the model with the 20% reduced prosthesis (b)

3-D FINITE ELEMENT ANALYSIS OF PROSTHETIC JOINT IMPLANTS FOR THE THUMB

K Hollerbach, Computational Biomechanics Group, Lawrence Livermore Laboratory, Livermore, USA

S Perfect, Computational Biomechanics Group, Lawrence Livermore Laboratory, Livermore, USA

A Hollister, Louisiana State University Medical Center, Shreveport, USA

M Truman, ArthroMotion, Warsaw, USA

E Ashby, Computational Biomechanics Group, Lawrence Livermore Laboratory, Livermore, USA

INTRODUCTION

Nonlinear, three-dimensional (3-D) finite element modeling and analysis are powerful new tools for the evaluation and design of joint replacement prostheses. We used this technology to model several different implant designs and to perform 3-D, nonlinear, large displacement analysis that replicates loads that would be experienced by the prosthetic implant *in vivo*. Material failure is common in implant designs containing polyethylene components. Wear and fatigue are correlated with the pattern and magnitude of stresses experienced by the material. Orthopaedic implant materials, especially polyethylene, have nonlinear stress-strain relationships. Polyethylene wear occurs primarily in sub-surface regions, which cannot be adequately modeled in two-dimensional (2-D) analyses. In addition, implants are three dimensional and experience forces in all dimensions in the patient. Fully 3-D, nonlinear modeling tools can produce more accurate results than the 2-D, linear analyses.

METHODS

Finite element models of three thumb carpometacarpal joint prostheses were developed using TrueGrid (XYZ Scientific Applications, Inc.), and NIKE3D (Lawrence Livermore National Laboratory) on a Silicon Graphics Indigo² workstation. 3-D surface descriptions of the implants were obtained from Initial Graphics Exchange Specification (IGES) format descriptions generated by the orthopaedic industry using computer aided design tools. Finite element models were developed for each implant design by generating hexahedral meshes that closely approximated the surface descriptions. Material properties for the implant materials (e.g., polyethylene, CoCr, and titanium) were assigned based on data available in the literature. Boundary conditions, defining net joint forces for a specified set of grasps, were calculated based on simulations derived from a hand model originally developed by Giurintano et al. (1995).

RESULTS & DISCUSSION

The implant designs modeled include (1) a ball and socket style joint with a polyethylene socket and a CoCr ball and stem, (2) a ball and socket style joint with a titanium backed polyethylene socket and a CoCr ball and stem, and (3) an asymmetric saddle-shaped joint, in which the metacarpal component consists of a titanium stem with a polyethylene articular surface and a CoCr trapezial component. 3-D, nonlinear finite element analyses of these models show articular surface and sub-surface stress patterns when the implants are loaded with forces representative of common grasping modes (e.g., key pinch, tip pinch, and screwdriver grasps). These loads are static and eccentric to the joint components. Metal backed and unbacked polyethylene socket models varied in stress patterns and levels of maximum stress. The implant stress analyses appear to correlate with failures observed clinically in the ball and socket designs, where metal backed sockets have been shown to be more subject to polyethylene failure. Both ball and socket designs were also compared to the newer design with an asymmetric saddle-shaped implant. With similar eccentric loading, stress patterns varied between implant designs, and maximal stresses were lower in the asymmetric implant.

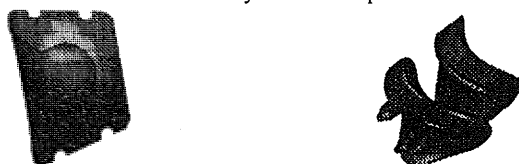


Figure 1: (a) ball & socket joint stresses (b) saddle joint stresses.

3-D, nonlinear finite element analysis should contribute to an improved understanding of the mechanism of implant failure in patients and to improved product design. This modeling can easily be applied to other types of joint implants such as total hip replacements.

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Acknowledgments

This research was supported in part by the United States Department of Energy under contract with the Lawrence Livermore National Laboratory.

A 3-D FINITE ELEMENT MODEL OF THE HUMAN INDEX FINGER

K. Hollerbach, Computational Biomechanics Group, Lawrence Livermore Laboratory, Livermore, USA

S. Perfect, Computational Biomechanics Group, Lawrence Livermore Laboratory, Livermore, USA

A. Hollister, Louisiana State University Medical Center, Shreveport, USA

J. Pearlman, Science & Engineering Research Semester, Lawrence Livermore Laboratory, Livermore, USA

B. Beaubien, Science & Engineering Research Semester, Lawrence Livermore Laboratory, Livermore, USA

INTRODUCTION

Human joint biomechanics are inherently three dimensional (3-D), nonlinear processes that cannot be realistically modeled in two-dimensional (2-D) analyses that have previously been attempted. The quality of the analysis is influenced by the ability to use 3D nonlinear techniques, the anatomic and material models, meshing techniques, and the ability to simulate in vivo joint loading conditions.

We present a three-dimensional (3-D) finite element model of the index finger, including bones, ligaments, and tendons of the DIP, PIP, and MP joints. This research is part of a model of the entire hand, which currently includes the index finger and thumb, as well as all bones of the hand and wrist. Related work in this area includes finite element models of other human joints and prosthetic joint implants (Hollerbach et al., 1996).

METHODS & RESULTS

The finite element model of the index finger was developed using VISU (Lawrence Livermore National Laboratory, LLNL), TrueGrid (XYZ Scientific Applications, Inc.), and NIKE3D (LLNL) software on a Silicon Graphics Indigo² workstation. The finger model is based on a CT dataset with an effective pixel size of 0.15 by 0.20 mm, with a z-axis step size of 0.17 mm. The dataset obtained provides the high resolution required for joint surface definition and facilitates semi-automated segmentation of surfaces, i.e., delineation and labeling of individual tissue types. 3-D surfaces were generated from the segmentation results, and hexahedral meshes were developed to model the bone and joint geometry. 3-D surfaces were generated from the segmentation results, and hexahedral meshes were developed to model the joint geometry. Boundary conditions, such as loading forces, and material properties for the joint tissues are taken from the literature. The finite element model (Fig. 1) can be used to perform static simulations of the finger, in which the joints are placed in a position for a particular grasp, and the corresponding joint reaction forces are applied. These forces can be applied either directly to the bones, or they can be applied to the bones via the tendons. In addition, we are able to simulate 3-D dynamic articulation in which joint motion is determined entirely by the loading conditions (representing tendon forces actuating the joint and ligament forces constraining the joint) and by the joint geometry. The kinematic mechanism is determined by the 3-D geometry of the articular surfaces. In static grasps, joints are loaded eccentrically, as they are in human joints. Results obtained from the simulations include the kinematics, which can be compared against experimental data, the articular surface stresses, and the ligament stresses. Joint kinematics and tissue stresses are visualized on a workstation, where color maps are used to easily distinguish regions of high stress in the soft tissues.

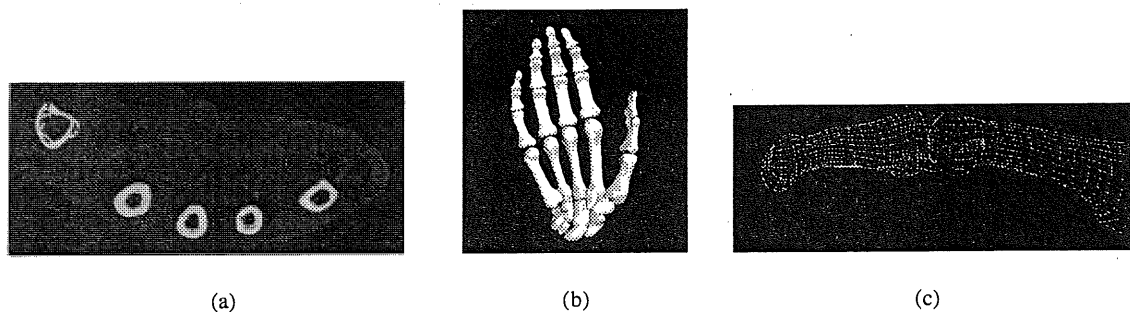


Figure 1: (a) Single slice of CT data defining model geometry (b) meshes of hand bones from CT data (c) meshes of bones and ligamentous structures surrounding PIP joint of the index finger.

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ACKNOWLEDGMENTS

This research was supported in part by the United States Department of Energy under contract with the Lawrence Livermore National Laboratory.

DYNAMICAL MODEL OF THE HUMAN FINGER FOR ASSESSING MULTIJOINT TENDONS KINEMATICS WITH ARBITRARY INTERCONNECTION STRUCTURE

V.A. Sholukha¹, K.J. van Zwieten², P.L. Lippens², A.V. Zinkovsky¹

¹ St. Petersburg State Technical University, St. Petersburg, Russia

² Limburgs Universitair Centrum, Diepenbeek, Belgium

INTRODUCTION

Most studies on the biomechanics of the human finger are either based on considerations concerning the relation between anatomy and function (van Zwieten and Lauw, 1985) or by using a simplified formalising model based on qualitative analysis of the position of the elements in a kinematic scheme (Leijnse and Kalker, 1995). In this report a computer model of goal oriented movement synthesis (Zinkovsky et al., 1996) is presented, in which both approaches are incorporated. It takes into account the main anatomical elements of the finger and is based on calculations of multi-joint tendon movement with arbitrary interconnection structure. The model allows for estimations of pulling forces of the main tendons and for the distribution of the attachments of the elements of the spiral fibers.

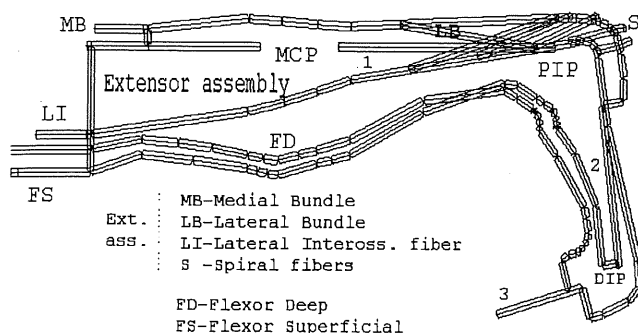


Fig. 1. Kinematic scheme of the finger.

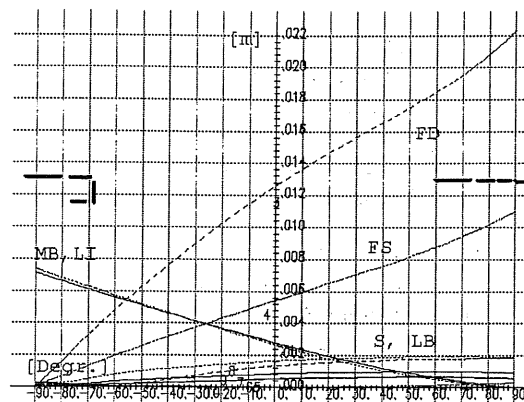


Fig. 2. Tendons elongations.

METHODS AND RESULTS

For the synthesis of the movement of the kinematic chain the mixed problem of dynamics is solved by using equations of constraint, realising the anatomical form of the composing elements of the joints, as a result of contact surfaces digitisation. Start/stop modes of the movement are synthesised. Mass-inertial characteristics of the phalanxes are derived from previously published data. The changes in positions of the phalanxes are recalculated in function of the movements of the main tendons and connecting fibers. Intermediate positions of the tendons are defined by the surfaces of the phalanxes and dependencies of the joint angles are assessed empirically. Thus, the main tendons are simulated as flexible unexpanded wires, which are tensible on a mobile contour, the form of which depending on the positions of the phalanxes. From variation of positions of points located on the main tendons the intermediate positions of tendons, ligaments and spiral fibers are determined. From smoothing of the results of the calculations of the changes in relative length of the tendon the velocity of muscle contraction is derived, which is an indispensable parameter for the realisation of the finger dynamics. Tendon positions were determined experimentally by measurements in the ring finger of an amputated hand. Dimensions and positions were measured in fully elongated position (0°) and 90° angles in M.C.P., P.I.P. and D.I.P. joints. In the 4-link human finger model with 10 degrees of freedom the consecutive and simultaneous flexing in the joints was simulated. Fig. 1 shows the kinematic scheme of the model. The change of the length of the tendons during simultaneous flexion in P.I.P. and D.I.P. joints (90° in 1 sec) as a function of the absolute angle of the rotation of the distal phalanx, is given in Fig. 2. Changes in length of the lateral bundle and spiral fibers remain within the limits of measurement errors. Maximum of velocity of the deep flexor tendon is about 40 mm/sec.

DISCUSSION

The model approximates a realistic representation of flexion in the human finger. By changing the parameters it may be adapted to any real situation. The adequacy of the model is dependent on the accuracy of the measurements of its anatomical constituents. Further refinery of the model will be achieved by introducing factors like forces of muscle contraction and of visco-elastic properties of ligaments elements.

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GASTROCNEMIUS ACTIVATION AND PLANTAR PRESSURE DISTRIBUTION DURING THE SPRINT START

M. J. Harland², A. Mero¹, S. Kuitunen¹, P. V. Komi¹

1. Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

2. Department of Biomedical Science, University of Wollongong, Wollongong, Australia

INTRODUCTION

Few studies have examined the influence of start block face angle on start performance parameters during the sprint start. Guissard et al. (1992) found that as the front block angle decreased the start performance increased. As the triceps surae group was progressively stretched in the less inclined block positions the authors reported a larger contribution of the medial gastrocnemius muscle due to earlier activity onset. This study was designed to integrate the activity pattern of the gastrocnemius (GA) muscle with the plantar pressure distribution during the sprint start.

METHODS

Ten skilled male sprinters (height; 1.78 ± 0.07 m, weight; 73.2 ± 6.4 kg and age; 25.3 ± 4.6 years) with a personal best 100 m time of 10.86 ± 0.34 s volunteered for the study. The subjects completed eight maximal starts over a distance of 20 m, from two randomised block angle positions (both front and rear blocks adjusted to either 40° or 65°). Right and left limb electromyographic and plantar pressure data were recorded (Paromed Medizintechnik GmbH, Germany) at 250 Hz.

RESULTS AND DISCUSSION

Figure 1 shows the development of GA activation and plantar pressures during the block phase of the sprint start (65° angle).

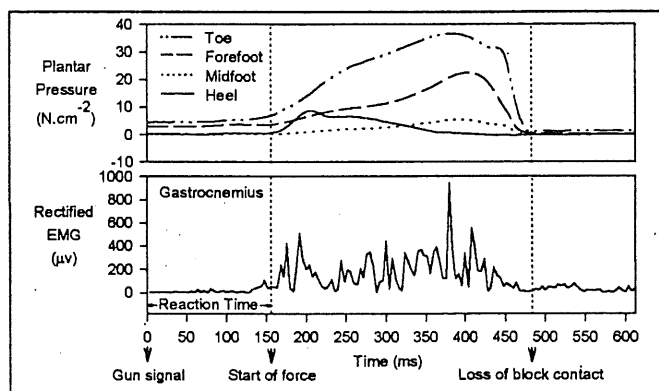


Figure 1: An example of front limb plantar pressures and GA activity (65° angle) during a block phase.

Higher plantar pressures in the block 'set' position on the rear foot (in both block conditions) were correlated to higher peak pressures during the propulsion phase for both heel (40° ; $r = 0.85$, 65° ; $r = 0.80$, $p < 0.05$) and midfoot (40° ; $r = 0.78$, 65° ; $r = 0.81$, $p < 0.05$) pressures. Also larger peak pressures exerted on the rear foot during the block start (in both block conditions) correlated to 20 m running times for both heel (40° ; $r = -0.72$, 65° ; $r = -0.77$, $p < 0.05$) and midfoot (40° ; $r = -0.72$, 65° ; $r = -0.82$, $p < 0.05$) pressures.

Electromyographic data showed that for the rear limb in the 40° condition, the GA muscle displayed a longer motor-time (38 ± 24 ms) compared to the 65° condition (2 ± 24 ms; $p = 0.003$). The 40° block angle start also displayed greater IEMG activity during both the block phase (29% greater activity; $p = 0.0001$), and during the 48 ms pre-block phase (128% greater activity; $p = 0.024$) for the rear limb compared to the 65° condition.

This earlier and greater activation of the GA in the 40° block angle, and the observed greater block velocity ($p = 0.007$) at the same angle (compared to 65°) suggest that the lower block angle is beneficial in maximising start performance on the blocks.

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CHANGES IN MUSCLE ACTIVITY PATTERNS WITH INCREASING RUNNING SPEED

H. Kyröläinen and P.V. Komi

Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

INTRODUCTION

Knowledge of EMG patterns of the leg muscles is important for understanding biomechanics in running. The extensor muscles usually increase their EMG activity with increasing running speed (e.g. Mero and Komi, 1987) although results are not always uniform (Ito et al. 1985). The purpose of the present study was to examine muscle activity patterns of the leg muscles and their relationships with ground reaction forces at different speeds.

METHODS

A total of 17 young endurance runners participated in this study. They ran at 12-13 different running speeds ranging from $3.25 \text{ m}\cdot\text{s}^{-1}$ to individual maximum ($8.31 \pm 0.75 \text{ m}\cdot\text{s}^{-1}$). 3-D ground reaction forces were measured simultaneously with telemetric EMG recordings. Surface electrodes were placed between the center of the innervation zone and the distal tendon of the vastus lateralis (VL), biceps femoris (BF), gastrocnemius (GA) and tibialis anterior (TA) muscles as well as on the upper part of the gluteus maximus (GM) muscle. EMGs were full-wave rectified and integrated in four phases: preactivity (0-50 and 50-100 ms before the contact), braking and push-off phases. Muscle activity patterns were created by averaging signals intra- and interindividually.

RESULTS AND DISCUSSION

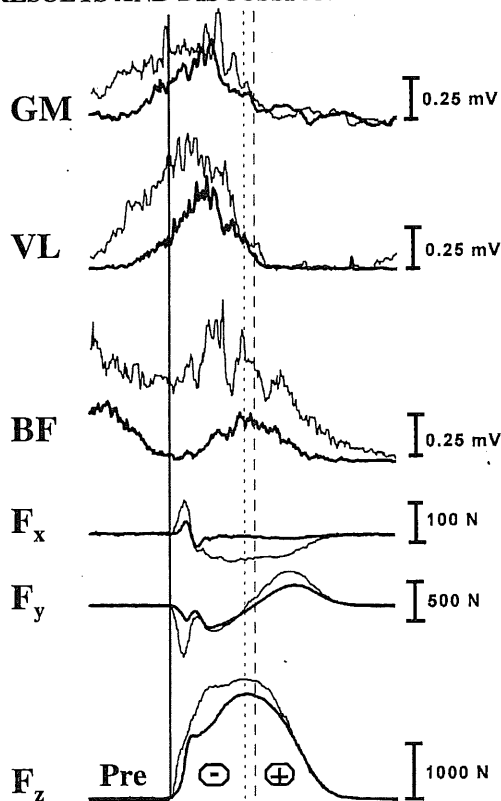


Figure 1. The mean, time-normalized EMG and ground reaction force curves of 85 contacts (5 contacts/17 subjects) when running at $4.00 \text{ m}\cdot\text{s}^{-1}$ (thick line) and maximal speed (thin line).

The mean contact time shortened gradually with increasing running speed from 0.227s (braking time 0.110s, push-off time 0.117s) to 0.115s (0.054s, 0.061s). At the same time, stride rate increased from $2.79 \pm 0.08 \text{ Hz}$ to $4.09 \pm 0.19 \text{ Hz}$ ($p < 0.001$), and the ground reaction forces increased in all directions. The highest increase was observed in F_y values: $116 \pm 20 \text{ N}$ vs. $242 \pm 45 \text{ N}$ ($p < 0.001$) in the braking phase and $123 \pm 20 \text{ N}$ vs. $271 \pm 46 \text{ N}$ ($p < 0.001$) in the push-off phase. At the same time, the directions of the averaged resultant forces decreased with respect to the horizontal level in both phases.

These changes in reaction forces were characterized by clear changes in EMG activity patterns. Figure 1 shows that the EMG amplitudes increased with increasing running speed in all muscles. For the GM, VL and GA muscles, this increase occurred primarily in the preactivity and braking phases. The TA EMG increased also just before the touch-down ($156 \pm 85 \mu\text{V}$ vs. $327 \pm 148 \mu\text{V}$, $p < 0.05$). However, the highest change was noticed in the BF muscle. Its EMG amplitude increased clearly both in the preactivity and contact phases (Fig. 1.).

The results emphasize the role of the hamstring muscles as well as the importance of the high pre- and eccentric activity at higher velocities. This could suggest optimizing the stiffness characteristics with increasing impact loads.

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BIOMECHANICAL ANALYSIS OF HURDLE RUNNING : IN COMPARISON WITH SPRINTING

A. ITO and M. TOGASHI

OSAKA UNIVERSITY OF HEALTH AND SPORT SCIENCES, KUMATORI, OSAKA, JAPAN

INTRODUCTION

Hurdle running can be divided into two phases. One is interval running and the other is hurdling. Although it had been believed that the hurdling was currently different from sprinting, hurdling is becoming more similar to sprinting with increasing of the world record. In the present study, hurdle running by different hurdlers with different hurdle running velocities was analyzed kinematically and kinetically to determine whether better hurdlers' hurdling and interval running were similar to sprinting.

METHODS

Subjects were nine male hurdlers (18 to 22yrs) whose best record for 110mH ranged from 14.33 to 15.88 seconds. Five hurdles (height: 1.067m, interval distance: 9.14m) were set, and the subjects were asked to repeat the maximal hurdle running while their movements were recorded by a 16 mm movie camera and ground reaction forces were recorded by a force platform from the 3rd hurdle to the 4th hurdle (the 1st step after landing, the successive 2nd step, take-off and landing). The mechanical work was estimated from the joint power computed from the film and force analysis. For comparison, the same analysis was done for sprinting.

RESULTS and DISCUSSION

The horizontal velocity of the center of gravity decelerated at the first half of the foot contact and accelerated at the latter half of the foot contact in all steps during hurdle running. Acceleration at each step was larger than deceleration except at take-off. Deceleration at take-off demonstrated a significant negative correlation ($r=-0.801$, $p<0.01$) with the hurdle running velocity. Faster hurdlers demonstrated a higher maximal velocity of high knee movement (take-off leg) before foot contact for the take-off ($r=0.745$, $p<0.05$). The maximal swing back velocities of the landing leg (represented by a line from the hip to the ankle) were higher in faster hurdlers ($r=0.848$, $p<0.01$).

The faster hurdlers showed greater mechanical work in the following muscles and movement phases. That is, the hip flexors and the knee flexors of the take-off leg during high knee movement before foot contact to take off ($r=0.717$, $p<0.05$; $r=0.715$, $p<0.05$, respectively), and the hip extensors of the leading leg when landing ($r=0.708$, $p<0.05$) showed positive correlation with hurdle running velocity. However, the faster hurdlers showed less mechanical work in the following muscles and movement phases. The knee extensors during the support period at take-off ($r=-0.798$, $p<0.01$), and ankle extensors of the landing leg ($r=-0.763$, $p<0.05$) showed negative correlation with hurdle running velocity.

According to these kinematic and kinetic results, the faster hurdlers' values were nearer to those of sprinting. Therefore, the present study suggested that faster hurdlers demonstrated in leg movements and muscle activities similar to those of sprinting.

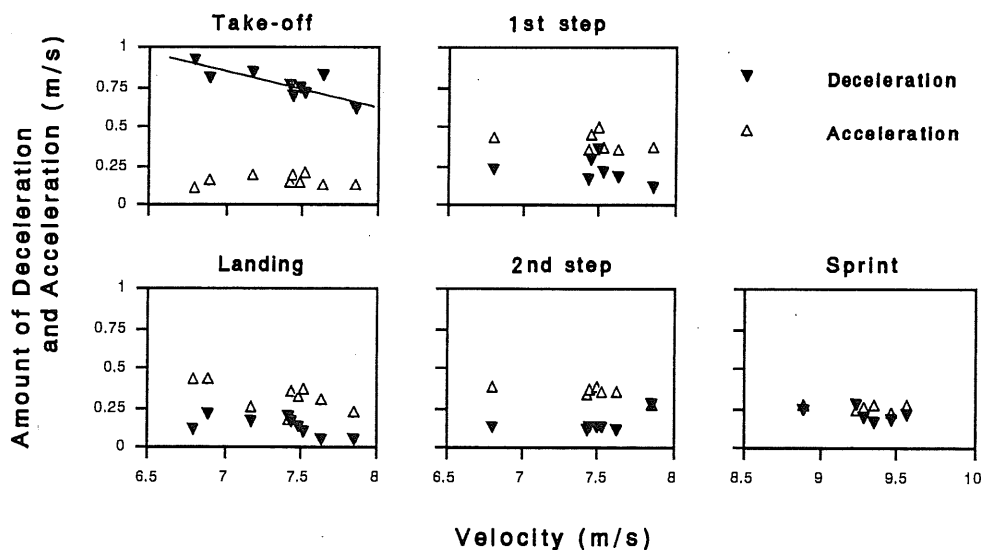


Figure 1. The relation between the amount of deceleration and acceleration during foot contact, and the velocity of hurdle running and sprinting.

INTERRELATIONSHIPS BETWEEN GROUND REACTION FORCES AND VELOCITY DURING ACCELERATION PHASE OF THE SPRINT START

A. Mero¹, M. Harland², S. Kuitunen¹ and P.V. Komi¹

1. Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

2. Department of Biomedical Science, University of Wollongong, Wollongong, Australia

INTRODUCTION

The running velocity during sprint acceleration depends on starting block positions, acceleration technique and force production in the blocks and after the blocks. The force-time characteristics of the sprint start have been studied in the block phase (e.g. Henry 1952) and during the first contact after leaving the blocks (Mero 1988). The aims of the present study were to investigate ground reaction forces in the blocks and during the six contacts after leaving the blocks at two different block obliquities and to relate them to running velocity during acceleration.

METHODS

Ten male sprinters (height 1.78 ± 0.07 m, weight 73.2 ± 6.4 kg and 100m record time 10.86 ± 0.34 s) performed in a hall eight starts on a long force platform at block angulations of 40° and 65° in a randomised order. Force-time data were recorded on a computer at 833 Hz from the force platform consisting of sixteen force plates (each 1.25 m long and 0.60 m wide; Raute Precision, Finland) arranged in two adjacent rows to a length of 10 m. The force plate area was covered with a paper roll to record the first six stride lengths. The running times were recorded with photocells placed at 5 m, 10 m, 15 m and 20 m.

RESULTS AND DISCUSSION

The force-time curves of one subject are presented in Figure 1. In general, the F_z - curves during the sprint start demonstrated a gradual change from a double shape curve to that of a "bouncing ball" which was often reached already by the end of the 10 m start.

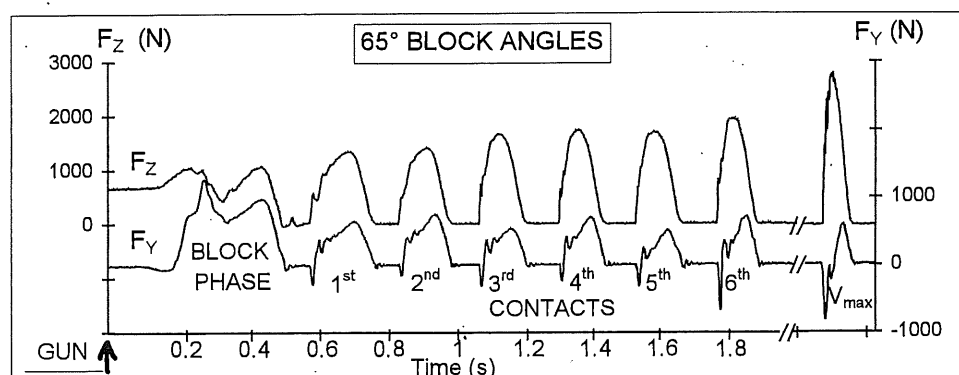


Figure 1. An example of horizontal and vertical force-time curves in the blocks, during acceleration and at maximal velocity (V_{max}).

As compared to the 65° condition the 40° block angle was characterized by a longer duration of force production on the rear block ($p=0.012$), greater horizontal force impulse on the rear block ($p=0.022$), greater total block impulse ($p=0.005$) and greater block leaving velocity ($p=0.007$). Correspondingly the first stride length was shorter ($p=0.000$) and the first stride rate was higher ($p=0.019$) in the 40° block angle. The stride rate during the first six strides correlated with the 20 m running time both in the 40° conditions ($r=-0.72$, $p<0.02$) and in the 65° conditions ($r=-0.86$, $p<0.01$). Also in both conditions the block velocity correlated significantly with the 20 m running time ($r=-0.76$, $p<0.01$ for 40° and $r=-0.79$, $p<0.01$ for 65°).

The great block velocity and stride rate are advantageous in the sprint acceleration but there were no differences in the 20 m time between the 40° and 65° block obliquities although the block velocity was greater in the 40° conditions.

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EVALUATION OF TRAINING BASED ON FEEDBACK ON PERFORMANCE IN THE 110 METERS HURDLES

C. Papadopoulos¹, S. Prassas², K. Valasotiris³, A. Giavroglou³ and L. Tsarouchas³

¹Aristoteles University of Thessaloniki, TEFAA Serron, 62110 Serres, GREECE; ²Colorado State University, Fort Collins, CO 80523, USA; ³Hellenic Sports Institute, 37 Kiffisias Ave., 37-15123, Marrousi, GREECE

INTRODUCTION AND PURPOSE

Improvement in sports' performance can be achieved by improving physical conditioning and/or technique. To achieve this, practitioners utilize a variety of methods and training regimes. The purpose of this study was to evaluate the effect on performance of a training program based on feedback in the 110 meters hurdle.

METHODS

Sixteen hurdlers of the same caliber, randomly divided into experimental (n=8) and control (n=8) groups, participated in the study. Pre-and-post test data were collected utilizing a CYBEX II Isokinetic Dynamometer, a KISTLER force plate, and a LOCAM high speed camera set at 100fps. The experimental group participated in a 10 week training program, in which, in each session, feedback related to dynamic and kinematic characteristics of their performance was provided. The control group was informed of the pretest results, but was not given specific feedback during training. Pre-and-post variables examined were: 1) maximum knee joint extension and plantar flexor isometric and dynamic torque, 2) maximum heights in jumps with (CMJ) and without (SJ) countermovement, 3) angles of the line connecting the center of mass (CM) with the planted foot measured CCW from the positive horizontal axis, before (B) and after (A) the hurdle at landing (L) and at take off (TO) (THETABL, THETABTO, THETAAL, THETAATO), 4) the angle of the velocity vector of the CM with the positive horizontal axis before the hurdle at take off (THETA VCM), 5) vertical velocity of the CM before the hurdle at take off (YVELCMBTO), 6) horizontal velocity of the CM after the hurdle at take off (XVELCMATO), 7) foot ground contact time after the hurdle (CONTIME), 8) flight height and flight time--over the hurdle (FLHEIGHT and FLTIME, respectively), and 9) 45 meter hurdle race time (TTIME). Paired t-tests between pre-and-post performances were conducted.

RESULTS AND DISCUSSION

Results revealed that training with or without feedback did not significantly change knee and ankle joint torque measurements nor maximum heights in CMS and SJ jumps in both groups. These results indicated that the initial level of physical conditioning was high in both groups. Thus, further improvement in performance could be achieved only by improving

TABLE 1
Descriptive Statistics (M, SD) and t-scores for the experimental group

Variable	Pretest		Posttest		t-score	p
THETABL (deg)	117.3	4.00	117.0	3.90	1.99	n/s
THETABTO (deg)	67.0	2.20	65.0	2.20	3.03	<.05
YVELCMBTO (m/sec)	2.11	0.22	1.95	0.23	2.32	n/s
THETA VCM (deg)	16.0	2.00	14.0	1.20	3.66	<.01
FLHEIGHT (m)	0.23	0.05	0.19	0.02	2.85	<.05
FLTIME (sec)	0.41	0.03	0.37	0.03	4.36	<.01
THETAAL (deg)	95.0	2.60	93.0	2.70	2.29	n/s
THETAATO (deg)	79.1	2.60	78.6	2.70	2.28	n/s
XVELCMATO	7.72	0.32	7.97	0.26	2.37	<.05
CONTIME (sec)	0.11	0.01	0.10	0.01	7.47	<.005
TTIME (sec)	6.82	0.23	6.62	0.22	4.80	<.05

technique. T-test scores for the control group showed no performance improvement (TTIME did not decrease significantly) as well as no statistically significant changes in technique (variables 3 through 9 in methods section) after 10 weeks of training without specific feedback. In contrast, as Table 1 indicates, performance (TTIME) significantly improved in the experimental group. This improvement was achieved by positive modifications in technique, such as decrease in CONTIME, THETA VCM, and THETABTO. These technique modifications, chosen by the athletes on the bases of their biological make up and the requirements of the specific task (Papadopoulos, 1990), decreased FLTIME, FLHEIGHT, and increased XVELCMATO. It was concluded that specific feedback during training positively affected performance in the 110 meters hurdle.

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SWING PHASE KINETICS OF PREFERRED AND NONPREFERRED SPEEDS OF WALKING AND RUNNING

Mary M. Ryan and Robert J. Gregor

Department of Health and Performance Sciences, Georgia Institute of Technology, Atlanta, USA

INTRODUCTION Comparisons between stance phase kinetics for preferred and nonpreferred walking and running has been reported (Ryan and Gregor, 1994), however no comparisons of swing phase kinetics have been made between preferred and nonpreferred locomotor conditions. The purpose of this study was to examine the generalized muscle moments of the lower extremity during the swing phase of preferred and nonpreferred walking and running to highlight possible mechanical reasons leading to a preferred gait at a particular speed.

METHODS A linked segment rigid body model of the lower extremity was used for a 2-dimensional analysis of the swing phase of gait. Data were collected on three subjects, who locomoted overground at walking speeds of 1.5, 2.0 and 3.2 ms^{-1} and at running speeds of 2.0, 2 and 4.1 ms^{-1} . Walking at 3.2 ms^{-1} and running at 2 ms^{-1} were classified as nonpreferred speeds. Lower extremity movement was captured at 200 Hz using two high speed video cameras (NAC, Inc.). Generalized muscle moments (GMMs) of the foot, shank and thigh were calculated using standard inverse dynamics equations (Winter, 1990).

RESULTS The ankle GMM profiles for all walk conditions (open symbols) showed an increase in magnitude throughout swing as speed increased (Figure 1). A similar trend was also observed for the run conditions (closed symbols), however only minimal increases in magnitude were displayed at the end of stance across run conditions. In the preferred walk conditions, the knee and hip moments displayed similar magnitudes for the first 70% of stance, however, the intermediate walk (2 ms^{-1}) showed an increase in magnitude for the remainder of swing (Figure 2 and 3). During the first 30% and between 75-90% of swing, the magnitude of all GMMs for the fast nonpreferred walk were up to three times greater than for the preferred walks and all run conditions.

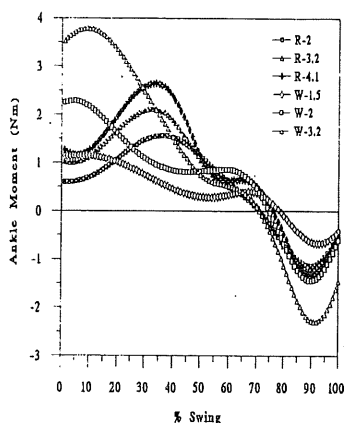


Figure 1

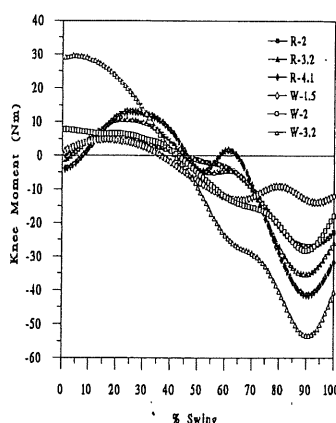


Figure 2

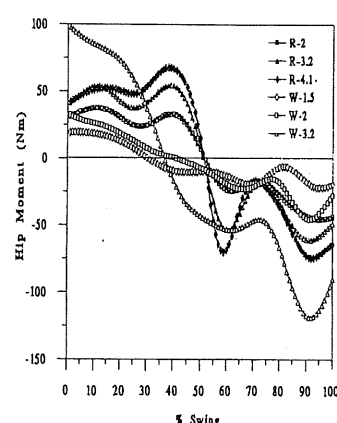


Figure 3

DISCUSSION The GMM patterns for all running conditions were consistent with increases in speed, with no apparent differences between preferred and nonpreferred speeds. While the GMM profiles for the walk conditions followed the same general pattern across speeds, the magnitude of the nonpreferred walk was substantially greater than expected and exceeded the values for even the two fastest run conditions. The large increases in knee and hip GMM magnitude at the beginning of swing were consistent with the increased knee extensor and hip flexor moments present at the end of stance as reported by Ryan and Gregor (1994). The GMMs for the walk and run at similar speeds were greater in the nonpreferred walk than the preferred run (3.2 ms^{-1}) but did not display such large deviations between the nonpreferred run and preferred walk (2 ms^{-1}). Larger GMM values reflect the greater muscular demands needed to produce moments necessary for maintaining the fast walk speed. An increased muscular demand is consistent with EMG records presented by Nilsson et al. (1985) which showed increased EMG activity for both walking and running as speed increased. Therefore, the lower GMM magnitudes present in the run at 3.2 ms^{-1} compared to the walk at this speed, would suggest that running is preferred to walking due to the decreased demand on the musculature to locomote at this speed. These results may be important for understanding the natural transition between walking to running gaits and why certain gaits are chosen for a particular speed.

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DISTANCE RUNNING BIOMECHANICS IN RELATION TO INJURY - A CASE STUDY APPROACH

K.R. Williams, Department of Exercise Science, University of California, Davis

Davis, CA, USA

INTRODUCTION

Investigations into the relationships between biomechanical aspects of distance running and injury using groups of runners have had some limited success, though the relationships identified are often not strong and of limited generality. One inherent problem in group studies is the high anatomical and biomechanical variability present in a normal population of runners that may obscure more specific relationships with injuries. An alternative research paradigm that can be useful for analyzing running injuries is the case study approach, though it is seldom used except in clinical investigations because the analysis may be viewed as too specific to individuals and a typical statistical analysis is not feasible. It is possible, however, that such an approach can identify relationships among the injury history, functional and structural characteristics, and biomechanical measures for the individual that can also enhance our understanding of the mechanisms of injuries, and may provide insights that will help us better focus group studies. Presented here are two such examples of case studies.

METHODS

Subjects described here are examples from a larger group of more than 50 elite male and 40 elite female distance runners evaluated in an interdisciplinary program at the United States Olympic Training Center in Colorado Springs, CO between 1985 and 1990. Data were collected at running speeds of $5.36 \text{ m} \cdot \text{s}^{-1}$ for women. Kinematic data for one side of the body were collected from a side view at 100 (cine) or 200 (video) fps and for a rearfoot view at 200(cine) or 400(video) fps, depending on which year data were obtained. Coordinate data were filtered using a Butterworth low-pass filter and used to calculate various linear and angular kinematic measurements. Ground reaction forces were collected at 500Hz during running outdoors on a synthetic track surface over a Kistler force platform, with data from five trials for left and right feet averaged while subjects ran in training, racing, or both types of shoes. Anthropometric data and a lower extremity evaluation were obtained using typical methods, a complete injury history for each runner was obtained, and a clinical analysis was performed by a podiatrist. Results are presented here for two elite female middle distance runners, though other similar examples can be found among the subjects tested.

RESULTS

Case #1: Elite Female Middle Distance Runner

This runner had recent severe Achilles tendon problems on the right side and a history of bilateral plantar fascii irritation, both severe enough to limit training and prevent competing. Anthropometric data showed a 1cm longer left leg and flat feet, but with very little eversion possible from a neutral subtalar position. Stride length relative to leg length (417%LL) was long compared to the mean for other female elite runners (395%LL), accompanied by a relatively high vertical oscillation. She had a relatively large decrease in A-P velocity during the braking phase of support, and the most distinct aspect of her kinematics was a very large knee flexion angle during support (57.1°) vs. a mean of 44.2° . The rearfoot angle was very supinated at footstrike (18° L, 21° R), but both feet showed a small maximal pronation angle (-4.3° L, -4.2° R). Maximal vertical forces (2.55BW-R, 2.61BW-L) were very low compared to mean values (3.14BW), with the vertical force showing a typical rearfoot pattern despite a midfoot strike index (43%-L, 55%-R). Analysis of the data suggested that the Achilles tendon problems were related to the large knee flexion (and slight higher than average maximal dorsiflexion) occurring in support, with greater flexion putting greater stress on the tendon. The low maximal vertical force was likely a result of the high knee flexion during support, and these patterns may have been an adjustment to the limited pronation angle achieved during the support phase. The low pronation angle was probably the result of the restricted subtalar eversion, and this, combined with the large amount of knee flexion in support, may have caused the plantar fascii irritation.

Case Study #2: Elite Female Middle Distance Runner

This 1500m runner had had three 3rd metatarsal stress fractures in the right foot in the previous four years of running. The left leg was 0.6 cm longer than the right, arches were somewhat flat, and the left foot abducted more than the right foot. Stride length relative to leg length (380%LL) was slightly less than mean values, and while maximal pronation was within normal values, the right foot pronated more (-12.3°) than the left (-7.4°). Maximal vertical forces were extremely high (3.9BW-R, 3.76BW-L), and strike index indicated an extreme forefoot striker (65%-R, 83%L). The extreme forefoot strike and high vertical forces, combined with higher pronation on the right side, probably predisposed this runner to stress-related injuries in the right foot. This may have been exacerbated by doing fast interval training in lightweight spikes, a situation where impact stresses might have been even higher.

DISCUSSION

These examples show how kinematics and ground reaction forces can be combined with anatomical measurements to show relationships that seem to be directly related to injuries in runners. In the search to establish more definitive mechanisms for injuries in running, it may be that examining individuals for relationships between biomechanical factors and injury can be useful. Once relationships are established for individuals, it should be possible to gradually develop a larger database of specific interactions with particular injuries. Studying individuals may be an equal or more suitable alternative to epidemiological studies or more general investigations using groups of runners.

THE BIOMECHANICAL ANALYSIS OF FATIGUE IN TIBIA

Z. Fang

Beijing University of Physical Education, Beijing, China

INTRODUCTION

During the phase of running and jumping, the tibia bears large load where lays the common part of fatigue fracture. Considering the case reports, the location apparently relates to the types of motions. The fatigue fracture mostly occurs on the up-middle location of tibia in motions of jumping, such as high jumping and basketball. However in the motions of speed running, such as 100 meters' race, long jump and hurdle, the location is likely on the down one third. Combined with the 130 case reports, the fatigue fracture locations of frequent occurrence and the causes of fracture were analyzed so as to provide a theoretical basis for the diagnosis, treatment and prevention of the fracture.

METHODS

1. Using the trigonometric method of vector resolution, we raise the computational theory of the contracted force on tibia. Figure 1 shows the force diagram about leg. J is the femur-tibia joint force on tibia. We can identify the magnitude and direction by the trigonometric method of vector resolution after having attained the joint force on ankle R . Figure 2 shows the force diagram about foot. We can identify the magnitude and direction of force R by using the following function:

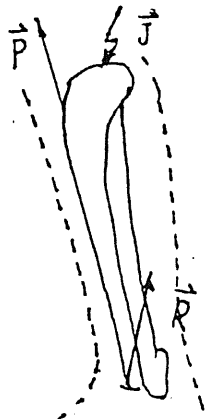


Figure 1.

$$\begin{cases} J_A \ddot{\theta} = \sum M_A \\ m\ddot{x} = \sum F_x \\ m\ddot{y} = \sum F_y \end{cases}$$

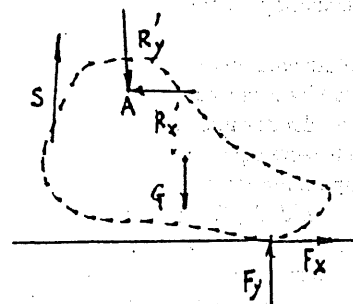


Figure 2.

2. By searching materials, we obtain 130 references about fatigue fracture in tibia.

3. Through analyzing the forces and the stress distributions on tibia, we use the method of biomechanics to get the mostly location that the maximum stress occurs.

RESULTS AND DISCUSSION

During the take-off of men's high jump, the horizontal component in ankle $R'_x \ll$ the vertical component R'_y . The axis pressure on tibia is about 900 kg on the up-middle of tibia. During the speed running, a torque is on the distal of tibia. Tibia bears the torsion load. There is a shear-stress τ on the cross section.

$$\tau = \frac{T_1 \cdot \rho}{I_p}$$

It is about twice as many the shear-stress on the distal of tibia as the one on the proximal of tibia.

JOINT COORDINATE SYSTEM (JCS) APPLICATION TO DIFFERENT JOINT MECHANISMS

R L Van Vorhis

Interdisciplinary Ergonomics Research Program

Lawrence Livermore National Laboratory

Livermore, California

A M Hollister

Department of Orthopaedic Surgery

Louisiana State University Medical Center

Shreveport, Louisiana

K Hollerbach

Institute for Scientific Computing Research

Lawrence Livermore National Laboratory

Livermore, California

INTRODUCTION

The Joint Coordinate System (JCS) of Grood and Suntay [1] has been used to represent human joint motion and uses non-orthogonal coordinate axes. The first coordinate (e_1) is fixed relative to the proximal bone and usually represents flexion-extension (FE) movements. A second coordinate (e_3) is fixed relative to the distal bone and is used to represent internal-external (IE) motions. A third coordinate (e_2), a function of the other two, moves and is used to represent abduction-adduction (AA) movements. This system has been applied to human joints without regard to the actual joint mechanism(s). Since the nature of the joint mechanism and the number of degrees of freedom are important in calculating limb statics and dynamics, as well as control theory, the effect of variations in the joint mechanisms on the ability of the joint coordinate system to represent movements of the mechanism is crucial. Several joint mechanisms have been shown to be represented by from one to three revolute [2,3]. The authors have generated the forward kinematics for several known joint mechanisms and have calculated the inverse kinematics for each mechanism using the JCS. A revolute coincident with the initial reference coordinate, an arbitrary revolute, two revolute and three revolute mechanisms were investigated.

METHODS

Location of a fixed and a moving axis system were established in simulations implemented in LabView 4.0 on a Macintosh computer. The kinematics of arbitrary rotation axes [k_i] were implemented and specified by a_i and b_i orientation angles relative to the reference coordinate directions as well as the position of a point lying on each axis. Two and three revolute systems were created by linking the revolute with their a and b angles and the vector separating the revolute.

RESULTS

A revolute parallel to an axis coordinate direction gave an exact match of Grood and Suntay JCS variables with the actual joint mechanism. An arbitrary revolute, one that is not parallel with any reference frame coordinate, produced large discrepancies between the actual and calculated axes' rotations however, the displacements were equal.

The simulation for two arbitrary axes was configured as typical of a knee joint. The first coordinate (e_1) is aligned with the first revolute simulating the FE axis, and the second coordinate

(e_3) is aligned with the second revolute and offset 10mm simulating the IE axis. The revolute's rotations and the G&S JCS parameters correspond exactly. If the revolute were not aligned with the JCS coordinates, discrepancies occurred between the actual and the measured rotations.

For the three axis system, a wrist example was used with, one revolute representing Flexion/Extension motion, one revolute representing Radial/Ulnar deviation and one revolute representing Pronation/Supination motion. These revolute are mutually non orthogonal and non intersecting. A simulation was performed in which two of the three revolute were aligned. Orientation and displacement results of the three revolute did not match the G&S JCS variables.

DISCUSSION

The JCS can be used to represent a one revolute joint such as the elbow if one of the JCS axes is aligned with the joint's revolute. If this is not done, significant discrepancies occur between the motion of the mechanism and the recorded rotations. Erroneous motion can register in all three JCS parameters even though the actual motion source is one degree-of-freedom. The magnitude of the errors are non-linearly related to misalignment and increase with increasing misalignment.

For a two revolute joint system, the discrepancies are avoided if the first coordinate is aligned coincident with the first offset revolute, and the second coordinate axis is aligned with the second offset revolute. Increases in alignment discrepancy increases the error and decreases correlation of the JCS coordinates with motion for the mechanism.

For three independent revolute mechanism joints, the JCS cannot represent the joint mechanism and hence cannot represent the motions of this joint.

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COMPUTATIONAL TECHNIQUES FOR ESTABLISHING FOOT SIZING SYSTEMS FOR BIOMECHANICAL APPLICATIONS

Gautham Gopalakrishna, Nachiappan Chockalingam, Surendra M and Das B N
Shoe Design and Development Centre, Central Leather Research Institute, Adyar, Chennai 600 020

INTRODUCTION:

The anthropometric data of any population shows certain dissemination even within a given age group. The foot sizes also exhibit a definite range, which, in turn has to be reflected by the size range of the footwear offered for the local population. To be able to produce properly designed footwear a well elaborated sizing system is required. Lasts, the three dimensional representation of the human foot manufactured in other countries of the world cannot serve the purpose in any other country. The reason for this could be attributed to the anthropometric differences of the population and to the wearing conditions, urbanisation, climate, shoe consumption etc., Therefore, to obtain a reliable data on foot proportions it is imperative to conduct a nation wide survey to collect statistical data for analysis. Results of this study would help more accurate biomechanical foot modelling of a specific population. However, the reported study concentrated on India and its diversified population. The results would in turn help the development of new orthoses and prostheses.

METHODS:

The human foot is a complex three dimensional object and it would be very difficult to collect the necessary numerical data for describing it by the traditional and manual measuring methods. Therefore, a new method for data collection and sampling was evolved. The method developed uses the video digitizing technique which essentially is a system comprising of two video cameras that measures the foot from orthogonal positions viz., the bottom and side profile. The live pictures of the foot were grabbed and digitised. The digitised data are then stored as picture files. The data collection and analysis procedure included the transformation of pixel-graphic picture file to a vectorised data stream describing the shape contours. The curves and contours are analysed mathematically in terms of sizes, distances and angles. The entire country was zoned and a sampling size representative of the entire population was computed. These zones were based on anthropological, biological, cultural and linguistic variables. About 40000 people across the country were measured.

RESULTS AND DISCUSSIONS:

The entire database containing all measurements throughout India was integrated into one master database. This was then indexed on age and sex to arrive at the various categorisations. Various statistical tests like the multiple regression analysis, t-tests and Chi-square tests were carried to delineate various size groups. It was found from the study of correlation coefficients that below the age of 12 years the foot size is determined only by age and not by gender.

From 12 years to 18 years the foot growth for Girls and Boys is markedly different. Whilst, the subjects above 19 years had a foot growth rate, that is insignificant. Altogether, 24 sub-groups were identified and sub-sets were created in each of these groups for 72 variables. These variables represented various foot parameters. A spline fit for the curves was carried out and statistical mean of these parameters were determined that helped us to evaluate the increments of these parameters from size to size. Thus, we were able to attach footwear sizing numbers and generate cross tables of footwear sizing and measurements using the statistical data. We were then able to generate the geometric data for last modelling of these groups.

This geometric data evolved would be a vital input for any further biomechanical modelling of foot. The studies that are currently being carried out on the prosthetic foot would use these data for aesthetic modelling and functional comfort. Further, the mathematical analysis, using FEM for catering to foot/footwear comfort could also make the best use of the reported results.

LOCAL PROPORTIONAL SCALING OF TIME SERIES DATA OF HUMAN MOVEMENT

K. Kanatani-Fujimoto¹, B.V. Lazareva², V.M. Zatsiorsky¹

¹Biomechanics Lab., Department of Kinesiology, The Pennsylvania State University, University Park, PA, U.S.A.

²Biostatistics Unit, Cornell University, Ithaca, NY, U.S.A.

INTRODUCTION

A method of analysis of time series data, local proportional scaling (*LPS*), is suggested and its applications in biomechanics and motor control are discussed. The *LPS* is a further development of the two procedures widely used in human movement science to analyze time series data; scaling and segmentation. The method is based on comparison of two time curves, a reference curve $x(t)$ and a 'transformed' curve $x'(t')$. By assumption, $x'(t')$ was received from $x(t)$ by local affine transformations; local extensions/compressions along the amplitude (x) and time (t) axes [$x(t) \rightarrow x'(t')$].

The aim of the *LPS* method is to discover the underlying transformations, including: 1) the time epochs (*TE*) on which the local transformations have been performed; 2) the gain indexes [$GI(k)$], the coefficients of the local extension/compression along the time and amplitude scales; 3) the velocity quotients (*VQ*), characterizing the ratio of the velocity of the two curves; V_1/V_2 , where V_1 and V_2 are the signal velocity of the reference and 'transformed' curves, respectively; the velocity quotients are independent from the local affine transformations; 4) the time segments (*TS*), the periods of time during which the velocity quotients are constant; and 5) the time quotients (*TQ*), characterizing the overall time transformation of the $x'(t')$ and numerically equal to the inverse of the velocity quotient.

PROCEDURES

LPS is a two-stage procedure. First stage is a local affine transformation of $x'(t')$ preserving the signal velocity (time derivative of the signal) but not the signal continuity. The *TE*s are determined and *GI*s are calculated at this stage. The process is performed on the monotonic curves—compare an ascending portion of the reference curve with the ascending portion of the 'transformed' curve, and the descending portion with the descending portion. In the second stage, a time-vs.-time (*TxT*) comparison of the two curves is performed. *VQ*s and *TS*s are calculated next. The whole procedure is equivalent to the inverse affine transformation [$x'(t') \rightarrow x(t)$].

Vertical ground reaction force data of running with two different velocities are utilized to illustrate the procedures. **Step 1**) *TE*s are defined for both reference and 'transformed' curves. This can be done by identifying local maxima and minima (Figure 1). **Step 2**) *GI* is calculated for each *TE*. *GI* (k) is defined as the ratio of the range of data values (R) between $x(t)$ and $x'(t')$ for each corresponding *TE*; $GI = R[x(t)]/R[x'(t')]$. To preserve the signal velocity of the $x'(t')$ for each *TE*, a local extension/compression of $x'(t')$ is performed on both x and t axes using the same *GI*, k (Figure 2: $k[x'(t')]$). **Step 3**) Sequentially measure the time of $k[x'(t')]$ corresponding to $x(t)$ by dynamically interpolating $k[x'(t')]$, where $x(t_n) = k[x'(t'_n)]$ (Figure 3). **Step 4**) Repeat the Steps 2 and 3 for each *TE* and draw a linked plot for the entire data of $x(t)$ and $x'(t')$ (Figure 4). **Step 5**) Plot the *TxT* plots for $x(t)$ vs. $x'(t')$ and $x(t)$ vs. $k[x'(t')]$ (Figure 5). The *TxT* plot for $x(t)$ vs. $x'(t')$ can be derived from $x(t)$ vs. $k[x'(t')]$ plot by dividing the signal velocity of $x(t)$ vs. $k[x'(t')]$ values of each *TE* by corresponding *GI*. The slope of the $x(t)$ vs. $k[x'(t')]$, $\{x(t_{n+1}) - x(t_n)\} / \{k[x'(t'_{n+1})] - k[x'(t'_n)]\}$, represents the velocity quotient for the data points n and $n+1$ in the transformed time scale (*TVQ*). To obtain the velocity quotient in the actual time scale (*AVQ*), it is necessary to divide the *TVQ* by the appropriate *GI*. **Step 6**) Find the inflection points and define the time segments from the *TxT* plot for $x(t)$ and $k[x'(t')]$. Because $x'(t')$ was previously adjusted for time and amplitude, time segments (*TS*) defined from the *TVQ*s represent a 'pure' time segment. They do not represent, however, the relative timing of the actual signals. To estimate the relative timing, *TS*s can be identified from the *AVQ* using the *TS* defined for *TVQ* (Figure 6). **Step 7**) For each *TS* defined, actual and transformed regional velocity quotients can be calculated and used to explain underlying phenomena. In this example, three *TS*s are defined suggesting the existence of three underlying movement changes due to altered velocity.

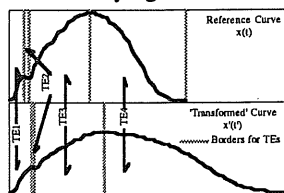


Figure 1. $x(t)$ and $x'(t')$ with *TE*s identified (gray lines represent end of each *TE*).

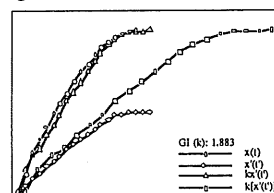


Figure 2. Local scaling of $x'(t')$ for *TE*3 ($kx'(t')$: amplitude scaled using k ; $k[x'(t')]$: amplitude and time scaled using the same k).

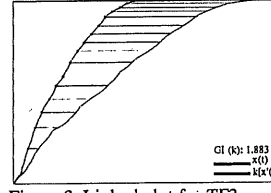


Figure 3. Linked plot for *TE*3 (corresponding time where $x(t_n) = k[x'(t'_n)]$ is measured).

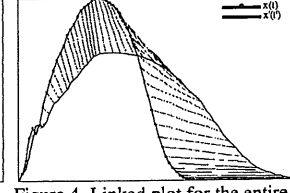


Figure 4. Linked plot for the entire data range in the original x and t scales.

DISCUSSION

As is any mathematical method the *LPS* should be applied to appropriate objects. For the *LPS*, the objects are the time series curves mapped by the local or global affine transformations. If the curves do not satisfy this requirements the method does not hold. The *LPS* should be used when either local or global affine transformations in the time and space scale, are suspected. Though the method is not immediately bound with any theory of biomechanics or motor control, it can be used for testing hypotheses on invariant relative timing, local and global scaling, generalized motor programs, etc. In general, the *LPS* method can be used for: (a) comparing the time series curves in a concise transparent manner; (b) scaling the curves, bringing $x'(t')$ in conformity with $x(t)$; (c) automatic segmentation of the time series data; and (d) data classification.

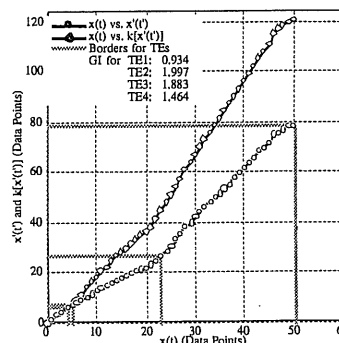


Figure 5. *TxT* plot for $x(t)$ vs. $x'(t')$ and $x(t)$ vs. $k[x'(t')]$.

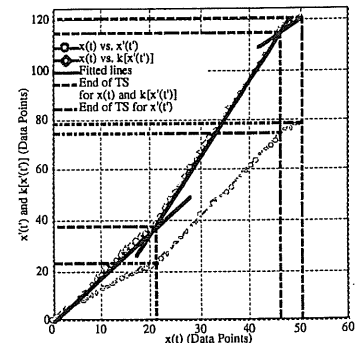


Figure 6. Identified time segments from $x(t)$ vs. $k[x'(t')]$ and $x(t)$ vs. $x'(t')$.

UNIQUENESS AND COMMONALITY ANALYSES OF TIME SERIES DATA OF HUMAN MOVEMENT

K. Kanatani-Fujimoto¹, G.A. Smith², V.M. Zatsiorsky¹

¹Biomechanics Lab., Department of Kinesiology, The Pennsylvania State University, University Park, PA, U.S.A.

²Biomechanics Lab., Department of Exercise and Sport Science, Oregon State University, Corvallis, OR, U.S.A.

INTRODUCTION

A method of continuous analysis of time series data to identify 'substantial uniqueness' (*SU*) and 'substantial commonality' (*SC*) in a human movement is presented and its applications in biomechanics and motor control research are discussed. The basic idea of the method is to sequentially perform statistical testings along the duration of the data among time series data.

PROCEDURES

The procedure consists of three major steps: 1) normalization, 2) calculation of continuous statistics (*C-Stat*), and 3) evaluation of the statistics. Linear displacement of golf club head in medio-lateral (*XD*), antero-posterior (*YD*), and vertical (*ZD*) directions of 13 elite professional golfers performing driver (*1W*) and 8-iron (*8I*) shots (from set up through impact) is utilized to illustrate the procedures. The data are normalized twice for the duration and amplitude of each data series (Figures 1a through 1c).

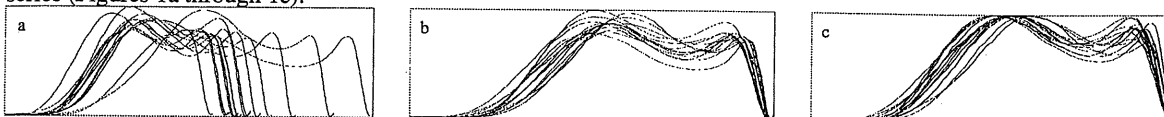


Figure 1: *ZD* for 13 subjects. a) absolute, b) time normalized, and c) time and amplitude normalized.

CONTINUOUS UNIQUENESS ANALYSIS

Series of 'continuous' *t*-tests (*C t-test*) are calculated for each data point across the duration of the swing. The calculated *t*-values are plotted against the duration of the swing (Figure 2; shaded areas indicate significant difference bands at $\alpha=.10$). By examining the *C-Stat* plot, it is possible to identify the data range where subjects performed 'unique' movement (shaded areas) which significantly differs from other attempts. To visualize the result of the *SU* analysis, ensemble mean plot with 90% confidence intervals (*CI*) is generated and combined with continuous plot results of the zero-null hypothesis testing (*C-H₀ plot*); the horizontal bar graph at the top (Figure 3; shaded areas indicate *SU*).

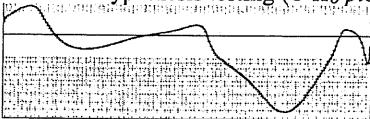


Figure 2: *C-Stat* plot for the *t*-test between *1W* vs. *8I* for *ZD*.

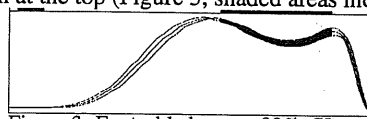


Figure 3: Ensemble mean, 90% *CI*, and *C-H₀* plots for *1W* vs. *8I* for *ZD*.

The use of a filtering routine to process biomechanical data assumes continuity and smoothness of a human movement. Hence, the filtering process virtually eliminates the possibilities of outliers of data. Although the borders between *SU* and non-*SU* may be fuzzy due to the lack of statistical power, this method would effectively identify data ranges where *SU* are present.

CONTINUOUS COMMONALITY ANALYSIS

In order to evaluate the existence of *SC* in *XD*, *YD*, and *ZD*, all the three curves are simultaneously examined. Series of *C t-test* for each data point across the duration of the swing for all three dimensions are simultaneously performed. The *C-Stat* plot for each dimension is generated (Figure 4a through 4c; shaded areas indicate statistically non-significant difference at $\alpha=.10$).

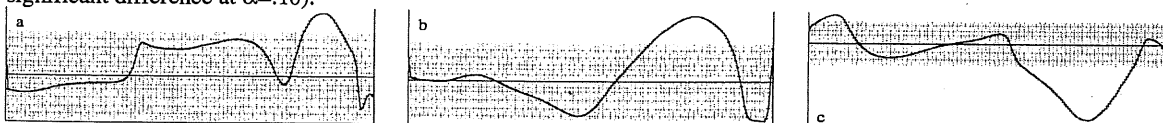


Figure 4: *C-Stat* plot for the results of *t*-test between *1W* vs. *8I* for: a) *XD*; b) *YD*; and c) *ZD*

Because the commonality analysis is designed to identify the existence of similarities, the method focuses on the data ranges where statistically non-significant differences are present. However, non-significant *t*-test results may appear due to: 1) large variance and 2) similar movement. In order to identify the *SC*, 10% of the data range of each data series is utilized as an evaluation criterion. In other words, the variable is identified to possess *SC* if the results of the *C t-test* showed non-significant results across dimensions while the variance at the specific data point across dimensions is within 10% of the data range. In order to illustrate the result of the *SC* analysis, the ensemble mean plot with 90% *CI* for each dimension is generated and combined with *C-H₀* plots across dimensions (Figures 5a through 5c; shaded areas indicate *SC*). The use of 10% variance criteria as well as the assumption made for the filtering routine make the method effective to identify the existence of *SC*. However, the borders between *SC* and non-*SC* ranges may be fuzzy due to the lack of statistical power.

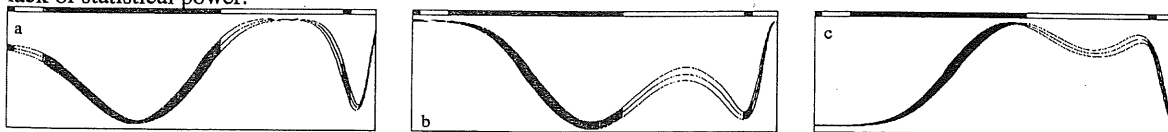


Figure 5: Ensemble mean plot and 90% *CI* and *C-H₀* plot between *1W* vs. *8I* for: a) *XD*; b) *YD*; and c) *ZD*.

DISCUSSION

The method facilitates continuous evaluation of human movement in which temporal and spatial characteristics of movement are simultaneously evaluated. In particular, it enables the evaluation of the existence of the unique and common movement patterns. Although due to the large number of statistical tests performed, the method does not provide estimates of the statistical significance, it does provide useful results that can be applied to many areas of human movement studies including biomechanics and motor control. This method may be particularly useful where a practical significance is more important than a statistical significance. The applications of the method can be extended to segment and joint kinematics, angular kinematics, and kinetic data. In addition, the use of other statistical methods such as ANOVA and MONOVA can be equally effective.

EFFECTS OF SPONTANEOUS RESPIRATION ON PULMONARY ARTERY INPUT IMPEDANCE

P. Castiglioni, M. Morpurgo and M. Di Rienzo
LaRC, Centro di Bioingegneria, Milan, Italy

INTRODUCTION

The pulmonary artery input impedance, PAII, is a quantitative coupling index between the right ventricle and the pulmonary vascular bed in the frequency domain, being defined as ratio between oscillations in blood pressure and blood flow at the entrance of the pulmonary circulation. Because of the low blood pressure values characterizing the pulmonary circulation, the changes of intrathoracic pressure associated with breathing should influence the PAII. Surprisingly, the available literature does not report any significant difference between PAII values of heart beats occurring during inspiration and expiration both in spontaneously breathing men (1) and animals (2). In all the previous studies, however, PAII has not been assessed during the transition between the inspiratory and the expiratory phases of respiration, though this transition is characterized by important biological changes, like a marked increase of intrathoracic pressure and concomitant changes in right heart dynamics (3). The aim of the present study is to focus on this issue by assessing the changes of the PAII during the transition from inspiration to expiration in spontaneously breathing dogs.

METHODS

Pulmonary artery blood pressure, PAP, and pulmonary artery blood flow velocity, FV, were measured at the entrance of the main pulmonary artery in 5 anesthetized, spontaneously breathing dogs via a high fidelity Mikro-Tip catheter with pressure and velocity sensors at the same location. Airflow was simultaneously measured. The average duration of the recordings was 81 s. For each respiratory cycle PAII was separately assessed over the heart beats falling completely in inspiration (reference beats), and the heart beats occurring during the transition from inspiration to expiration (transitional beats). The position of each transitional beat within the transition between inspiration and expiration was quantified in terms of expiratory fraction, E_{FR} , defined as percent ratio between the fraction of the beat falling in expiration and the total beat duration.

RESULTS

We observed in all animals that for transitional beats 1) PAII at 0 Hz, i.e., input resistance, linearly increased with E_{FR} ; 2) with respect to the reference condition, the modulus of PAII at the frequency of the heart rate progressively decreased up to -50% for E_{FR} from 0 to 50%, and then symmetrically increased for E_{FR} from 50 to 100%; and 3) the phase of PAII at the frequency of the heart rate was greater than in reference beats for E_{FR} below 40% and lower than in reference for E_{FR} above 50%. Characteristic impedance did not show any systematic modulation during the transition. Changes of input resistance and impedance modulus and phase at the frequency of the heart rate mirrored similar changes in the mean value, and in the first harmonic modulus and phase of the PAP wave. Negligible changes were observed during the respiratory transition in FV wave and in the heart rate.

DISCUSSION

The beat-by-beat analysis of PAII and the classification of transitional beats according to E_{FR} indicate that there is a respiratory influence on PAII during the transition from inspiration to expiration. Results suggest that the observed impedance changes are mainly due to modifications in the shape of the PAP wave, presumably caused by the raise in intrapleural pressure at the start of expiration.

Since variations of PAII imply modifications in the work done by the right ventricle, the PAII changes of transitional beats may be one of the determinants of previously reported changes in right heart dynamics during the transition from inspiration to expiration (3).

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CHARACTERISTICS OF REGIONAL HEMODYNAMICS IN FREE-MOVING RATS.

M. Di Rienzo, A. Daffonchio, P. Castiglioni
LaRC, Centro di Bioingegneria, Milan, Italy

INTRODUCTION

Circulatory regulation provides each district of the vascular system with the fraction of cardiac output required by the local needs. The strategies of such blood flow control are still mainly unexplored. In this study we addressed this issue by analyzing 1) the spectral characteristics of muscular, splanchnic and renal blood flow velocities in free-moving rats and 2) the correlation between flow velocities and blood pressure.

METHODS

Arterial blood pressure was measured by an intra-vascular catheter inserted into the femoral artery in 9 conscious rats. Three blood flow velocities were simultaneously measured by pulsed Doppler flowmeters placed in the abdominal aorta (which supplies the hindquarter muscular district), in the superior mesenteric artery (splanchnic district) and in the left renal artery (renal district). These districts were selected because they account for most of the total flow and because they are paradigmatic of different biological needs (muscular activity, digestion, vital function). Each recording lasted about 3 hours. During the monitoring period the rats were spontaneously behaving inside a large Plexiglas box.

The recordings were sampled at 156 Hz and edited from artifacts. Mean arterial pressure (MAP), and mean aortic (AV), renal (RV) and mesenteric (MV) flow velocities were derived on a beat-to-beat basis. For each beat-to-beat series of MAP, AV, RV and MV the power spectrum was computed as follows. The series was split into 120 sec segments with a 50% overlapping, and tapered with the Hanning window. The spectrum was estimated in each segment by FFT and integrated over 3 frequency bands: LF (0.025-0.1 Hz), MF (0.1-0.6 Hz) and HF (0.8-3 Hz). These bands received particular attention in literature because their power may reflect the activity of specific control mechanisms of the cardiovascular system (1). In each animal the LF, MF and HF powers were averaged for the whole recording and expressed as percentage of the total power (LF+MF+HF). The statistical significance of the difference between LF and MF powers and between LF and HF powers was assessed by the paired t-test. The degree of linear correlation between MAP and each of the 3 flow velocities was quantified by the squared coherence, k^2 . Coherences between MAP and AV, MAP and MV, MAP and RV were estimated between 0.02 and 3 Hz.

RESULTS

The normalized powers in LF, MF and HF are reported in the table I. The relative distribution of the powers among the 3 bands differs according to the signal. LF power prevails in MAP and AV; by contrast, MF and HF powers are predominant in MV and RV.

Table I: Normalized power (%). Asterisks indicate significant power differences ($p < 0.05$) with respect to the LF power.

	LF	MF	HF
MAP	43.61	37.47	18.92*
AV	40.84	25.74*	33.42
MV	24.55	37.68*	37.77*
RV	18.85	43.01*	38.14*

The figure shows the 90% confidence intervals of the mean coherence between MAP and each of the 3 flow velocities. The coherence between MAP and AV is very low over the whole frequency range considered. By contrast, fluctuations of MAP show a certain degree of linear correlation with MV and RV in the MF band, as indicated by the coherence peaks at 0.14 Hz for MV and at 0.43 Hz for RV.

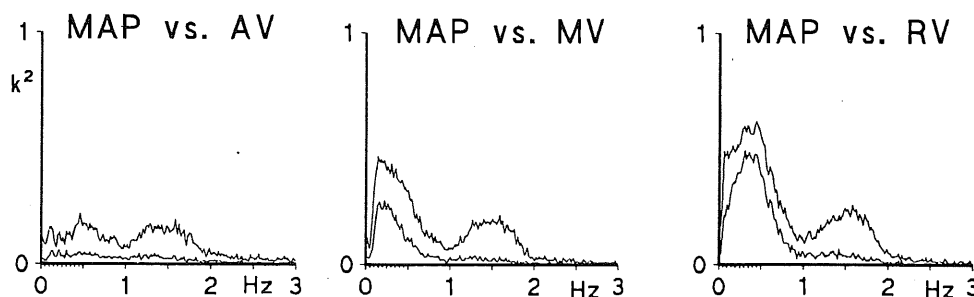


Fig. 1. 90% confidence intervals of the mean coherence between MAP and the 3 flow velocities.

DISCUSSION

Our results clearly indicate that regional hemodynamic control is exerted through different strategies. In particular, in the renal and splanchnic districts, which are characterized by constant or very slow varying flow demands, mainly MF and HF fluctuations are present, and these fluctuations are somewhat correlated with corresponding fluctuations of MAP. Conversely, in the aortic district, which supplies muscles, blood flow demand is variable and strictly related to muscular activity. This is reflected by a greater contribution of the LF components to the local overall variability.

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DEFORMATION-INDUCED HIERARCHICAL FLOWS AND STREAMING POTENTIALS IN BONE : PREDICTIONS FROM A COMPUTATIONAL MODEL

J.D. Zhang and A.F.T. Mak

Rehabilitation Engineering Centre The Hong Kong Polytechnic University, Kowloon, Hong Kong

INTRODUCTION

Bone is a highly structured composite material with a hierarchical network of flow channels from the level of vascular flows through the Haversian canals to the flows in the tiny lacunae-canalliculi channels immediately around the bone cells and the interstitial flows through the microporosity of the collagen-hydroxyapatite bone matrix. Most earlier studies of interstitial flows in bone have only examined the contributions from different flow systems separately, such as the flows through the microporosity, the canalliculi, and the Haversian canals. An overall model encompassing the hierarchical microstructure is important to our understanding of the actual physics of flows in bone. The flow-induced drag forces and streaming electrical potentials could interact with the osteocytes to effect biological responses.

THEORETICAL MODEL

The biphasic poroelastic theory was used to describe the interstitial fluid flow through the deformable collagen-hydroxyapatite microporosity. The lacunae-canalliculi flows system was modeled as a "superposing" discrete network of one-dimensional channels emanating from a Haversian canal. Cross flow at a given point between the canalliculi and the microporosity was assumed to be proportional to pressure difference at that point between the two compartments. The Haversian canals were modeled by hydrostatic elements. A finite element mesh was generated to represent an idealized model for the bone microstructure. Sixteen channels emanating from each Haversian Canal were introduced to simulate the canalliculi. A pure bending situation was simulated. The Young's Modulus was 15 GPa and the Poisson's Ratio was 0.25. The hydraulic permeability of the matrix microporosity was taken to be $0.13 \times 10^{-15} \text{ m}^4/(\text{N}\cdot\text{s})$. The ramp time was taken to be 2 sec. The maximum strain rate was 0.1 % per sec.

RESULTS AND DISCUSSION

Solidity in the channels representing the amount of filling within the canalliculi was varied parametrically. It was found that fluid velocity dropped almost linearly with the solidity in the canalliculi filling became higher. The diffusive drag force within the canalliculi increases dramatically as the canallicular solidity approached 1. It was noted that the velocity in the canalliculi predicted by the current model is much higher than that in the microporosity, suggesting that the presence of the canalliculi may significantly affect the fluid flow pattern. The normalized streaming potential distribution along a line cutting through 3 osteons was estimated from the pressure profile. (Figure 1) The compressive and tensile sides during pure bending were to the left and right of the model respectively. It was noted that the cusp-like potentials experimentally measured by Pollack et al (1981) could be reproduced qualitatively here.

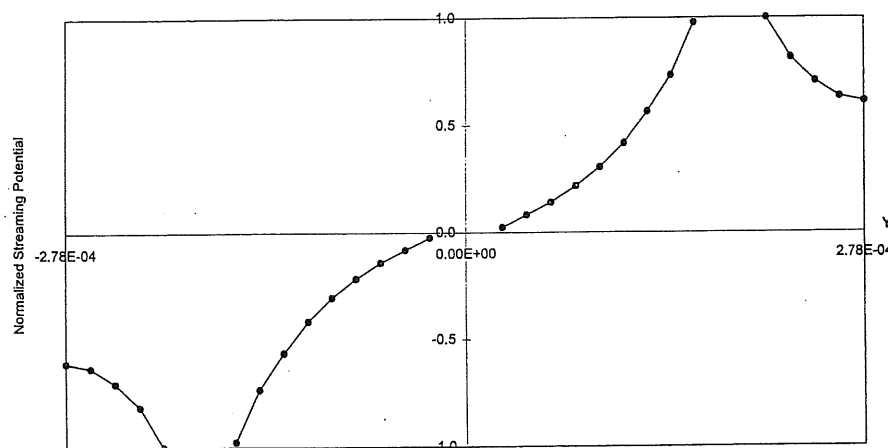


Figure 1 - Normalized Streaming Potentials Across 3 Osteons. The osteon on the left was under compression, the right was under tension. The one in the middle was nearest to the neutral plane.

THE BUILDING-UP OF SUBCUTANEOUS TISSUE COMPACTION AND DISTORTION UNDER COMBINED PRESSURE AND SHEAR EPIDERMAL LOADINGS

J.D. Zhang and A.F.T. Mak

Rehabilitation Engineering Centre, The Hong Kong Polytechnic University, Kowloon, Hong Kong

PURPOSE

This study was to examine quantitatively the role of epidermal shear traction in the development of tissue compaction and distortion within a soft tissue layer covering a bony prominence. This work was motivated by a lack of a theoretical framework to understand the biomechanical processes associated with the development of pressure sore.

METHODOLOGY

A 2-dimensional nonlinear finite element model of a soft hydrated poroelastic tissue covering a rigid and impervious bony prominence was established for this study. The epidermal surface was subjected to an uniform compressive displacement with or without a concomitant shear displacement. The material constants were assumed to be 156 kPa for the equilibrium modulus, 0.16 for the Poisson's Ratio and $3.5 \times 10^{-14} \text{ m}^4/\text{N.s}$ for permeability. The compressive and shear displacements were applied in a ramp-plateau manner (500 sec ramp time). At the end of the ramp phase, an overall 20% compression and 15% shear were generated in the soft tissue right above the bony prominence. Internal tissue compaction, maximum shear stress, and the interstitial fluid pressure as functions of time were calculated at mid layer right above the bony prominence using this finite element model.

RESULTS AND DISCUSSION

Both the maximum shear stress and the interstitial fluid pressure exhibited overshoots at the end of the ramp phase. The overshoots were 2.5 to 3 times higher than for the case with an additional 15% shear displacement on the epidermal surface. However, both the shear and the interstitial pressure transients relaxed after the ramp phase with the latter back to zero towards equilibrium. Tissue compaction continued to rise after the ramp phase with the latter back to zero towards equilibrium. At 1000 sec, tissue compaction was 3.5 times higher for the case with additional epidermal shear than for the case without. The difference became smaller in percentage as time increased. These results documented the significant effects of epidermal shear on all the internal biomechanical variables within the tissue layer and demonstrated biomechanically the adverse effects of shear on the clinical development of non-superficial pressure sore.

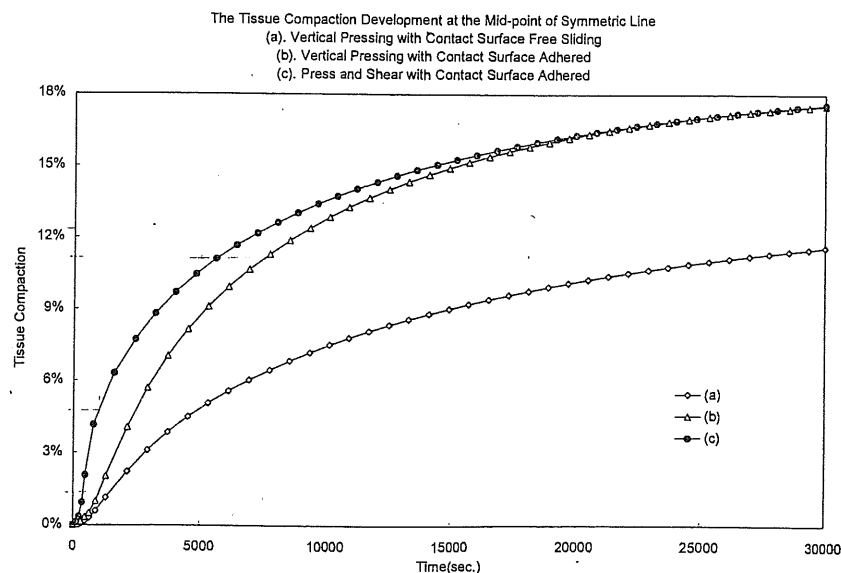


Figure 1. The build-up of tissue compaction at the mid tissue layer directly above the bony prominence.

MODELING MECHANICAL RESPONSES OF BIPHASIC CARTILAGE UNDER DYNAMIC LOADING

J. Z. WU, W. HERZOG and M. EPSTEIN

Human Performance Laboratory, Faculty of Physical Education

Department of Mechanical Engineering, Faculty of Engineering

The University of Calgary, Calgary, Alberta, Canada

INTRODUCTION

Experimental findings show that dynamic loading can regulate the biosynthetic activities of chondrocytes, and so affects the extracellular matrix in cartilage layers. In the initial stages of osteoarthritis, cartilage layers become thicker, softer, and more permeable than layers of the corresponding healthy cartilage. The ever-changing material properties of degenerating cartilage influence the stress and strain field in the cartilage layers. The present study was aimed at simulating numerically the fields of stress and fluid pressure in normal and "diseased" cartilage layers for static and dynamic loading.

METHODS

Based on the linear, biphasic constitutive law proposed by Mow et al. (1980), and Ateshian et al.'s (1994) asymptotic solution, an analytical solution for the contact pressure, and the stresses in the cartilage layers was derived for axi-symmetric joint under dynamic loading. Assuming the time period t is discretized into equal parts Δt , i.e., $t_i = i\Delta t$ ($i = 1, 2, 3, \dots$), so that the contact pressure is $P_A(r, t_i) = P_{Ai}(r)$. The contact force, F_i , can be considered approximately constant over each time step. The contact pressure and the stresses in the cartilage layers depend not only on the current contact force, but also on the loading history of the contact forces.

RESULTS AND DISCUSSION

The diseased cartilage was modelled by changing the thickness, permeability, shear modulus, and Poisson's ratio of the cartilage. The results (Fig. 1) show that, for a given loading condition, the contact areas increase and peak stresses decrease in the diseased compared to the normal joint. This finding implies that areas of normal joint contact become unloaded, and areas of little or no contact pressure become overloaded compared to the situation in normal cartilage. The proposed solution can be used to simulate the effects of changing joint geometry and cartilage properties as they may occur in joints following injury or disease.

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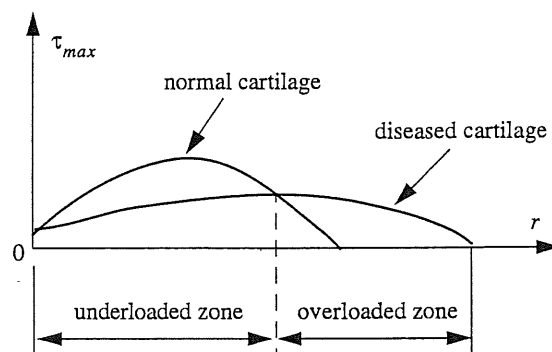


Figure 1: Comparison of the distribution of the maximal shear stress at the cartilage-bone interface between the normal and degenerated cartilage layers

GRASP SPECIFICATION USING A 3-D GLOVE IN VIRTUAL REALITY BASED POINT-AND-DIRECT TELEROBOTICS

Myung Hwan Yun*, Jae Hee Park**, David Cannon***, Andris Freivalds***

* : Department of Industrial Engineering, Pohang University of Science and Technology, Korea.

** : Ergonomics Laboratory, Korea Institute of Standardization and Sciences, Korea

*** : Department of Industrial and Manufacturing Engineering, Penn State University, USA

INTRODUCTION : Hand posture and force, which define aspects of the way an object is grasped, are features of robotic manipulation. A means for specifying these grasping "flavors" has been developed that uses an instrumented glove equipped with joint and force sensors. Point-and-Direct robotics was initially explored by the second author at Stanford University. This initial research demonstrated that an operator could direct a robot to perform tasks in a natural and interactive way by pointing to objects and destinations while giving directives such as "put that there". In demonstrations, this point-and-direct (PAD) Telerobot put trash in a wastebasket, and blocks on a pallet, and tools in a toolbox. In the Human Motion Laboratory at POSTECH and the CIM laboratory at Penn State, hand posture and force data were collected for manipulating bricks and other items that require varying amounts of force at multiple pressure points. The feasibility of measuring desired grasp characteristics was demonstrated for a modified Cyberglove impregnated with FSR (Force Sensitive Resistor) pressure sensors in the fingertips. A joint/force model relating the parameters of finger articulation and pressure to various lifting tasks was validated for the instrumented "wired" glove (Yun, 1994).

METHOD: An angle transducer glove system (Cyberglove™ CG1801,) and a set of data acquisition programs specific to finger joint measurement were developed. The Cyberglove system provided measurement of 18 joint angles. There were two flexion sensors on each of the five fingers. On the thumb, these two sensors measured the metacarpophalangeal (CMC) and interphalangeal (IP) joint flexions. On the remaining four fingers, the two flexion sensors measured the metacarpal phalangeal joint (MP) and proximal interphalangeal (PIP) joint flexions. Two wrist motions, flexion/extension and radial/ulna deviation angle, was also measured. For the measurement of finger forces, a sensor matrix with ten Force Sensitive Resistors was developed. This set up has been used to measure pressure distributions on foam grip handles, to study finger pinch forces, and hand-tool coupling effects. Each of the FSR sensors were covered with 2 mm thick plastic glue over a 12 mm² sensing area for effective force measurements. The sensors had an effective force sensing range of 1 to 50 N with 1 N precision. Sixteen different tool grip tasks were tested for six subjects. The joint angle data and the force data were collected for three seconds while the subjects were using the tools.

RESULT AND DISCUSSION : Overall, it was found that the grip force is a complex function of the size of the object, shape of the contact surface, and the types of the tools used. However, it was found that finger forces, index finger in particular, can be measured for the purpose of specifying the amount of force needed to maintain the grasp posture of the hand. Together with the joint angle data of the hand at that specific posture, a fairly good designation of position and force requirements could be possible for a robot. The force specification, is presumed to be initiated from the human operator based on his or her knowledge of the objects being grasped. The human operator holds an object similar to the one in the remote environment such as a brick. With the measurement system developed in this study, hand and finger posture as well as force to be applied are then relayed to the robot based on grasping performance using the instrumented glove. The force and position sensors attached to the Cyberglove™ system recognize the posture and the amount of force exerted to the finger tips and transfer the data as a specification to the manipulator for it to maintain during an otherwise autonomous grasping endeavor. With a basic robot algorithm to adjust for movement with gravity and to increase forces with the onset of slip, the instrumented glove specification system should allow the human to provide important data yet operate at a much higher level of strategic control than is possible with continuous telemanipulation. This simultaneous specification of force and joint articulation serves as a first step toward improving human supervision of grasping for remote telerobotics particularly in the case where humans point and give directions rather than provide continuous telemanipulative input. Robotics community may progress further toward such strategic control of grasping as more sophisticated robot mechanisms are developed. With such robotic capability adapted to the point-and-direct paradigm, the role of the human may be raised to the level of giving target grasping angles and forces while letting the robot adjust and maintain a reliable grasp during its own complex trajectories. Future research may consider the proper means to specify articulation angles and forces for robot grippers even when these grippers do not exactly resemble the human hand or the instrumented glove used to develop the grasp specifications.

ACKNOWLEDGMENT : This research was partially supported by the Research Grant of the Pohang University of Science and Technology, Pohang, Korea (No. P 96019).

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MODELING COMPACT BONE AS A COMPOSITE MATERIAL: PREDICTION OF THE ELASTIC PROPERTIES OF SINGLE LAMELLAE AND OF OSTEONS

E. Lucchinetti*, G. Kress°, E. Stüssi*

*Biomechanics Laboratory, ETHZ and °Institute for Construction Methods, ETHZ, Zurich, Switzerland

INTRODUCTION

There is no consensus in the related literature on the microstructure of compact bone in the *sub-micron* order of magnitude. One point of view is that of an ordered pattern of the collagen fibre orientation in bone lamellae, whilst another is that of an isotropic collagen fibrils' network. Moreover, the shape and the exact location of the apatite crystals are additional subjects of controversy. However, the bony material in a lamella displays some kind of anisotropy. The aim of this study was to calculate the material properties of the collagen-apatite composite material, and to use the calculated material data for prediction of osteons' elastic behavior. The results were compared to experimental data from literature.

METHODS

The Halpin-Tsai relationships [2] were used to calculate the elastic moduli *in the plane* of the lamella itself as a function of mineralization, in the case the composite is built of platelet-like apatite crystals embedded in a collagen matrix. In the case the crystals are needle-like the theory of Chamis for continuous fibers [1] was applied (this assumption is legitimated by the crystals' high aspect ratio). Perfect bonding between apatite and collagen was assumed. The material properties *out of plane*, that is in the direction normal to the lamella, were assessed using a generalized Reuss model. The osteons' models consist of stacks of bone lamellae (up to 16), each 4 μm thick, rolled into co-axial cylinders. Different collagen fibers orientations were taken into account. The osteons elastic properties were assessed using a software created for the prediction of the structural behavior of thick-walled composite tubes [3].

RESULTS

The material properties of the apatite reinforced collagen (platelet-shaped crystals) are plotted as a function of mineralization in Fig. 1. The calculated tensile and torsional properties of osteons are presented in table 1 and in table 2 (fiber direction 0° = longitudinal axis of the osteon).

Fig. 1:

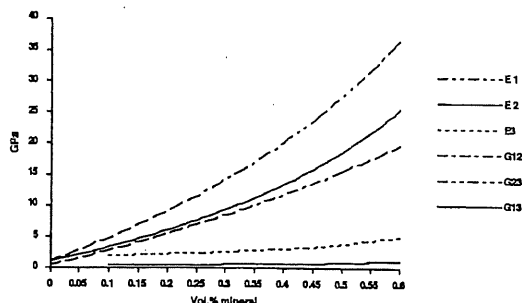


Table 1: Crystals assumed to be *needle-shaped*

fiber direction	tension	torsion
$\pm 5^\circ$	45.4 GPa	0.75 GPa
$\pm 45^\circ$	4.8 GPa	5.96 GPa
$\pm 85^\circ$	3.4 GPa	0.75 GPa
plywood	17.9 GPa	2.59 GPa

Table 2: Crystals assumed to be *plate-shaped*

fiber direction	tension	torsion
$\pm 5^\circ$	20.4 GPa	5.83 GPa
$\pm 45^\circ$	23.4 GPa	3.37 GPa
$\pm 85^\circ$	13.8 GPa	5.83 GPa
plywood	20.5 GPa	4.73 GPa

DISCUSSION

The results show very good agreement between the computed tensile properties of the osteons and literature values [4]. However the models provide an inadequate description of the elastic behavior of osteons in torsion. It is suggested that an improvement in the description of the bone composite material (e.g. allowing the platelet-shaped crystals to be rotated about some angle with respect to the lamellar plane) is needed, as well as further experimental work, to disclose the real nature of the collagen-apatite composite material.

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EXPERIMENTAL EVALUATION OF SKIN RUPTURE UNDER DIFFERENT LOADING CONDITIONS

W. Chang and A. A. Seireg

Department of Mechanical Engineering, University of Wisconsin-Madison, Madison, Wisconsin, USA

INTRODUCTION

A technique was developed to investigate the effect of simulated shapes and sizes of hard objects on the rupture of skin under pressure. Chicken skin was used for convenience. Air pressure was applied on the skin samples to simulate applied normal load. Six indenters with different shapes and sizes were used. Three sets of experiments were conducted. In the first experiments, the membrane strength of the skin was tested under evenly distributed pressure. In the second experiments, indenters were added to the same setup to simulate the effect of different shapes of bones or other hard objects on the skin when pressure was applied. Finally, tensile tests were also performed to construct stress-strain diagrams chicken skin.

METHODS

A skin sample with a diameter of 7.62 cm was held between two washers and tightened between a pipe and pipe cap. The top of the pipe cap had a one-way valve in order to apply pressure from the air source to the sample. The two washers had an inner diameter of 3.81 cm. Skin samples from chicken breasts and thighs were subjected to air pressure until they ruptured. The rupture pressure was recorded for each sample. In the second experiment, an indenter was attached to the other end of the pipe at the center with its tip below the skin sample. The length of the indenter was adjusted so it touched the skin sample without applying any pressure. The indenter could be easily detached from the second pipe cape. This allowed the accommodation of different shapes of indenters to simulate varied conditions of pressure under the skin. The first set of experiments were repeated with 6 different indenters (Figure 1). Three indenters were made from a dowel with a diameter of 1.91 cm. They had different shapes at the end which contacted the skin sample. The shapes of the indenters were flat, round, and sharp, respectively. The other 3 indenters were made of steel with diameters equal to 6.35, 3.81, and 2.29 cm, respectively. The thickness was that measured at the rupture location. Tensile tests were also conducted to study the stress-strain relationship of chicken skin.

RESULTS

It was observed that the location from which the skin samples were taken did not significantly affect the membrane rupture property of the skin. The pressure required to rupture the samples was proportional to the thickness. The average rupture pressure per unit skin thickness, P (Pa/mm), and the standard deviation, σ for the different tests are tabulated below. The results from the tensile tests show that for chicken skin the average rupture stress is 1.57 Pa, the average rupture strain is 19.91%, the average Young's modulus is 9.94 Pa, and the average residual strain is 3.80 %. A typical stress-strain diagram is shown in Figure 2.

Test	Without Indenter	Flat-end Indenter	Round-end Indenter	Sharp-end Indenter	Steel Indenter (D=6.35mm)	Steel Indenter (D=3.81mm)	Steel Indenter (D=2.29mm)
P	0.333	0.177	0.161	0.134	0.330	0.183	0.138
σ	0.085	0.044	0.027	0.037	0.125	0.041	0.074

Table 1 Average rupture pressure per unit skin thickness and standard deviation.

DISCUSSION

The results from tensile tests showed that chicken skin softened before rupture. By contrast human skin is known to stiffen as tensile stress increases (Zhivodyorov et al., 1989). The ultimate tensile stress and strain of the human skin were approximated at 4.7 MPa and 100%, respectively. The Young's modulus is approximately 1.48 MPa below a tensile stress of 0.7 MPa, and 55.12 MPa above that value (Kenedi et al., 1966). The differences of the mechanical properties of the skin from this two species may be attributed to the fact that humans have hair and chickens have feathers. An interesting result from the membrane stress tests is that the sharpness or roundness of the rigid objects did not significantly affect the required pressure for rupture.

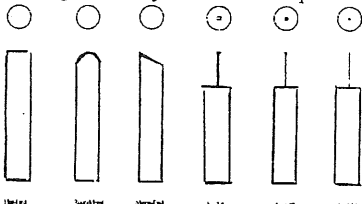


Figure 1 Indenters.

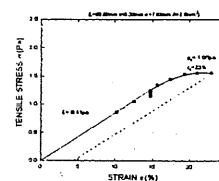


Figure 2 Typical stress-strain relationship of chicken skin.

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COMPRESSION OF SWINE BRAIN TISSUE; EXPERIMENT *IN VIVO*

Kiyoyuki Chinzei and Karol Miller

Biomechanics Division, Mechanical Engineering Laboratory, MITI, Tsukuba, Japan
Email: chin@mel.go.jp; kmiller@mel.go.jp

Introduction

Recent developments in Computer-Integrated and Robot-Aided Surgery [1], as well as advances in virtual reality techniques [2], call for closer examination of the mechanical properties of very soft tissues (such as brain, liver, kidney, etc.). The ultimate goal of our research into the biomechanics of these tissues is development of corresponding, realistic mathematical models. Such models would find applications in surgeon training and operation planning systems, based on the virtual reality techniques, registration [3], and control of surgical robots. The presented contribution, intended as a step towards the said goal, shows the experimental results of *in vivo* compression of swine brain tissue.

Materials and Methods

In vivo compression of swine brain tissue was performed in a testing stand, as shown in Fig. 1. The main testing apparatus was a two-linear-degree-of-freedom robot equipped with a sensitive load cell, build in our Laboratory. A head of a six-month-old, anesthetized swine was fixed rigidly in the robot base, Fig. 1. An opening in a skull of about 2.5 x 2.5 cm was prepared at the frontal lobe. The robot's cylindrical (10 mm diameter) end effector was placed so that it touched the exposed brain surface. The measurement began when the robot's end effector started vertical descent. The loading velocity was 1 mm/s. The robot movement was controlled and measured precisely by linear displacement sensors. The loading force was measured by a sensitive strain-gauge load cell.

The experiment consisted of two stages: loading and stress relaxation. This paper focuses on the results obtained during the loading phase. For compressive strains larger than 35%, measured *in vitro* [4], we observed unrecoverable damages in the tissue. Therefore, only one loading cycle was performed. The start of the loading phase was indicated by the first non-zero reading of the load cell. The end of loading phase and simultaneously the start of stress relaxation phase was indicated by the point of equalizing of the reading of the vertical displacement.

Results

Figure 2 shows loading force versus displacement curve. Surprisingly, even for high compression levels, the force-displacement relation is close to linear, with proportionality constant of 0.099 N/mm.

Discussion and Conclusions

This study presents the results of the *in vivo* compression of swine brain tissue. Close to linear force-displacement relation was observed. How to use the *in vitro* experimental results for defining models of brain mechanical properties remains an open question. Further research is needed to determine brain tissue constitutive models, which would incorporate the influence of the blood and cerebrospinal fluid pressure and flow.

Acknowledgments

The financial support of Japanese Science and Technology Agency and the New Energy and Industrial Technology Development Organization (NEDO) is gratefully acknowledged.

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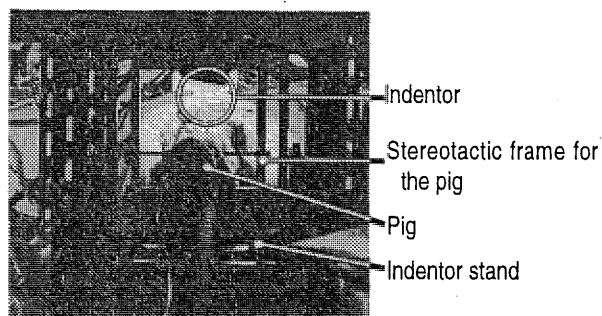


Fig. 1. Experimental stand

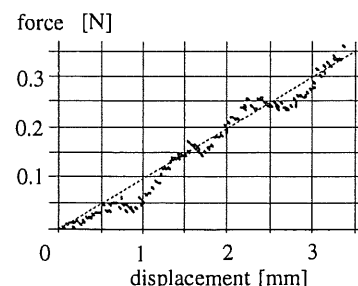


Fig. 2. Force - displacement curve obtained *in vivo*

STRESS RELAXATION MEASUREMENTS IN LINEAR REGION OF BRAIN TISSUE

Zizhen Liu, Nhan Phan-Thien and Lynne Bilston

Department of Mechanical and Mechatronic Engineering, University of Sydney, NSW 2006, Australia

INTRODUCTION

The mechanisms of brain injury are influenced both by the loading conditions and the mechanical behaviour of brain tissue. There are several previous studies on brain tissues: Galford and McElhaney (1970) studied the viscoelastic properties of human and monkey scalp, brain and dura. Shuck and Advani (1972) tested the human brain tissue in shear. Arbogast et al (1995) tested the stress relaxation of the brain of new-born pigs (age 2-3 days) at strain rates of 1s^{-1} to three levels of peak strain (2.5%-7.5%). Their experiments showed that brain is a viscoelastic material. We present an experimental measurement of the rheological properties of bovine brain tissue.

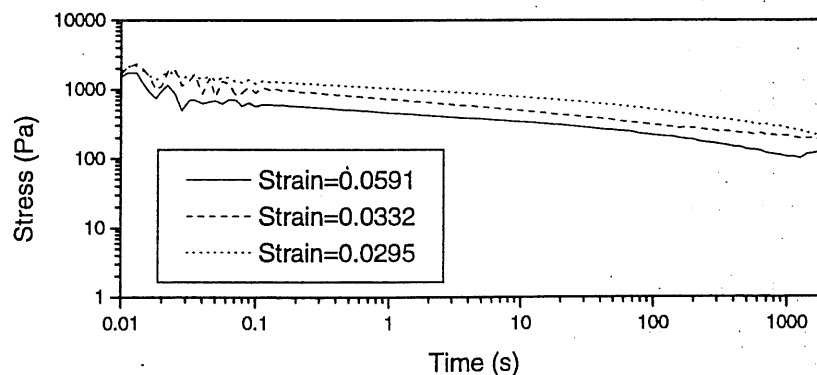
METHODS

The experimental protocol consisted of: 1) 10 cycles of preconditioning at 2Hz, 2) shear relaxation tests to measure the time dependent mechanical behaviour of the brain. Each brain was obtained from a nearby abattoir, freshly bathed in the artificial CSF immediately after removal from the skull. Then they were kept in 18°C mini container for eight hours prior to testing. The samples were transferred to an air-conditioned laboratory. All the tests were performed on a controlled strain rheometer, Bohlin VOR Rheometer, with parallel plate configuration, and a 93.286 gcm torque bar. The test temperature was set at 36.5°C and the pH values were controlled between 6.8 and 7.6. The brains were sliced to sample sizes of $\varnothing 30\text{mm} \times 1.5\text{mm}$ and $\varnothing 30\text{mm} \times 2\text{mm}$ thick with 75% to 90% white matter using a custom-designed cutting jig. Vaseline was applied to the outer edge of the sample to prevent moisture loss. Slippage between the sample and the plates was minimised by using sandpaper.

The stress relaxation shear tests were carried out at 0.02 seconds strain rise time and strain amplitude ranging between 0.0111-0.0666. After a sudden shearing displacement to the sample, the measurement of the sample was held at this strain for 3000 seconds. The data from the rheometer were converted to a PC for further computational analysis.

RESULTS

Figure 1. presents a typical test result for bovine brain tissues. The experimental results show that relaxation behaviours of the different specimens are similar in shape.



DISCUSSION

In this work, the data shows the brain tissue has an exponential stress relaxation with time, and the form of the response is consistent across different loading amplitudes. No stress plateau is seen in the first 3000 seconds, indicating a fluid-like behaviour. This is similar to the results reported by Arbogast et al. and Galford and McElhaney. Future experiments will establish the viscoelastic limit, calculate the relaxation spectrum and lead to the development of constitutive equations for brain tissues that account for their non-linear time dependent behaviour and anisotropy.

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PRECISION OF THE VESTIBULO-OKULAR REFLEX IN ATHLETES WITH DIFFERENT ROTATIONAL EXPERIENCE

Wolfgang Stangl, Albert Gollhofer

Department of Sport Science, University of Stuttgart, Germany

In normal subjects it is well documented that the vestibulo-ocular reflex (VOR), generated by head accelerations, is not responding precisely as a function of the dynamic parameters. Although, vestibular input has a basic sensory function in the recognition of orientation and movement, there exists only little information from sports elite athletes whether or not this reflex system is subjected to training experience or to discipline specific alterations. Therefore, the present study was designed to investigate the precision of the VOR in athletes with different rotational training requirements.

On the basis of the three-dimensional induction coil technique (Fetter 1987) since few years it is possible to analyse the properties of the VOR as a function of three-dimensional induced rotations. The subjects are seated in a completely dark capsule which can be rotated around individually determined axes. The axes of rotation were fitted to coincide with the three directions of the vestibular organ. Oscillations (amplitude ± 20 deg.) as well as continuous rotations (90 sec) were applied in the vertical („upright position“), in the supine („back position“) and in the 90 deg rotated position („right ear down position“). In each condition the acceleration ($100 \text{ deg} \cdot \text{s}^{-2}$) and the velocity ($100 \text{ deg} \cdot \text{s}^{-1}$) was kept constant. Stimuli were applied only around the vertical axis in order to prevent contribution from the otoliths. Competition gymnasts ($n=7$) with a three-dimensional rotation experience, professional ballet dancers ($n=7$), usually adapted to only one-dimensional rotations, and volleyball players ($n=7$), with no specific rotation experience, served as subgroups. In order to verify possible training effects the volleyball players were subjected to an intensive and systematic rotation intervention program of two weeks.

VOR was dominantly sensitive only in that direction the rotation stimulus was performed. The gain matrix (Tab.1) from all three subgroups shows considerable precision only in the diagonal structure. Compared to the findings from normal subjects the dynamic VOR in the highly trained athletes is significantly reduced. Within the different subgroups it can be shown that the precision is depending on the rotational experience both for the oscillation stimuli as well as for the rotations. The spatial precision represented by the non-diagonal values in the matrix revealed similar magnitudes as in normal subjects. Even, if the alterations couldn't be marked significantly, after two weeks of rotation training the volleyball player demonstrated reduced VOR precision.

	C_T (ccw)	C_H (up)	C_V (right)
E_T	$-.35 \pm .10$	$.00 \pm .04$	$.00 \pm .04$
E_V	$-.06 \pm .05$	$-.42 \pm .17$	$-.02 \pm .04$
E_H	$-.14 \pm .05$	$-.02 \pm .06$	$-.37 \pm .15$
E_T	$-.27 \pm .09$	$.00 \pm .06$	$-.01 \pm .06$
E_V	$-.09 \pm .05$	$-.39 \pm .07$	$-.03 \pm .09$
E_H	$-.14 \pm .07$	$-.07 \pm .07$	$-.39 \pm .15$
E_T	$-.45 \pm .09$	$.01 \pm .06$	$.01 \pm .09$
E_V	$-.04 \pm .08$	$-.57 \pm .18$	$.00 \pm .07$
E_H	$-.06 \pm .16$	$-.06 \pm .07$	$-.51 \pm .19$
E_T	$-.38 \pm .11$	$.05 \pm .04$	$.01 \pm .08$
E_V	$-.07 \pm .06$	$-.54 \pm .15$	$.00 \pm .08$
E_H	$-.13 \pm .06$	$.04 \pm .06$	$-.54 \pm .21$

GYMNASTS

BALLET-DANCER

VOLLEYBALL PLAYER PRE-Values

VOLLEYBALL PLAYER POST-Values

Tab. 1: 3x3 Matrix demonstrating the gains (Induced head velocity C_i / Observed eye-velocity E_i) obtained from the nystagmus after $\pm 20^\circ$ oscillations around the following axis: T (torsional axis), V (vertical axis), H (horizontal axis). The values are mean $\pm$ st.dev. from all subgroups. Additionally depicted are pre and post values of the two week training period performed from the Volleyball group ($n=7$).

It can be concluded that the vestibular input alone does not explain the differences in coordination necessary in the investigated sports disciplines. Moreover it may be assumed that in motor tasks with specific accentuation to the rotational orientation the suppression of the VOR is achieved by fixation. This improves the optical control of the visual field, but reduces the dynamical VOR. The observation that VOR loses precision already after a two week training period suggests the necessity of high plasticity in the adaptation in the various sport disciplines.

KINETIC DESCRIPTION FOR SEVERAL REGIMES OF MUSCLE CONTRACTION

V.P.Tregoubov and V.V.Baranov

Institute of Mathematics and Mechanics, St.-Petersburg State University, St.-Petersburg, Russia

INTRODUCTION

The correct mathematical description of muscle contraction must include both a model of muscle's contractile element and reflect its elastic properties. In spite of certain contradictions within experimental data the sliding-filament concept can presently be considered as the most valid basis for a mathematical description of a muscle contraction. Since A.Huxley (1957) suggested the well-known equation for the describing of the sliding-filament mechanism many research groups have presented different modifications. The most successful attempt has been made by Deshcherevsky (1971). He proposed the three-stage scheme of the cross-bridge work and governing equations which, in contrast to Huxley's equations, do not require the introduction of new adjusted parameters or the prescription of various values to the model parameters to describe various regimes of contraction. Nevertheless, Deshcherevsky's model is not useful in cases of the muscle stretch and it is useless for macroscopic descriptions of the whole muscle because it does not contain passive viscoelastic elements.

METHODS

Muscle has been modelled as the four-component mechanical structure that combines a contractile element, subjected to improvement by Deshcherevsky's equations, with a mechanical network including the damped serious elastic element and the parallel elastic one. To describe an eccentric regime of the contraction it was necessary to modify a scheme of the working cycle of the cross-bridges as shown on the figure, where γ , n , m , p are the number of the cross-bridges in free, "pulling", "braking", "superpulling" states respectively. The governing equations for the proposed model can be written as follows:

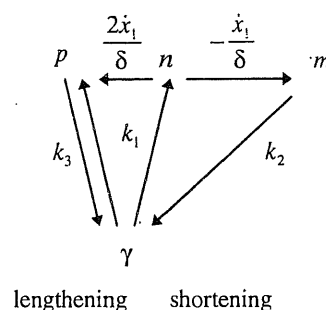
$$\frac{dn}{dt} = \varphi(\dot{x}_1)k_1(t)(\alpha - n - m - p) + \frac{1}{2N_1\delta} \frac{dx_1}{dt}(n - p),$$

$$\frac{dm}{dt} = -\varphi(x_1) \frac{1}{2N_1\delta} \frac{dx_1}{dt}(n - p) - k_2m,$$

$$\frac{dp}{dt} = \xi(\dot{x}_1)k_1(t)(\alpha - n - m - p) + \frac{n}{N_1\delta} \frac{dx_1}{dt} - k_3p,$$

$$M \frac{d^2x}{dt^2} = F(x_1, n, m, p) + P,$$

$$\frac{dx_2}{dt} = \frac{1}{b_2}(-c_2x_2 + c_1x_1 + f_0(2p + n - m)),$$



where P is external load, M is attached mass, $x = x_1 + x_2$ is the muscle length change, x_1 , x_2 are the length change of contractile element and serious elastic one respectively, f_0 is the value of a cross-bridge force, δ is the length of "pulling" region, φ and ξ are characteristic functions, $\alpha = \gamma + n + m + p$.

RESULTS

Three kinds of the regimes of the muscle contraction were reproduced by means of the derived equations for the mechanical model. The isometric regime was realized by fixing the ends of the mechanical model. It should be remarked that the S-shaped kind of the curve in the initial stage of the contraction has been obtained only with the activation rate constant k_1 being dependant on the time. The transient regime after stepwise change of the load was reproduced. The frequency and the amplitude of the model force oscillation in such a transition are appropriated to the experimental data cited by Deshcherevsky (1971). The eccentric regime was modelled by using of the proposed scheme of the working cycle of the cross-bridges. All computer simulation results closely fit to the experimental data obtained by other authors.

DISCUSSION

On the one hand using Deshcherevsky's approach allowed us to obtain Hill's well-known equation for the steady-state case. On the other hand only the proposed complications provided a correct reproduction of the above regimes of muscle contraction.

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NONLINEAR TWITCH SUMMATION OF THE HUMAN TIBIALIS ANTERIOR

J.J. Dowling and P. Kennedy

Department of Kinesiology, McMaster University, Hamilton, Ontario, Canada

INTRODUCTION

The mechanical response of muscle to a train of stimuli has been shown to be a nonlinear summation of the twitches. Adding a single pulse at a short interpulse interval has been shown to significantly increase muscular tension in the cat. Properties of stimulus trains that produce maximum force-time integrals have shown that each additional pulse has an initial less than linear summation followed by a more than linear summation (Parmiggiani and Stein, 1981). The optimum timing of the pulses has been thought to depend on the speed of the isometric twitch (Burke, et al. 1976). The purpose of this study was to compare the nature of the nonlinear twitch summations of human tibialis anterior to results obtained from in situ cat preparations. It was hoped that a model of the nonlinear summations could be developed for use in FES to optimally stimulate paralysed muscle.

PROCEDURES

Nine healthy subjects had the right tibialis anterior muscle stimulated via the peroneal nerve while situated in an isometric dorsi-flexion jig. The jig allowed the measurement of ankle torque which were sampled at 100 samples per second for 0.5 seconds and stored on a computer. Trains of pulses varying between one and four were delivered with inter-pulse intervals that maximized both the peak force and the impulse. The contribution of each twitch was determined by subtracting the previous train (i.e. the contribution of the fourth twitch was determined by subtracting the optimal three twitch record from the four twitch record). The nonlinearity of the summation of each additional twitch was determined by subtracting the initial twitch from each contribution (values < 0 indicate a "less than" linear summation - values > 0 indicate a "more than" linear summation).

RESULTS

Figure 1 shows, (A) the response to the four trains of stimuli, (B) the contribution of each stimulus and (C) the nonlinearity of the summations for a single subject. Similar to the cat data, subsequent twitches were larger and slower than the initial twitch. Unlike the cat data, however, the size of the doublet effect was much larger and none of the subjects showed an initial less than linear summation for the second twitch (2-1) and only four subjects showed it for the third twitch (3-1). The contraction time for this subject was 70 ms yet the fourth pulse does not reach its maximum contribution for 180 ms. Optimal interpulse intervals ranged from 4 ms to 10 ms but there was not a significant relationship between the speed of the isometric twitch and optimal interpulse intervals.

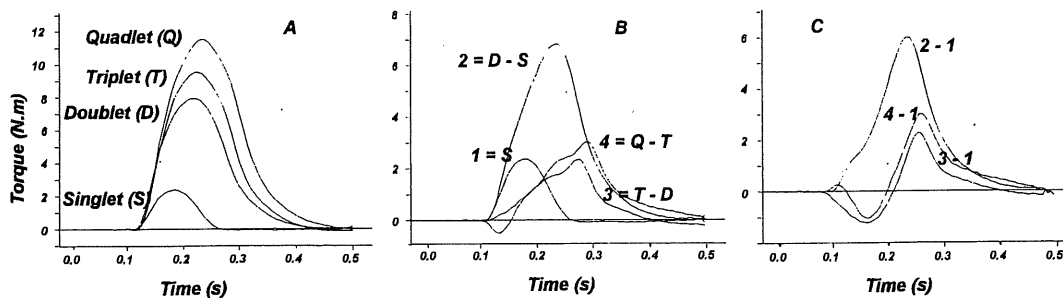


Figure 1. A: Pulse trains consisting one (Singlet) to four pulses (Quadlet) at 10 ms intervals. B: contributions of each pulse. C: Nonlinearity of the twitch summations.

DISCUSSION

The greater compliance of the in vivo human preparation was likely the cause of the larger effect of the more than linear summation when compared to the in situ cat preparations. The considerable variation between subjects makes it difficult to predict the nonlinearity of human tibialis anterior with a model. Perhaps a quantification of series elastic stiffness might account for much of the variability.

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A CUBIC FUNCTION DESCRIBES HAMSTRINGS (ECCENTRIC) / QUADRICEPS (CONCENTRIC) ISOKINETIC TORQUE RATIOS IN PROFESSIONAL FOOTBALLERS.

G. Garbutt and W. Dunn.*

Staffordshire University, Stoke-on-Trent, UK
University of Sunderland, Sunderland, UK*

INTRODUCTION

Eccentric muscle actions have a potential role in the aetiology of hamstring muscles injuries (Stanton and Purdam, 1989). Hamstring function is conventionally described using the isokinetic, concentric, hamstring/quadiceps peak torque ratio (Dvir, 1994) which does not account for the type of muscle action and the joint angle at which torque occurs. This information is important when considering the relative strength of antagonistic muscles in relation to normal muscle balance. This study aims to describe mathematically the angle specific torque ratio between eccentrically acting hamstrings (He) and concentrically acting quadiceps (Qc) throughout knee extension.

METHODS

Eight professional soccer players acted as subjects. Mean age, height and mass ($\pm$ SD) were 22.1 ($\pm$ 2.3) years, 179.8 ($\pm$ 5.3) cm and 80.4 ($\pm$ 16.0) kg respectively. Each performed a 5 min warm-up on a cycle ergometer at 120 watts followed by stretching of the leg musculature. Isokinetic torque produced by the knee extensors and flexors was determined on a Lido Active isokinetic dynamometer (Lido, Loredan, USA) according to the protocol in Table 1. Table 1. Protocol for measuring isokinetic peak torque of the knee extensors and flexors.

Muscle Action	Angular velocity (rad.s-1)	Repetitions
Quadiceps concentric/Hamstrings concentric	2.094	6
Quadiceps concentric/Hamstrings concentric	1.047	4
Quadiceps concentric/Hamstrings eccentric	1.047	4
Quadiceps eccentric/Hamstrings concentric	1.047	4

The He/Qc ratio was derived and plotted against knee joint angle every 5 degrees throughout the range of motion from 15 - 85°.

RESULTS

A cubic relationship was shown to exist between He/Qc ratio and the joint angle at which they occur ($P < 0.05$). This is illustrated in Fig 1 and described by the cubic regression equation:

$$y = 0.8425 + (0.0129)x + 0.0005x^2 + (3.4e^{-0.5})x^3$$

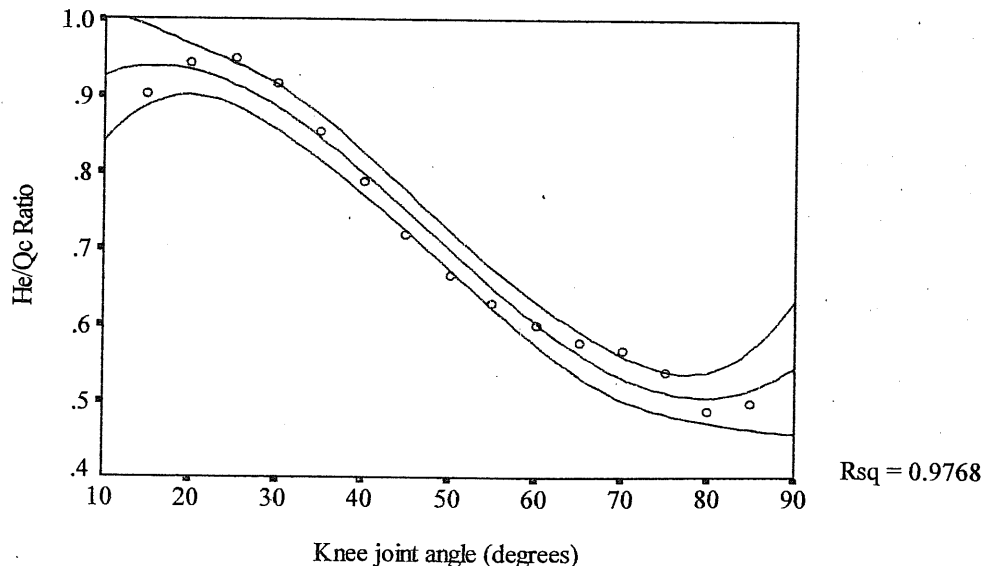


Figure 1. He/Qc ratios relative to knee joint angle, showing 95% confidence intervals.

DISCUSSION

This model takes into account the mode of action of the antagonistic muscles and the knee joint angle at which forces were produced. More comprehensive than previous models it compares torques produced via the active contractile component of the quadiceps with the maximal potential resistive properties of the eccentrically acting hamstring antagonist. Further research is required to evaluate the potential for such models to be used as a means of describing normal knee function and identifying pathologies.

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A METHOD FOR MEASURING VOLUNTARY MUSCLE ACTIVATION DURING DYNAMIC CONTRACTIONS

R.D. Herbert, S.C. Gandevia and J.B. Leeper

Prince of Wales Medical Research Institute, University of New South Wales and Faculty of Health Sciences, University of Sydney, Australia.

INTRODUCTION

The twitch interpolation method has been widely used to measure voluntary muscle activation during maximal isometric contractions (Merton, 1954). This paper describes a modified twitch interpolation method which enables high-resolution measurements to be made of voluntary activation during maximal dynamic contractions. The apparatus has been designed to measure voluntary activation during dynamic contractions of the elbow flexor muscles at elbow flexion velocities of up to 300 degrees.s⁻¹.

METHODS

Subjects are seated with the forearm firmly strapped to a rigid plate which rotates freely about the elbow axis. A strain-gauge type load cell connects the plate to a beam which otherwise rotates independently about the same axis, and which can be loaded with weights. Subjects are asked to perform maximal concentric contractions of the elbow flexor muscles by lifting the weights as fast as possible. A light switch is triggered as the bar passes through the horizontal, signalling a constant-current stimulator to deliver a single, 0.1 ms supramaximal stimulus to the elbow flexor muscles through electrodes firmly stuck to the skin over the mid-belly and distal tendon of the biceps. Force acting on the load cell is sampled at 1000 Hz with a sample-and-hold amplifier set at high gain. Angular velocity is obtained by feeding the output of an optical encoder through a frequency modulator. Integrating and differentiating this signal gives angular displacement and acceleration. Muscle torque is determined by correcting load cell torque for torques due to the inertia and weight of the arm and plate.

The stimulus produces torque transients ("interpolated twitches") superimposed on the torque produced voluntarily. The amplitude of these twitches, divided by the amplitude of twitches evoked from the resting but potentiated muscle, provides a measure of voluntary activation (Allen, Gandevia & McKenzie, 1995). During non-isometric contractions such as those employed in this study, muscle fibre length and velocity is continually changing, so muscle torque changes even when the level of muscle activation is constant. Therefore the amplitude of interpolated twitches is measured as the increase in force above the linear trend of the 25 ms of force immediately preceding the stimulus.

Resting twitches are obtained by stimulating the relaxed muscles as the arm is swung through the horizontal at a range of angular velocities. Surface EMG obtained from electrodes placed over triceps and brachioradialis muscles is displayed on an oscilloscope at high gain to confirm that subjects are relaxed at the time the stimulus is delivered. The size of the resting twitch declines linearly with angular velocity. Voluntary activation is calculated as $100 \times (1 - a/b)$ where a is interpolated twitch amplitude and b is the amplitude of the resting twitch at the same angular velocity as the interpolated twitch, calculated from the linear regression of resting twitch amplitude on angular velocity.

RESULTS

We have now tested 8 subjects with this method. Interpolated twitches were observed in all subjects during some maximal concentric efforts. The time course of these twitches scaled appropriately with twitch amplitude. It is concluded that healthy unfatigued subjects are usually capable of near-maximal or maximal voluntary activation during maximal concentric contractions of elbow flexor muscles.

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CONTACT PRESSURE DISTRIBUTION ON THE HUMAN FINGERPAD

D.T.V. Pawluk and R.D. Howe
Harvard University, Cambridge, MA, USA

INTRODUCTION

As a finger touches a rigid object, the skin contact area begins as a small spot, then expands to subtend much of the fleshy pad at the fingertip. The resulting pressure distribution across the skin is an important determinant of the mechanical behavior of the hand-object system in manipulation. It is also important for understanding the local mechanical conditions to which human tactile receptors respond. Previously, Dandekar [1] developed a detailed 3-D finite element model of the fingerpad; however, only data of the net force response was available for comparison. Experimentally, Srinivasan [3] examined the surface deformation of the fingerpad to a static line load for which he proposed a model of the fingertip as a fluid-filled membrane. This paper investigates the pressure distribution of the fingerpad to more complex stimuli as a function of indentation depth and velocity.

METHODS

A flat tipped motorized indenter applied controlled indentations normal to the fingerpad of the index finger. Sensors measured the position, acceleration and force on the probe. In addition, the distributed pressure was measured using an 8 element by 8 element capacitive pressure sensor with a 2 mm interelement spacing. A quasi-static indentation trajectory (0.2 mm/sec) was first applied to examine the elastic response of the system. In addition, fast (60 mm/sec) ramp and holds, as well as sinusoidal trajectories, were applied to investigate the dynamic response.

RESULTS

For all input trajectories, the contact area and the peak pressures increased significantly with increasing net force. The pressure distributions also appeared symmetric about the medial line of the finger (Figure 1), although there was a clear asymmetry in the proximal-distal direction of the fingerpad. Examining the form of the quasi-static distribution suggested that local pressure varied roughly parabolically as a function of radial distance from the center of contact (Figure 2). As a function of displacement and time, all the localized pressure responses showed nonlinear increasing stiffness and linear force relaxation to the input trajectories, similar to the net force response.

DISCUSSION

The parabolic form of the pressure distributions suggest that our data, at any fixed time, is more appropriately described by a Hertzian distribution than by a fluid-filled membrane model [3]. The data further suggests that a model combining Fung's quasi-linear viscoelastic model of tissue [2] with Hertzian contact can describe the mechanics of the fingerpad. We use such a model to fit the quasi-static response in Figure 2. The amount of error in the fit was small ($mse = 7\%$) and is mostly accounted for by the proximal-distal asymmetry. Future analysis will consider the verification of this model.

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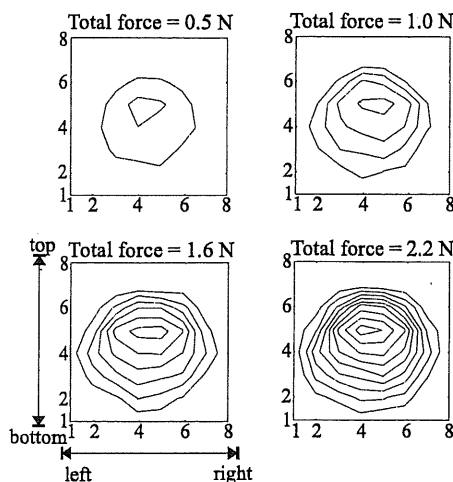


Figure 1: Contour plots of the distributed pressure response to a quasi-static indentation. Contours indicate increasing pressure levels of 4 kPa.

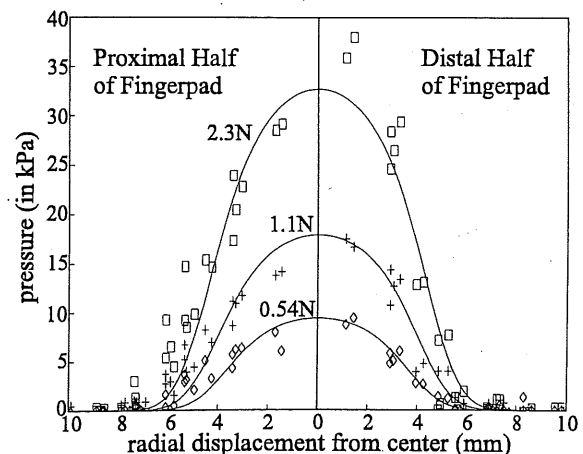


Figure 2: Pressure profiles in response to quasi-static indentation at forces of 0.54 N, 1.1 N and 2.3 N. Measured pressures are indicated by symbols. Solid lines are the model fit.

A NEW SYSTEM FOR THE MEASUREMENT OF THE FORCES GENERATED DURING LUMBAR SPINAL MANIPULATION

Philip J. Rowe, Jonathan McCrea & Patrick M. Salter

Department of Physiotherapy, Queen Margaret College, Edinburgh, UK

INTRODUCTION

Back pain is the most common complaint in physical medicine with some 80% of the population suffering from back pain during their lives (Waddell 1987). These disorders of the spine constitute a major cause of patient referrals to physiotherapists and other health care workers and also represent a substantial proportion of the physical disability prevalent in industrial societies (Nachemson 1991). Manual mobilisation of the lumbar vertebra is a commonly used method of conservative treatment for Low Back Pain (LBP). In this technique the patient is placed in the prone position on a treatment couch and the therapist applies a series of posterior - anterior forces to selected vertebra in the lumbar spine. This judicious application of force to the lumbar spine is designed to improve spinal motion and reduce pain and hence reduce the disabling effects of LBP. Little is known of the nature of the forces applied during manipulation, the benefits, in terms of pain relief and improved function, which accrue to the patient, or the mechanism by which these changes occur. These issues remain a topic for debate in the literature.

Recently considerable interest has been shown in the mechanics of the low back and in modelling the effect of spinal mobilisations on the lumbar spinal complex. In order for these mechanical models to successfully predict the response of the spine to treatment the forces input to the models must reflect current forms of treatment in terms of the magnitude of the forces applied, the duration of the application and the variations of force which occur during the treatment. Currently this data is not available to the researcher interested in the effect of spinal mobilisation on the mechanics of the lumbar spine.

METHOD

Few techniques have been reported which can measure the forces generated during spinal mobilisation. Those methods which have been used can be criticised in that they were expensive, complex, replaced or interfered with normal treatment or had a fixed location. This paper reports a new method of recording the forces generated during treatment. The system uses four, strain gauged force transducers, one placed under each leg of a treatment couch, to measure the vertical force developed during posterior - anterior mobilisations of the lumbar spine. The force transducers are powered by four strain gauge amplifiers the output of which is fed to an analogue to digital converter mounted in an IBM PC AT computer. Purpose built software is used to collect the data at fifty Hertz for up to a minute. The system is simple to use, inexpensive, portable and provides immediate results. It is therefore suitable for widespread use in hospital departments.

RESULTS

The system has been calibrated and exposed to a series of experiments to check the validity of the data obtained. These tests have included the effects of repeated loading and unloading, loading using forces orthogonal to the desired axis, moments around the principal axes, eccentric loading, dynamic loading, electrical noise, heat and mechanical vibration. None of these confounding variables has been shown to significantly effect the response of the system. A pilot study has been carried out, in which eight experienced physiotherapist conducted posterior - anterior mobilisations. The results indicate that within therapists there is some consistency in the pattern and magnitude of the forces generated. The peak forces generated during a series of tests and between mobilisations within a test varied by approximately 20% of the average value. However between therapists there were major differences in the pattern and magnitude of the forces applied. The magnitude of the average force applied varied from 100N to 400N and the variation of the force within the treatment was also different in different therapists.

DISCUSSION

The results of this study indicate that the new system is suitable for the scientific evaluation of the forces developed during manual mobilisation of the lumbar spine and a full scale clinical trial of the equipment is currently underway. Preliminary results from the pilot study indicate that significant differences are apparent in the magnitude and pattern of force exerted by different therapists. Researchers wishing to model the mechanical response of the lumbar spine to treatment should therefore base their models on data from a range of therapists while those that wish to evaluate the outcome of manual mobilisation should ensure that the effects of the therapists are equally distributed between comparison groups.

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FUNCTIONAL EVALUATION OF PSOAS MAJOR MUSCLE BY THREE-DIMENSIONAL MUSCLO-SKELETAL MODEL

Takeo Nagura¹, Yutaka Yabe¹, Kouichi Wakano² and Nobutoshi Yamazaki³

¹Department of Orthopaedic Surgery, Keio University, ²Department of Orthopaedic Surgery, Kawasaki City Hospital, ³Department of Biomedical Engineering, Keio University, Tokyo, Japan

INTRODUCTION

Psoas major muscle, the largest muscle at the lower lumbar levels, may have some important functions on lumbar spine, pelvis and hip joint. However, only the partial function on lumbar or hip region has been investigated. The purpose of this study is to investigate comprehensive function of psoas major muscle, by three-dimensional musculo-skeletal model constructed from MR images.

METHODS

Axial MRI scans of 5mm thick from L1/2 to lesser trochanter of femur with supine hip extended posture were collected from fifteen male volunteers using 1.5T system. Geometrical data of psoas major were obtained by tracing the circumference of the muscle on each slice. Anterior, lateral radiographs of eight male volunteers in the erect posture were digitized to determine the position of mechanical center of each joint. Centroid line approach was used to describe the muscle lines of action (Fig.1). The muscle forces and moments about each joint were calculated by multiplying each cross sectional area by potential of 60N/cm². The muscle actions with various lumbar and hip postures were also estimated by changing the relative position of skeleton in the model.

RESULTS

The difference of the muscle geometrical data among the subjects were not significant. Psoas exerted large compression force (CF) on each level, and anterior shear force (ASF) on L5/S1 and hip joint (Fig.2). Coronal and axial moments were maximal at lower lumbar levels, while sagittal moment was

maximal at hip joint (Fig.3). There were no significant changes of the activities in the differences of posture.

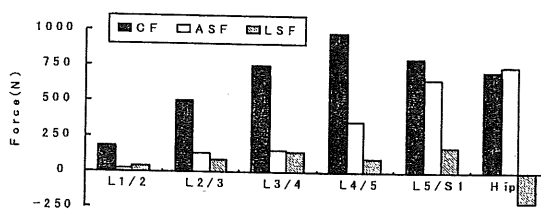


Fig.2 Calculated muscle force

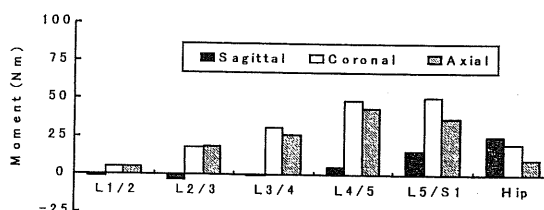


Fig.3 Calculated muscle moment

DISCUSSION

The results show various functions of the muscle. The maximal sagittal moment at hip joint means the flexor activity, while adductor and external rotator activities were also indicated. Lateral flexor activity of lumbar spine agrees with the several electromyographic studies, but it is relatively small comparing to extension moment exerted by back muscles. In spite of the large CF, the lumbar sagittal moments were remarkably small (1/3 to 1/50 of coronal moment). This fact suggest that psoas does not flex but stabilize the lumbar spine on sagittal plane. Moreover, femur may be stabilized to pelvis by CF and ASF, which increase with hip extension. Thus we conclude that psoas major acts not only as an actuator of hip joint, but also as the stabilizer of lumbar spine and hip joint. These functions are reasonable for human erect bipedal locomotion.

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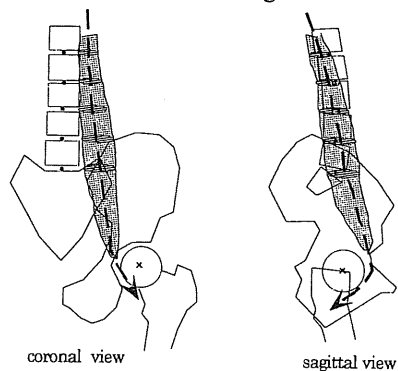


Fig.1 Three-dimensional musculo-skeletal model

A NEW METHOD TO MONITOR DENTAL IMPLANT OSTEOINTEGRATION BY VIBRATION TESTS

Federico Casolo and Vittorio Lorenzi

D.S.T.M. - Politecnico di Milano, ITALY

INTRODUCTION

The time required for the osteointegration of the dental implants -when they can achieve it-, is related to several factors. Surgical technique, strumentarium and implant shape, bone characteristics, such as subjective metabolic factors, can accelerate or slow down the osteointegration process, therefore the correct timing for placing and loading the artificial teeth is unpredictable. The experimental techniques which are now used on animals or volunteers for testing bone-implant interface require the extraction of the implant: the two best methods consist of the measure of the torque required to unscrew the implant and of the analysis of the interface by means of thin istological sections. Radiological methods do not require the extraction but are capable to detect only macroscopic degeneration of the tissue surrounding the implant. Our purpose is to set up and verify experimentally a new, not invasive method capable of monitoring the osteointegration of dental implants. Our aim is to use it as a clinical standard routine for the correct timing of the implant loading and teeth placement, as well as to follow the evolution of pathologic phenomena on perimplantar tissue.

MATERIAL AND METHODS

We assume that, during the healing after the surgery, the tissue around the implant modifies its mechanical characteristics and the boundary conditions; consequently the natural frequencies of the bone-implant system change, as well as the vibrations' amplitude. Because the vibrations frequency is much less affected by the experimental set up, compared to their amplitude, we choose the first as the key parameter to monitor the osteointegration in clinical environments.

Experimental apparatus consists of: a light piezoelectric accelerometer (1.7 gr.) that can be screwed by means of a special support to the standard implant used (PHI); an inductive micro vibrator for impulse generation and a signal conditioning and data acquisition system. The natural frequencies are obtained by f.F.t..

For each test, the impulse generator is gently pressed on the maxilla or mandible bone through the covering soft tissue, inside or outside of the mouth. The impulse transmitted by the bone originate vibrations of the implant.

In order to understand which vibration modes are detected and to try to understand the tissue characteristics variations, we set up mathematical models (Myklestad's meth. and F.E.M) of the bone-implant system. For the validation of the models we compared experimental and theoretical results concerning the vibration induced by the impulse to an implant mounted on a fresh bovine rib. The frequency results are in good agreement even when extra masses are added to the system. The mathematical models also supply the mode shape for natural frequencies.

EVALUATION OF THE EVOLUTION OF THE OSTEOINTEGRATION IN PATIENTS.

Even if it is quite difficult to evaluate separately the mechanic characteristics of the bone and of the other tissues surrounding a dental implant, our new experimental method is able to monitor the evolution of the constraining action of the tissue on the implant globally. Obviously the natural frequency of vibration of an implant depends on many factors such as the size of the implant, its position and the anatomic peculiarities of the patient. Therefore, up to now, it is impossible to check precisely the osteointegration of an implant knowing only its natural frequency at a certain time. On the contrary, our preliminary tests clearly show that the knowledge of the time history of the natural frequency can explain the phases of the healing process; this can be used for instance to determine when it is possible to load the implant. This method also demonstrated to be very sensible to the tissue damages: in a pathological case, in fact the system predicted the bone reabsorption of perimplantar tissue some weeks before the radiographic techniques. In the following example (fig.1) three implants in position 11, 13 and 24 on a patient (female 55 y.o) have been analyzed from the seventh day after surgery. All the data displayed are the mean value of five tests repeated on the same implant. The preliminary results of the tests are very encouraging, and are in agreement with the unscrewing tests on animals that we reported on previous works. They show that this implant has a relevant increase of stability in the first weeks, and then it can probably be loaded without problems.

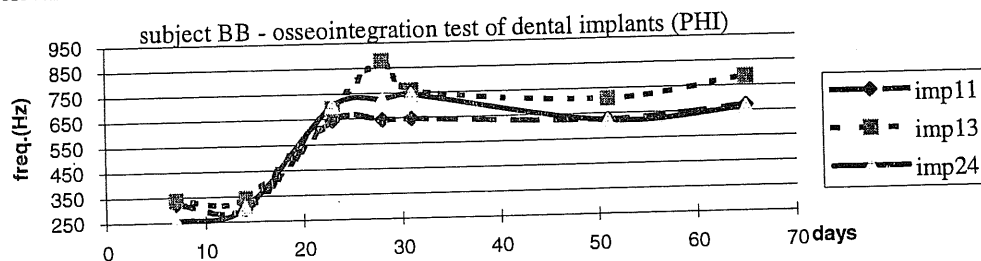


Fig.1: main nat. freq. of 3 impl. (position 11,13, 24) monitored in a patient, by imp. tests, for two months from surg.

CONCLUDING REMARKS

Even if we concluded only few preliminary tests, this new method demonstrated to be powerful and very useful to analyze the perimplantar bone conditions and the strength of the constraints of the implant. The protocol is simple, and tests are easy, relatively fast and not invasive and can be adopted as a standard clinical routine. Up to now we can not draw a conclusion by a single test but we must have the time history of the natural vibration frequency evolution

THE EFFECT OF ELECTRICAL STIMULATION ON SIT TO STAND TASK

M. Galli*, V. Lorenzi*, M. Crivellini*

* Dipartimento D.S.T.M., Politecnico di Milano, Milano, ITALY

* Laboratorio di Bioingegneria della Riabilitazione, Dipartimento di Bioingegneria., Politecnico di Milano, Milano, ITALY

INTRODUCTION

The sit to stand (STS) movement is one of the most analysed movements in biomechanics. This because of its frequent execution in daily life and because of its relevance in allowing people an independent life; moreover it allows to asses coordination and synergy of different body parts (trunk and lower limbs). Even if the kinetic and kinematic analysis of this task was largely discussed in a lot of previous studies, the role of lower limb musculature hasn't been sufficiently analysed yet. Particularly the effect of the strengthening of the lower limb muscles in STS has not been studied to date. Consequently the purpose of the present work is to study the behaviour of the lower limb musculature in STS task and how the STS movement is modified when the electrical stimulation (ES) therapy is used.

METHODS

Eighteen young subjects (from 23 to 30 years, mean 28 ± 3 years; weight 74.6 ± 11.6 kg; height 1.73 ± 0.08 m) participated to this study. A motion measurement system (ELITE, bts®, ITALY) provided the three-dimensional trajectories of reflective markers, a force platform (AMTI, Newton, MA) the ground reaction and the muscular activity was registered by a 6 channel telemetric EMG system.

Reflective markers were placed bilaterally on the acromion, greater trochanter, femoral condyle, lateral malleolus and on the head of the fifth metatarsal. Another reference marker was placed on the chair in order to indicate the seat-off. The EMG signals of quadriceps, gluteus and gastrocnemius were acquired using surface electrodes. The seat, armless and backless, was adjusted vertically for each subject in order to obtain the same knee flexion angle. The subjects were asked to sit in a self-selected position with arms crossed to prevent intersubject differences due to upper limb movements and with the feet over the force platform. The subjects were asked to perform four trials of STS at their natural speed. Eight subjects (4 males and 4 females) were submitted to a cycle of electrical stimulation of quadriceps, gluteus and gastrocnemius (i.e. a daily electrical stimulation, sine wave, $I = 0-150$ mA, for 20 min for each day, for 15 days) and the STS strategy was analysed in three different sessions: before, during and after the ES cycle. The used strategy was analysed from a kinematic point of view analysing the angles α and γ (strategy indexes) shown in Figure 1. By using a 3 segment planar model, from kinetic and kinematic data the torques acting on the hip, knee and ankle joint were evaluated. The anthropometric parameters used in computations were based on Zatsiorsky and Seluyanov¹. The activity of the lower limb muscles was analysed with the EMG signal elaboration (IEMG, RMS...).

RESULTS

The kinematic and kinetic results (Figure 1a) show that the not stimulated subjects, use two different strategies for STS task: a Forward STS strategy (78%) characterised by a fully forward flexion of the trunk as well as high moment at the hip joint and low back with small net moments at the knee and a Normal STS strategy (22%) characterised by a relatively vertical trunk position exerting greater moments at the knee. In Figure 1b are shown the results concerning the electrical stimulated subjects' (mean values and s.d). It is possible to observe that if before stimulation subjects rise with a typical FSTS strategy, the effect of ES is a gradual change from Forward to Normal STS strategy.

DISCUSSION

The change in strategy observed in ES subjects means that, with the strengthening of lower limb muscles, the subjects use the strategy with greater moments at the knee joint and with smaller net moments at the low back. The relevance of these results is that the use of ES therapy for patients with low back problems is particularly useful because the ES subjects execute the STS task without overloading the low back, a cause of worsening of the pathology itself. We are extending the same analysis, before during and after a ES therapy, to elderly people and the former results are very satisfying.

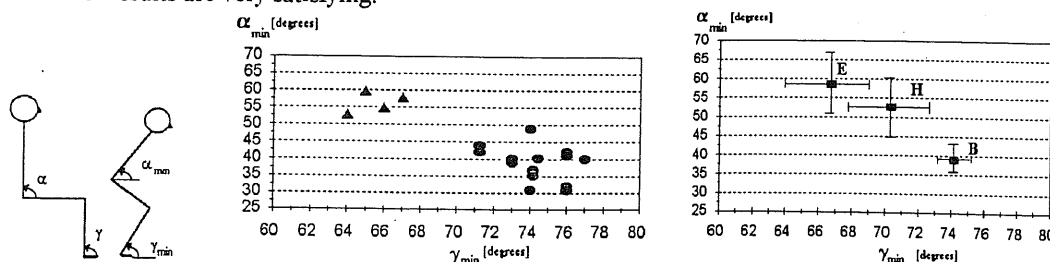


Figure 1: a) strategy used in STS task ● FSTS, ▲ NSTS b) effect of ES (■)B= before H= half and E= end of stimulation cycle.

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ABOUT THE EVIDENCE OF DIFFERENT BALANCE STRATEGIES DURING FUNCTIONAL REACH TEST

Sandro Fioretti, Tommaso Leo and Siobhan Murphy*
Department of Electronics and Automatica - University of Ancona - Italy; (*) CAMARN student

INTRODUCTION

Functional reach (F.R.) is a clinical measure of balance introduced by Duncan et al. (1990) that has been demonstrated, statistically, to have predictive validity in identifying elderly subjects at risks of recurrent falls (Duncan et al. 1992). Aim of this work is to analyse it from a biomechanical point of view, trying to put into evidence different balance strategies during the execution of this motor task. The experimental paradigm has been kept at the minimum level of complexity in order to minimise time and fatigue for the tested subjects once the FR test will be applied to an elderly population. At present, the test as been applied to a population of young subjects. Preliminary results show the presence of two different strategies in performing the motor task.

METHODS

Right-handed young subjects (age 21-25) were analysed while performing the F.R. test. Thirty male and eight female volunteers participated with informed consent. The subjects stood barefoot on a force platform (Bertec Co) and six active markers (IR-leds for CoSTEL stereometric system) were placed on the right side in correspondence of the following anatomical landmarks: lateral malleolus, calcaneus, 5th metatarsal head, greater trochanter, acromion process and 3rd metacarpal head. Consequently, a four segment biomechanical model has been taken as reference in the sagittal plane. A sampling frequency of 25 Hz was adopted for data acquisition. Anthropometric measures included height, weight, trunk length (measured from the spinous process of the VII cervical vertebrae to the coccyx), right shoulder height (acromion to floor), right arm length (acromion to distal end of right third metacarpal), right and left foot length (heel to the end of the great toe), and anterior foot length (anterior aspect of the medial malleolus to the end of the great toe). Subjects were asked to assume a normal, relaxed stance near the centre of the force platform. The inter-toes and inter-calcaneus distances were recorded at the end of the F.R. tests. The subjects were asked to make a fist and extend their arm frontally (position 1) and then to reach as far forward as they could (position 2) without losing their balance or taking a step. No attempt was made to control the subjects' methods of reach. Each subjects was given two practice trials and three test trials. The mean difference, over the three trials, between positions 1 and 2, is the clinical measure of FR. A leveled yardstick placed at the level of the acromion gave the subjects a visual feedback about the performance of their motor task.

RESULTS

FR as reported in literature, is an absolute measure of displacement. In the young subject population it resulted, on average, of 0.4 m (SD=0.04 m). This value is 27% larger with respect to the threshold for high-functioning and low-risk-for-falls subjects in the case of elderly population as reported in Duncan et al. 1990. Normalizing this measure with respect to the height of the subjects it resulted an average value of 23.0 % (SD=2.3%). Correspondingly, the average displacement of the centre of pressure (COP) resulted, in absolute terms, of 0.107 m (SD=0.015 m) and, normalizing with respect to the foot length, it resulted of 40.7 % (SD=5.9%). Consequently kinematic measurements resulted characterized by less variability than the force-platform ones. In the three trials of the F.R. test, each subject showed large variability both for functional reach measure and for centre of foot pressure displacement. This is not the case for the strategies adopted to fulfill the motor task. In fact every subject, in all the three trials, always used the same strategy. Only two different strategies have been recognized. The first kind of motor strategy results in an augmented plantar dorsiflexion; on the contrary, the second kind of motor strategy tends to enhance the plantar flexion by pushing backward the pelvis. This is confirmed by the angles of the leg segment w.r.t. the horizontal axis: in the first case the angle tends to decrease (about (-4) — (-6) deg) while in the second case it tends to increase ((+8) — (+10) deg). The trunk exhibits large flexion angles (40-50 deg) while the absolute orientation of the arm is maintained quite constant throughout the movement. In general the first strategy is accompanied by larger displacements of the COP in the posterior-anterior direction with respect to the second one. Moreover, looking at the COP trajectories it resulted that in all the tested subjects the displacement of this variable in the forward direction is always preceded by an initial displacement in the backward direction. This seems to be correlated with the motion primer, i.e. with the hip mechanism.

DISCUSSION

This paper describes very preliminary results relative to the F.R. test. This test has been performed until now on a population of young subjects with the twofold aim of having reference data with respect to which the performances of elderly subjects will be compared and with the aim of optimizing the assessment of the experimental apparatus before applying the test to the elderly. The first results show typical patterns at the level of strategies adopted to perform this maximal performance task. Statistical as well as dynamical analyses are in progress.

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Acknowledgments: Research funded by I.S.S.

THE SCREW LOOSENING AFTER SCREW TIGHTENING

H. Muratsu*, JT. Andrish**, HE. Kambic**, TE. Manning**

Department of Orthopaedic Surgery, Saijo Central Hospital, Hiroshima, Japan*
The Cleveland Clinic Foundation, Cleveland, U.S.A.**

INTRODUCTION:

The screw tightening generates the compression force under screw head. The retention of this compression force is essential for fracture treatment. The screw loosening would take place after the screw tightening due to the stress relaxation response of the initial compression force. Nevertheless, the magnitude of this stress relaxation have not been investigated.

In the present study, we analyzed the stress relaxation after the screw tightening using the quasi-linear viscoelastic theory (QLT), and evaluated the influence of the stiffness of the fixed bone on the magnitude of stress relaxation: i.e., mechanical screw loosening.

METHODS:

Thirty drill holes for the 3.5mm cortical screws were made perpendicular to the longitudinal axis of canine tibia using standard lag screw techniques. The lag screw tightening was simulated by applying tensile load on the screw head using MTS testing machine. Each drill hole was subjected to the non destructive pull-out test with 300N prior to the stress relaxation study. The stiffness of each drill hole was calculated. The max. load was presumed using linear regression line: max. load = $239 + 0.7 \times \text{stiffness}$ ($r^2=0.8$), which had been obtained by the previous pull out test.

The stress relaxation study was carried out with the standardized initial load of 50% of presumed max. load within 0.25 second. The consecutive change of tensile force after the initial loading was recorded for 1000 seconds.

The tensile force was normalized as the ratio to the initial value: $G(t)$, reduced relaxation function. According to the QLT, $G(t)$ is theoretically given by: $G(t) = A - B \times \log(t)$ [1] (t : time after the loading). By fitting the normalized experimental stress relaxation curve to the equation[1] using the logarithmic regression analysis, the constants A and B were approximated. The correlation of the stiffness to the constants: A, B and $G(1000)$ was evaluated using the simple linear regression analysis.

RESULTS:

Figure 1 demonstrates the typical normalized stress relaxation curve after initial loading. Each stress relaxation properties can be accurately described as linear in the logarithmic scale of time as in equation[1]. The constant A means the residual force 1 second after loading. The constant B represents the slope of the normalized stress relaxation curve in the logarithmic scale of time. It was found each parameters: A, B and $G(1000)$, has a high correlation to the stiffness measured by the non destructive pullout test. The stiffness of fixed bone has a negative correlation with constant B and positive correlation with constant A and $G(1000)$ (Fig.2).

DISCUSSION:

According to the present study, the stiffness of fixed bone has an important role on the screw loosening caused by the stress relaxation. With the less stiffer bone, the more stress relaxation: screw loosening, will take place after the fixation. We have verified that the reduced relaxation function: $G(t)$, calculated and obtained by the first 1000 sec. relaxation period was adequate to predict the stress relaxation at 3 days later. It had also reported that the axial force of bone screw seemed to decrease linearly in the logarithmic scale of time for about 3 weeks in an *in vivo* study. Therefore, these *in vitro* observations might be adequate to predict the stress relaxation in the long term after the fixation. This study reminds that the screw loosening after the screw fixation will differ among drill holes even if surgeon can tighten the screw equally. Especially, screw loosening caused by the stress relaxation must be considerably high in the case of elderly people who have osteoporotic soft bone.

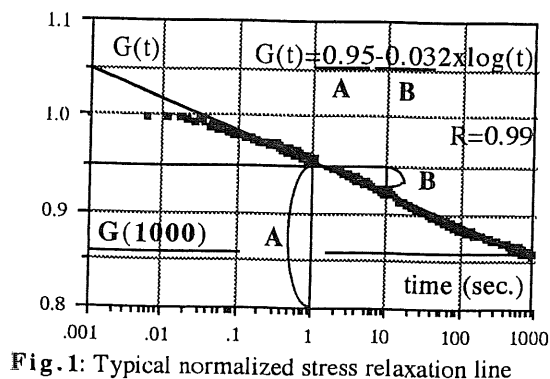


Fig. 1: Typical normalized stress relaxation line

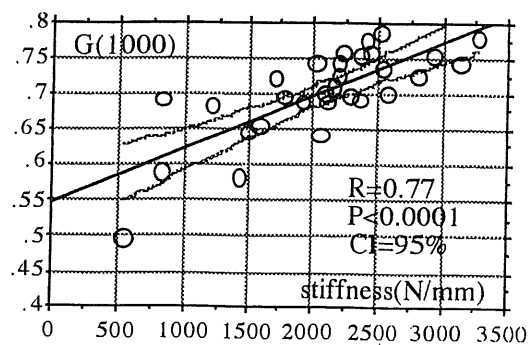


Fig. 2: The influence of the stiffness of bone on the residual force 1000 seconds after tightening

INFUSION OF SALINE INTO THE HUMAN NUCLEUS PULPOSUS AND ESTABLISHING THE PRESSURE-VOLUME RELATION

H. S. Ranu

Columbia University College of Physicians & Surgeons, New York, and
Life College, School of Graduate Studies, Marietta, GA 30060.

INTRODUCTION

Low-back pain can affect almost anyone at any time. In the United States, 8-15 percent of back injuries involve the spinal discs, which act as a cushion between each vertebra. When a human spinal disc slips or herniates, part or all of it bulges out from its usual setting between the vertebrae. The resulting protrusion may cause pain, numbness or muscular weakness, anywhere from middle of the buttocks to the bottom of the foot. The financial cost in the United States due to lost time, medical expenses, and workers compensation amounts to over \$35 billion annually. Over 80 per cent of all low-back problems are directly related to the intervertebral disc. Therefore, to gain further insight into the biomechanics of the human intervertebral disc and to determine a potential mechanism for causation and relief of symptoms related to a herniated disc, this investigation was undertaken.

METHODS

Human lumbar vertebrae with the interposed intervertebral discs and longitudinal ligaments intact were mounted vertically in a stabilizing frame. A miniature pressure transducer on a catheter tip was inserted into the nucleus pulposus. Pressure was measured continuously within the nucleus pulposus in 17 intact lumbar discs from human cadavers. The volume of the nucleus pulposus was increased at the slow, continuous rate of 0.034 ml/min by the pump regulated infusion of saline colored with methylene blue.

RESULTS

In 12 unloaded discs, nucleus pulposus pressure rose in a linear fashion (linear $r=0.96$) from an initial mean pressure of 174 (SD 81) kPa. The mean rate of pressure rise was 327 (SD 109) kPa/ml volume increase. The peak pressure measured was 550 kPa. Similar linear relations were obtained during infusion of saline into five vertically loaded discs fixed at the deformation produced by a 9.1 kg weight. This resulted in an initial mean pressure of 133 (SD 53) kPa, and the mean rate of pressure rise was 218 (SD 87) kPa/ml volume increase.

CONCLUSION

The data define the pressure-volume relation within the disc and show that the nucleus pulposus, surrounded by the relatively inelastic annulus and the solid vertebral end plates, has the properties of a tight hydraulic space in which a large pressure rise will regularly result from a small increase in volume, and opposite is also true. The data thus provide a biomechanical basis for the physiological variation in symptoms related to the disc, and for any benefit obtained from interventions designed to remove the disc tissue e.g. using chymopapain, partial/total nucleotomy/laserectomy of the nucleus pulposus.

EFFECT OF COMPUTER KEYBOARD DESIGN ON WRIST AND FOREARM POSITION WHILE TYPING

G. G. Simoneau, R. W. Marklin, J. F. Monroe

Physical Therapy and Industrial Eng. Depts., Marquette University, Milwaukee, WI USA

INTRODUCTION

While the exact cause of occupationally induced cumulative trauma disorders (CTDs) in computer keyboard users is not known, the ulnar deviation of the wrist and the fully pronated position of the forearm dictated by the design of the conventional flat keyboard are often implicated in the etiology of hand and wrist CTDs. Several "alternative" computer keyboards, with design features aimed at reducing wrist ulnar deviation and excessive forearm pronation when typing, have been introduced into the marketplace. The purpose of this study was to determine the effectiveness of three types of alternative computer keyboards in placing the forearm and wrist in a more neutral position, as compared to the wrist and forearm position present when a conventional computer keyboard is used.

METHODS

Each of the 86 participants in this study were full-time clerical workers between the ages of 21 and 55 who typed for at least two hours per day on the computer. All participants also typed at least 45 words per minute, were asymptomatic of any pain or discomfort that would interfere with typing on a computer keyboard, had been typing for more than 5 years and used a conventional keyboard at work.

Three commercially available keyboards were used in the study: A) a fixed-angle split keyboard, B) an adjustable-angle split keyboard and a C) a vertically-inclined keyboard (resembling a drawbridge).

Each subject used one of the three randomly assigned alternative keyboards in her/his office to perform her/his regular work activities for a minimum of 10 hours. Testing took place in the lab, where electromechanical goniometers were used to measure wrist and forearm motion. Ten samples of 30 seconds of data were collected over a 16-minute period as the subject typed text that appeared on the VDT screen. Each subject was tested using their assigned alternative keyboard and a conventional keyboard (in order to serve as their own control). The order of testing was randomized.

RESULTS

Data are presented below for the mean position of the right wrist and forearm for a total of 300 seconds of typing for each subject.

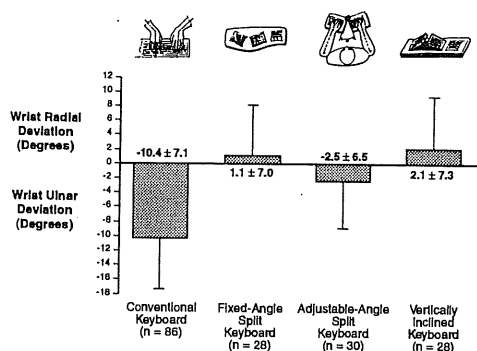


Figure 1. Mean ± 1 S.D. for wrist radial/ulnar deviation angle with the conventional and three alternative keyboards.

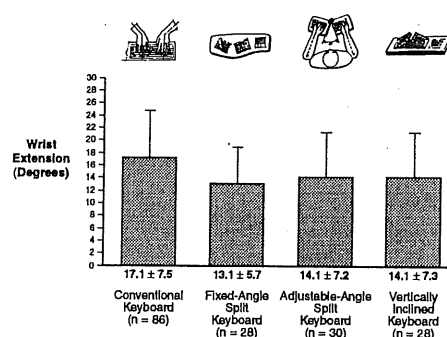


Figure 2. Mean ± 1 S.D. for wrist extension position with the conventional and three alternative keyboards.

DISCUSSION

This study is unique based on the number of subjects investigated and the amount of practice time given to the participants to accommodate to the alternative keyboard. All three alternative keyboards placed at least one component of wrist and forearm posture in a more neutral position than the conventional keyboard. Both split keyboards significantly reduced ulnar deviation ($p < .001$) while the inclined keyboard significantly reduced both ulnar deviation and forearm pronation ($p < .001$). Only small differences were noted in wrist extension angle with the alternative keyboards. Based on our results, it appears that alternative keyboards of the type used in this study place the wrist and forearm in a more neutral posture than a conventional keyboard. The effect of this change of wrist and forearm position on the incidence of CTDs is yet unknown.

ACKNOWLEDGMENTS

The authors would like to thank the National Institute for Occupational Safety and Health (NIOSH) for partially sponsoring this study (#1 RO3 OH03 184-01A1).

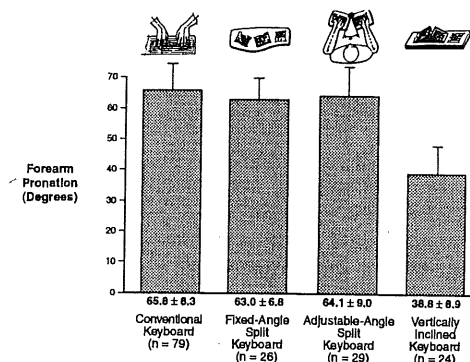


Figure 3. Mean ± 1 S.D. for forearm pronation angle with the conventional and three alternative keyboards.

INFLUENCE OF PREVIOUS ANKLE SPRAINS ON NEUROMUSCULAR RESPONSE DURING SIMULATED INVERSION INJURY

Wilfried Alt, Heinz Lohrer, Albert Gollhofer

Department of Sport Science, University of Stuttgart, Germany
Institute of Sports Medicine, University of Frankfurt, Germany

INTRODUCTION

One of the major reasons of ankle joint instability are recurrent sprains. Especially in basketball there is a high prevalence of ankle injuries. For preventive reasons most athletes use tapes, braces or specific stabilizing shoes to minimize the risk of injury.

The aim of our study was to evaluate neuromuscular and mechanical parameters during a simulated inversion injury and to determine the differences between preinjured and noninjured ankles. The influence of a semi-rigid functional orthotic device on the biomechanics of the simulation should be quantified.

METHODS

22 male ball sports athletes (24.8 ± 5.5 years old) participated in the study. All of them performed in the second or third national division. By clinical investigation, including manual ankle examination, case history and stress x-ray the degree of ankle joint instability was determined. An ankle joint injury was simulated by a special constructed device both for the right and left leg separately: In the upright stance with full body weight support on one leg a tilt platform was released. These stimuli were applied randomized and unexpectedly. The induced displacement was 30 degrees inversion combined with 15 degrees plantarflexion.

Angular excursion was registered „in shoe“ with twin axis goniometers. Neuromuscular activity was assessed with surface electromyography from mm. peroneus lg., tibialis ant., gastrocnemius med. and the vastus med. A total set of 5 inversion stimuli was averaged separately for the condition with and without wearing a semi-rigid ankle brace. All EMG-data were normalized to the condition without ankle brace.

RESULTS

40% of stress x-ray images showed an angle of 5 degrees or more in the talar tilt. There was neither a correlation between clinical investigation and x-ray talar tilt nor between talar tilt and „subjective“ instability score. The induced inversion amplitudes from the platform tilts were reduced significantly ($p \leq 0.01$) from 28.8degrees (± 3.9) without to 18.6degrees (± 7.2) with semi-rigid orthotic device. The rectified and averaged EMG-signals of the m. peroneus following ankle injury simulation were reduced when an orthosis was used. Magnitude as well as integrated reflex responses were decreased to 85% compared to the condition without brace.

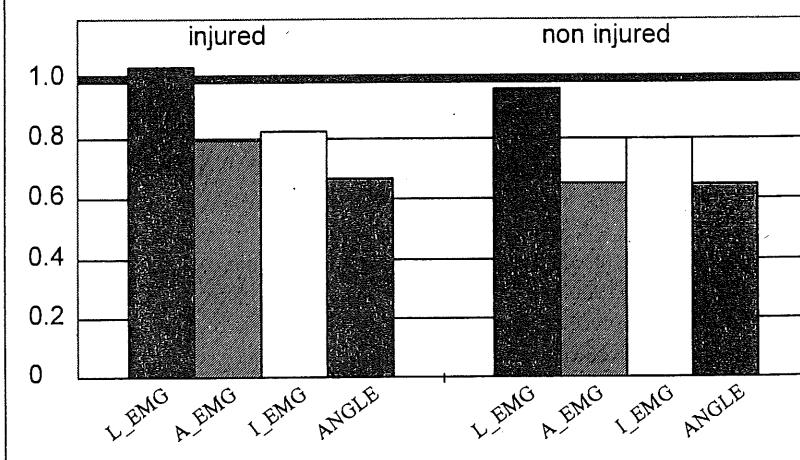
In the m. tibialis ant. and the m. peroneus lg. a significant ($p \leq 0.05$) increase in reflex latency could be found in those subjects with previous history of recurrent giving-way.

DISCUSSION

Prolonged muscle reflex latency due to previous injury is documented in literature. In our investigation we used a testing movement that is very close to real inversion injuries. Whether the mechanoreceptors of the injured capsule and ligaments or the stretch or strain sensitive receptors in the tendons and muscles are affected by recurrent sprains is still under debate. Increased latencies following mechanical stress can either be explained by differences in the injury related set of active sensory receptors or by alterations in the reflex pathway indicating a shift from spinal to transcortical regulation. Although

it is often discussed that mechanoreceptors, especially cutaneous receptors around the joint can modulate proprioceptive response, an increase in reflex activities using a semi-rigid orthosis could not be determined. Moreover, the application of an orthosis reduces EMG during the simulated inversion injury which is primarily a consequence of the reduced mechanical stress for the joint complex. However, in high risk sport disciplines, where prevention of recurrent ankle sprain is important, it seems that all factors stabilizing the ankle joint must be regarded: the mechanical aspects of the orthotic devices as well as the status of the proprioceptive input to the muscles acting around the ankle joint.

Figure 1: Latency, max. amplitude, 100ms integral of m. peroneus and inversion angle β during injury simulation from frequently injured and non injured athletes. Data are normalized to the trial without ankle brace (=1.0)



INFLUENCE OF BODY POSITION ON THE ANKLE RESPONSE TO FUNCTIONAL ELECTRICAL STIMULATION

M. Kljajić^{1,2}, F. Gider¹, M. Tomšič¹, U. Bogataj¹, J. Florijančič²

¹University of Ljubljana, "Jožef Stefan" Institute, Ljubljana, Slovenia

²University of Maribor, Faculty of Organisational Sciences, Kranj, Slovenia

INTRODUCTION

During the implantation of a subcutaneous peroneal electrostimulator the implantable electrode positioning is one of the most important problem. A quantitative method for the prediction and evaluation of ankle movement with implantable peroneal stimulators has been investigated. The methods comprise EMG monitoring the major ankle motion controlling muscles (tibialis anterior, peroneus longus and triceps surae as well as the angles in sagital and transversal plane. Influence of body position to the ankle response invariance during functional electrical stimulation was analysed.

METHODS

Ankle goniogram and EMG response (M-waves) performed in three positions: sitting, lying and standing. Three pairs of in vivo metric surface Ag-AgCl electrodes are used for EMG recording. Each pair of electrodes is placed 30 mm apart and fixed between the muscle belly and muscle tendon. A wraparound grounding electrode is placed between the stimulation and detecting sites. The active stimulation surface electrode was moved over the peroneal nerve until the best foot movement, was found. Through the measurements the stimulation with amplitude of approx. 15 mA, pulse duration of approx. 150 μ s and frequency of approx. 1 Hz was used.

RESULTS AND DISCUSSION

The method has been tested on 9 healthy subjects (average age: 30 $\pm$ 5 years) and 2 CVA patients. ANOVA test was performed to estimate the influence of body position on the ankle response. There is no significant difference between tibialis anterior, peroneus longus as well as dorsal flexion in all three position. EMG response of triceps surae muscle group was negligible, which means that stimulation did not activate the foot plantarflexors. The ankle eversion was significantly different between sitting, standing and lying position ($p < 0.01$). Different proportion of evtor and dorsiflexor activity in the composite movement in two patients can be observed. The optimal activation of each stimulated muscle was determined according to visual assessment of composite ankle movement. The right hemiplegic patient D.D. was also measured by applying dual channel surface stimulation at the motor points of tibialis anterior and peroneus longus muscles in order to elicit a composite ankle movement. The stimulation of tibialis anterior causes inversion of the foot while the stimulation of peroneus longus causes an excessive eversion, only stimulation of both muscles results in a good movement. The representative responses to single stimulation pulse in sitting position are displayed on Fig. 1.

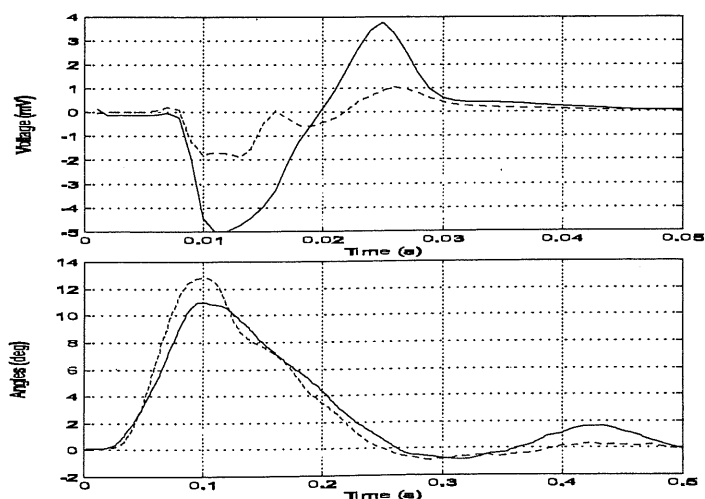


Fig. 1. The ankle response of the patient D.D. to single pulse, in sitting position (Top: M-waves, solid - tibialis anterior, dashed - peroneus longus. Bottom: goniograms, solid - flexion, dashed - eversion).

The new method provides; functional testing of the ankle response before implantation, electrode positioning during the surgical procedure of the implantation and follow up of the ankle response after implantation. In general, M-waves are less sensitive to the body position than goniograms. However, significant impact of body position on ankle eversion response clarifies needs for dual channel implantable stimulation of paretics' ankle movement control.

THE INFLUENCE OF ANKLE FLEXIBILITY ON THE PERFORMANCE OF THREE TASKS

Moseley AM[#], Crosbie J[†], Adams R[†], Dickson H[‡]

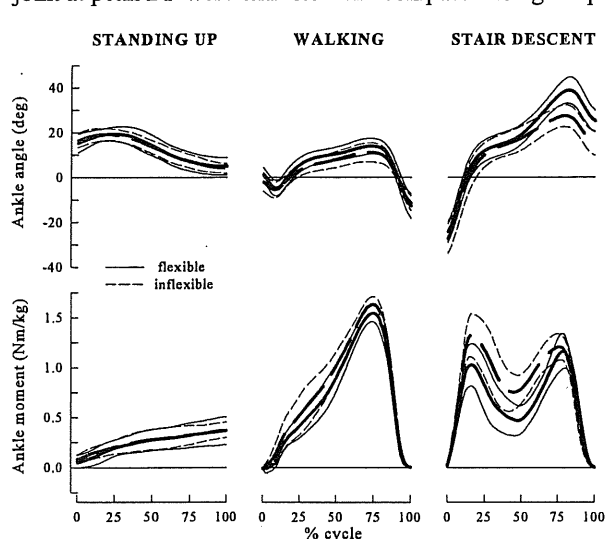
[#] Brain Injury Rehabilitation Unit, Liverpool Health Service, [†] School of Physiotherapy, The University of Sydney,

[‡] Department of Rehabilitation & Geriatrics, Bankstown Hospital, Sydney, Australia

INTRODUCTION: Considerable intersubject variability exists in ankle flexibility measurements in normal subjects, with standard deviation values as high as 7 degrees for passive ankle dorsiflexion movement¹. It has not been established whether this intersubject variability in ankle flexibility affects the performance of everyday motor tasks. Investigation has been limited to the percentage contribution of the passive ankle moment to the total internal moment developed about the ankle during stance phase of walking². The aim of this study was to investigate the influence of ankle flexibility on ankle kinematics and kinetics during standing up from sitting, walking and stair descent.

METHODS: Eighteen healthy male subjects were selected from a larger sample³ based on their passive ankle dorsiflexion movement - the 'flexible' group (N=9) being more than one standard deviation above the mean, and the 'inflexible' group (N=9) being more than one standard deviation below the mean. Both groups were matched for age, height and weight.

Subjects were videotaped using an NEC video camera placed perpendicular to the plane of progression. Reflective markers were placed on the left head of the 5th metatarsal, heel, lateral malleolus, lateral epicondyle, greater trochanter, ASIS, PSIS and acromion to define a 4-segment model. Ground reaction force data were collected for the left leg using an AMTI force platform. For standing up from sitting, seat height was adjusted to 100% of each subjects lower leg length. A 10 m walkway was used for the walking trials. While for stair descent, a staircase consisting of 3-steps, with the force platform forming part of the second step was used. Movement time or cadence was externally paced for each task using a metronome. Joint kinematics and net muscle moments were calculated for the ankle using the synchronised force platform and x-y coordinate data in accordance to the form set by Bresler and Frankel⁴. Peak dorsiflexion (DF) angle, timing of peak DF and the muscle moment about the ankle joint at peak DF were extracted and compared using a repeated measures ANOVA.



RESULTS: Mean (bold lines) and standard deviation ankle angle and muscle moment versus percent cycle curves for the extension phase of standing up and stance phase of walking and stair descent are illustrated in the figure. There was a main effect for both task ($p < .05$) and group ($p < .01$) for peak DF angle, with walking requiring the least DF and stair descent requiring the most. Post hoc testing revealed that peak DF was only significantly greater for the 'flexible' group during stair descent ($p < .01$). Peak DF was, on average, 39.36 (SD 5.71) degrees for the 'flexible' group compared to 27.94 (SD 4.81) degrees for the 'inflexible' group. The groups did not differ in terms of the timing of peak DF and the muscle moment about the ankle joint at peak DF. These two variables were dependent, however, on the task being performed.

DISCUSSION: The results of this study demonstrate that subjects from the extremes of the normal ankle flexibility distribution did exhibit differences in the biomechanics of stair descent but not for standing up from sitting or walking. This finding suggests that subjects must be modulating the active contraction of their plantarflexor and dorsiflexor muscles to achieve the same net muscle moment during standing up and walking. One would expect that pathological loss of ankle flexibility (for example, ankle plantarflexion contractures following traumatic head-injury) may have an impact on standing up and walking.

This study provides insights into the influence of flexibility on the performance of everyday motor tasks. The data generated, in addition to other biomechanical research, provide a normal model to which patient assessments can be compared.

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RELATIONSHIP OF IN-VIVO BONE RESONANCE ANALYSIS TO HEALING CALLUS MICROSTRUCTURE

Mellick J Chehade¹, Nick L Fazzalari², Mark J Pearcy³, Timothy J Lawes¹, Anthony P Pohl¹

¹Department of Orthopaedics & Trauma, Royal Adelaide Hospital; ²Division of Tissue Pathology, Institute of Medical and Veterinary Science, Adelaide; and ³School of Mechanical, Manufacturing and Medical Engineering, Queensland University of Technology, Brisbane, Australia.

INTRODUCTION

The aim of this study was to examine the use of vibrational analysis in the assessment of fracture healing. Techniques that monitor long bone resonant frequency changes have been studied with the aim of indirectly assessing the mechanical properties of healing fractures. No previous studies however have correlated *in-vivo* resonant frequency monitoring with the histomorphometric properties of a healing fracture.

METHODS

Using a sheep tibial fracture healing model, *in-vivo* resonant frequency changes were measured during fracture healing using a bone resonance analysis (BRA) technique and the callus microstructure was determined using histomorphometric analysis. The BRA technique was modified to obtain reproducible resonant frequency determinations in sheep (inter- and intra-observer coefficient of variability < 1.5%). Forty skeletally mature merino wethers were used with 8 sheep randomly assigned to each of 5 groups for culling at 2, 4, 6, 8 or 10 weeks. After application of an external fixator the fracture was created by osteotomy to the tibial mid-shaft and resonant frequencies were obtained both prior to osteotomy and at the time of culling. A BRA stiffness index (SI) was calculated by squaring the fracture to pre fracture resonant frequency ratio. Histoquantitation of callus morphometry¹ was measured from adjacent the fracture line. Direct measurements of Bone Volume and Specific Bone Surface were made from ground sections using an image analyser whereas Trabecular Thickness and Trabecular Number were calculated using a parallel plate model. Changes in the callus microstructure were examined as a function of time and the BRA stiffness index was then correlated with the histomorphometry using Pearson's correlation coefficient.

RESULTS

During fracture healing there was an increase in the bone volume and in the thickness of the trabeculae. The number of trabeculae and their surface area however decreased. Strong and highly significant correlations were found with the BRA and both the callus Bone Volume ($r = 0.77$, $p < 0.00001$, (Fig. 1)) and the callus Trabecular Thickness ($r = 0.76$, $p < 0.00001$, (Fig. 2)). Highly significant *negative* associations were also found with Specific Bone Surface ($r = -0.70$, $p < 0.00001$) and with the Trabecular Number ($r = -0.68$, $p < 0.00001$).

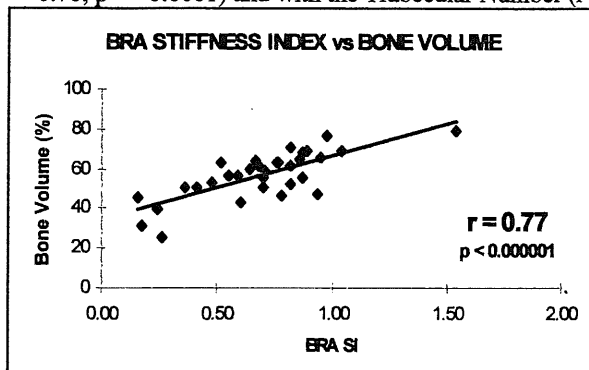


Fig 1

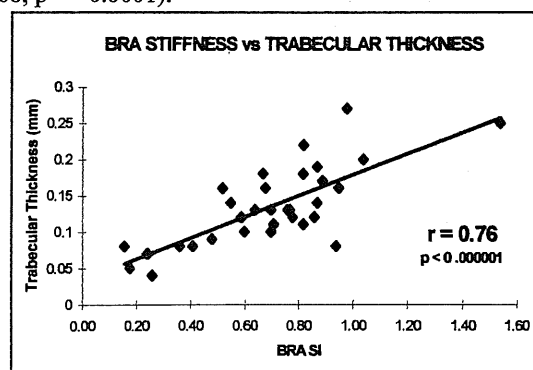


Fig. 2

DISCUSSION

In this study a new and highly reproducible technique of BRA has been developed and described. The changes in callus microstructure during early fracture healing have also been described. This is the first study to assess *in vivo* resonant frequency changes with respect to fracture biology and the results have conclusively demonstrated that a strong relationship exists between the callus microstructure and the resonant frequency using this BRA technique. This suggests that the resonant frequency can be used as an indicator of callus formation and early maturation. This part of the study has not specifically looked at the role of BRA in the determination of an endpoint to fracture healing but these results suggest that BRA should at least have clinical application in demonstrating early progression (or non progression) towards union. This would enable much earlier decisions to be made with regards to possible intervention than is currently possible.

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IS HIKING SAFE AFTER TOTAL KNEE REPLACEMENT?

M. Kuster¹, E. Spalinger², M. Schoch², U. Wyss²

¹Klinik für Orthopädische Chirurgie, Kantonsspital, 9007 St. Gallen, Switzerland

²Sulzer Orthopaedics, Winterthur, Switzerland

Introduction

The most destructive wear in total knee arthroplasty is fatigue, which occurs through repeated high loads above the yield point of polyethylene (25 MPa). Earlier biomechanical studies of knee joint loadings have estimated maximal joint compressive forces to be 3-4 times body weight (bw) during normal ambulatory tasks and this range of values has become a design criterion for most currently used knee joint prostheses. A recent study, however, has shown that tibio-femoral loads during downhill walking may be as high as 8 times bw at flexion angles of 40 deg (Kuster et al). There are no biomechanical evaluations of total knee replacements available for this newly published loads. It was the purpose of the present study to evaluate three different total knee design using tibio-femoral loads as high as 8 times bw in order to see whether hiking may be safe after total knee replacement.

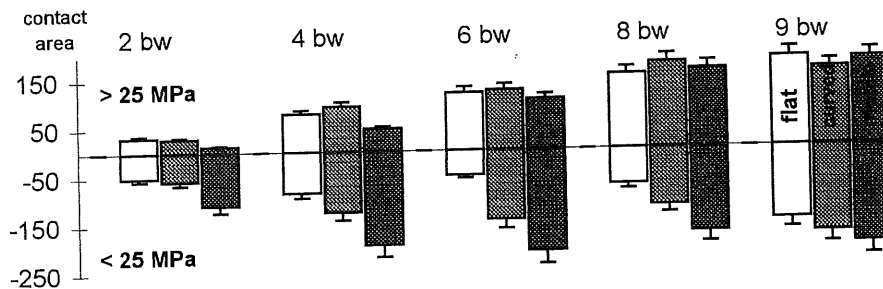
Methods

Three different total knee designs were evaluated using Fuji pressure sensitive films. The designs consisted of a flat tibial inlay, a curved inlay and a "conforming" inlay with mobile bearings. The loads imposed on the prostheses were 2 times bw, 4 times bw, 6 times bw, 8 times bw and 9 times bw at 40 degrees of flexion. Five measurements were conducted for each load. The Fuji pressure sensitive films were then scanned and analysed using an image analysis program. The measures obtained for each load was the total contact area, the mean contact pressure and the contact area below and above 25 MPa, which was considered the yield point of polyethylene.

Results

The mean contact pressure of the conforming design was significantly reduced for all tested loads compared to the flat or curved design. At 8 times bw the flat design had a mean contact pressure of 33.6 +/- 1.2 MPa, the curved design 30.1 +/- 1.1 and the conforming design 26.5 +/- 1.4 MPa. The mean contact pressure was above the yield point of polyethylene for all three designs. Figure 1 shows the contact area above 25 MPa (positive bars) and below 25 MPa (negative bars) for the flat inlay, the curved inlay and the conforming inlay at loads of 2, 4, 6, 8, and 9 times bw. As can be seen the contact area below 25 MPa was significantly bigger in the more conforming inlay compared to the flat inlay for all loads. The contact area above 25 MPa, however, was reduced for the more conforming prosthesis at loads of 2 and 4 times bw only. Above 4 times bw the contact area with stress levels above 25 MPa did not differ for the different design types indicating that the beneficial effect of a more conforming prosthesis decreases with increasing loads. A reason for this finding may be the fact that during downhill walking the loads occur at 40 deg of knee flexion where most currently available mobile bearing designs are not fully conforming. We conclude that fast downhill walking produces high stress levels and may endanger the polyethylene inlay of most total knee designs. The present study considered a static load only and the subsurface stress which may be more important has not been calculated. Thus additional data including dynamic investigations and subsurface stress calculations seem necessary to further evaluate our findings.

Fig. 1



References:

Kuster M et al. J Bone and Joint Surg (Br). In print

HOW TO DEFINE THE CONTACT POINT OF THE TIBIOFEMORAL JOINT OF THE PROSTHESIS IN THE IN VITRO BIOMECHANICAL TESTING

Jiann-Jong Liao, Cheng-Kung Cheng and Chun-Hsiung Huang*

Institute of Biomedical Engineering, National Yang Ming University Taipei, R.O.C.

*Department of Orthopaedic Surgery, Mackay Memorial Hospital, Taipei, R.O.C.

INTRODUCTION

Although some investigators have been reported the contact areas and pressures of the tibiofemoral joint of total knee prostheses, but no standard testing method was reported because there are some difficulties in the testing procedure. The contact point of the tibiofemoral joint plays an important role in the testing procedure but no literature reported how to define it. The objective of this research was to define the contact point of the tibiofemoral joint in the in vitro biomechanical testing.

METHOD

Three well-known commercialized knee prostheses (Omnifit, Genesis, and AMK) were used for testing. A three-dimensional digitizer was used to measure the curves of the femoral component of knee prosthesis and to define the instant rotation center at the center of the flexion radius during flexion / extension. We applied a compression load at the middle line of medial-lateral condyle and the center of the flexion radius of femoral component and used Fuji-prescale film to measure the contact areas at different flexion angles (0° , 10° , 30° , and 60°). We defined the normal contact point at which maximum contact areas achieved and used the sensitivity analysis and digital image analysis of the Fuji-prescale film to find the normal contact point at different flexion angles.

RESULTS

From the results, we found that the normal contact point of the 3 prostheses were all located at 30%-60% of the depth of the tibial dimensions related to posterior edge and symmetrically related to the middle line of the mediolateral width. We also found that 3 prostheses have the same tendency in which the contact point first moves anteriorly during 0° to 10° flexion and then moves more posteriorly after 10° flexion (Fig 1).

DISCUSSION

In this study, we used the Fuji-prescale film and image analysis to find the normal contact point at different flexion angles in biomechanical testing. We have two important assumptions: 1. the contact area is symmetrical between medial and lateral side; 2. the load is symmetrical between medial and lateral side. The assumptions were validated by the contact area and pressure measured in medial and lateral side which had no significant difference. The nonhomogeneous property of the Fuji-prescale film could also contribute to the error of the testing results. The results of the contact point of 3 prostheses were compatible with previous publications.

In conclusion, the present method of the maximum contact areas used to define the contact points during different flexion angles ought to be applicable.

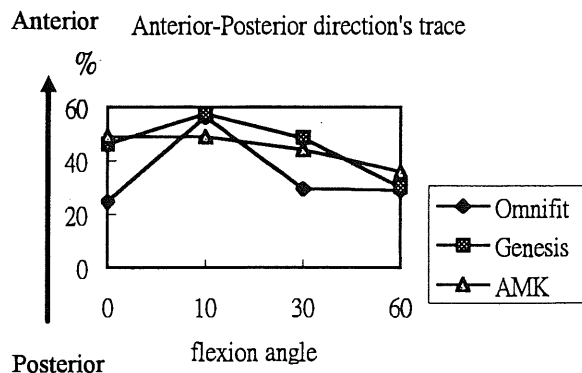


Figure 1 : The contact path in anterior-posterior direction of the three prostheses

3D PELVIC MOVEMENT AND POSTURAL MUSCLE RESPONSES DURING TASK PERFORMANCE IN PARAPLEGIC SUBJECTS.

Y.J.M. Potten^{1,2)}, H.A.M. Seelen¹⁾ and J. Drukker²⁾

1) Institute for Rehabilitation Research, Hoensbroek, The Netherlands.

2) University Maastricht, Maastricht, The Netherlands.

INTRODUCTION

During rehabilitation spinal cord injured (SCI) people have to learn to compensate function loss of the erector spinae (ES) muscle by using parts of the sensorimotor system which are still intact. They learn to use non-postural muscles like the latissimus dorsi (LD) and the ascending part of the trapezius muscle (TPA) to control their sitting balance. Besides differences in neuromuscular control there are differences in gross kinematics between SCI people and able-bodied subjects, especially concerning pelvic tilt and spinal movement. Since SCI people differ from non-SCI people in the ability to tilt their pelvis during sitting, the correlation between pelvic movement and postural changes during task execution was investigated.

METHODS

The experiment included ten people with a complete high thoracic SCI (T2-T8), ten people with a complete low thoracic SCI (T9-T12) and ten people without a SCI. The SCI subjects had completed their active rehabilitation at least six months before and had no secondary pathology. Groups were matched for sex, age, height and weight. Subjects had to perform bimanual reaching movements towards button pairs placed on a table at 15, 30, 75 and 90% of each individual's maximum reaching distance in antero-posterior direction. Muscle activity was recorded bilaterally of the erector spinae (ES) at level L3, T9 and T3, the ascending part of the trapezius (TPA), the latissimus dorsi (LD), the serratus anterior (SA), the pectoralis major (PM) and the oblique abdominal muscles (OA). Sitting balance was monitored by measuring changes in location of the centre of pressure (COP) using an AMTI force platform mounted underneath the chair. Two different chair configurations were tested, i.e. A) 0° tilting and 10° reclining (standard chair) and B) the standard chair fitted with cushion that causing 10° forward inclination of the seat. Pelvic tilt and spinal movement were 3D recorded using four infrared cameras. Passive markers were attached to the following anatomical landmarks: spinous processes at L4, L2, T12 and T10, posterior superior iliac spines, anterior superior iliac spines, greater trochanters and lateral epicondyles of the femur.

RESULTS

Lateral views of marker displacement during forward reaching of a non-SCI, a low SCI and a high SCI subject while sitting in the standard chair are shown in figure 1.

Evaluation of marker displacements while reaching towards 90% of the maximum revealed a clear difference in pelvic movement and flexion of the spine between the three groups. Non-SCI people produce more pelvic tilt and flexion of the trunk compared to SCI people. Low SCI subjects showed less pelvic tilt than non-SCI people but exposed an extension of the upper part of the spine. High SCI subjects revealed hardly any movement, they used the backrest of the chair for support while making the reaching movement. Different levels of impairment resulting in different sensorimotor capacities lead to more possibilities to compensate the function loss of the ES in low SCI. These findings are corroborated by the EMG data of the ES, SA, LD, TPA, PM and OA muscles recorded simultaneously.

DISCUSSION

Paraplegic people are not able to actively tilt their pelvis forward due to the growing lack of activity of the ES at increasing levels of SCI. To compensate for instability of the pelvis and the lower part of the spine, high SCI subjects passively tilt their pelvis backwards and use the backrest of the chair for support, i.e. to increase the base of support. This position is advantageous for a second reason: it creates possibilities for more cranial muscles like the LD and TPA to become active in support of the erect position of the spine. People with a low SCI, having more residual function, counteract the forward displacement of the upper limbs, which coincides with a disturbance of sitting balance, by extending the upper part of the spine and moving the head backwards. In this way the centre of gravity is kept within the base of support.

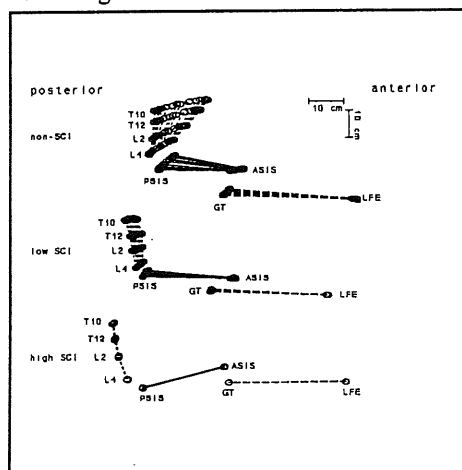


Figure 1. Marker displacements

BIOMECHANICAL EVALUATION OF TAPING IN CONTROLLING KNEE JOINT EXTENSION AND ROTATION ANGLE

Y. Urabe, K. Kawaguchi and K. Onari

Institute of Health Sciences, School of Medicine, Hiroshima University

Kasumi 1-2-3, Minami-ku, Hiroshima 734, Japan

PURPOSE

For the athlete with an injured knee, determining appropriate treatment is often a difficult problem. Although the athlete should regain enough physical activity and function be able to return to the sports activity, lack of research data makes it difficult to make solid clinical decisions. The purpose of this study was to examine the effectiveness an anterior cruciate ligament (ACL) protective taping method, and to search for a rationale to determine a suitable method of taping.

METHODS

Participating in this study were ten healthy athletes, all of whom were active volunteers with no known knee joint ligamentous abnormalities. Five men were college soccer players from the Hiroshima University, and five women were elite basketball players. The ACL taping method was as follows. Many strips of 2 inch wide hard elastic tape were applied beginning on the anterior leg directed proximally and medially, spiraling behind the knee turning posteriorly and superiorly to end laterally on the anterior thigh. This tape was designed to prevent terminal knee extension and lower leg external rotation with respect to the femur in an open kinetic chain (OKC). An OKC knee extension was monitored by a high speed cinecamera (NAC), and the changes of terminal knee extension angle analyzed. Maximum knee extension in sitting was tried before and after taping. Three-dimension space analyzer ISOTRACK (polhemus) was used to register changes of rotation angle between the lower leg and the femur. The rotation angle was measured in both OKC and closed kinetic chain (CKC) motions respect. CKC motion was similar so called "forward range" following knee extension to flexion. Terminal knee extension and rotation angle were analyzed with OKC versus CKC and before versus after taping.

RESULTS

The terminal knee extension angle decreased after ACL taping in OKC. A statistically significant difference ($p < 0.05$) was seen between before and after taping. Knee extension velocity was not restrained after taping. The rotation angle was significant less ($p < 0.05$) after taping than before taping. The rotation angle was not significant different between OKC and CKC.

DISCUSSION

In this study we evaluated the mechanical effects of taping in vivo. It indicated that tape restrain some knee motion. The taping method was designed to limit the terminal knee extension angle and rotation (especially "kneeing-in" motion). An ideal taping technique would permit other movements to be normal. We believe that controlling abnormal tibiofemoral strain is able to prevent ACL rupture.

Upper Extremity Torques during Tennis Forehand Volley

Naoko Inoue¹⁾, Youichi Iino and Takeji Kojima

Aoyama-gakuin University, Kanagawa, Japan¹⁾

The University of Tokyo, Tokyo, Japan

The purpose of this study was to determine upper extremity torques during forehand volley in tennis.

Five advanced tennis players, four male and one female, who have mean age of 34.2 years performed their forehand volley. The tennis balls were delivered by an electric ball machine with 16.0 ± 0.8 m/s ball speed. Several performances were filmed using two phase-locked 16mm cine-cameras at 200ftf film speed. Spherical markers (20 mm in diameter) made of styrofoam were attached to player's racket and some parts of body such as upper extremity, joint centers of the shoulder, elbow and wrist. One successful volley from each player was chosen as the object. The movements estimated by the marks were digitalized by a film motion analyzer. Three dimensional positional data of the extremity and the racket were determined from the digitized data using a DLT method. Then, the positional data were smoothed using a Butterworth type digital filter. Torques around the joints were determined from the smoothed data using an inverse dynamic method. The convention of coordinate system that was reported by Miyanishi et al.(1996) was used.

Results which were seen in common among the subjects during a forward swing are the followings. Torques

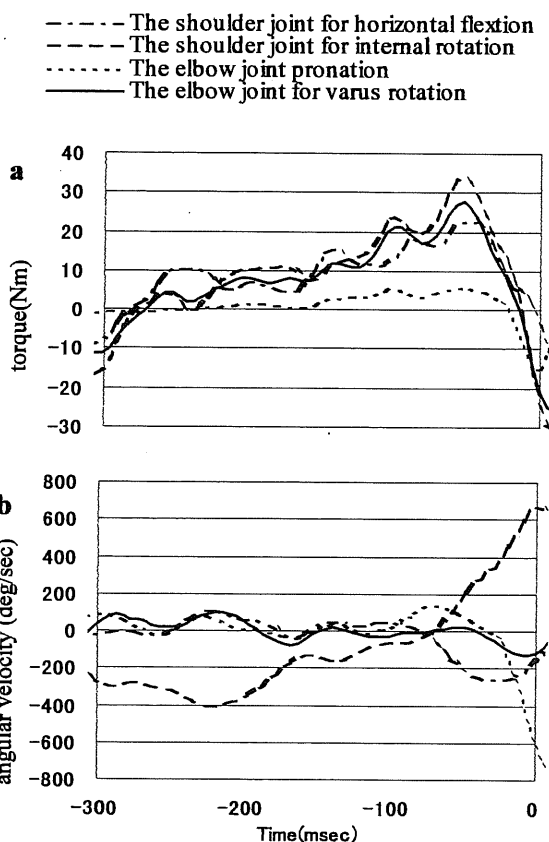


Fig.1 Torque and angular velocity

around the shoulder joint for horizontal flexion and internal rotation of the upper arm and torques around the elbow joint for varus rotation and pronation of the forearm were larger than other torques (Fig.1-a). The racket could be mainly driven by the internal rotation torque since angular velocities of the horizontal flexion and pronation were small (Fig.1-b), and the varus rotation was not allowed by anatomical constraints. The torques for the horizontal flexion and pronation served mainly to overcome inertial forces during the drive.

Reference

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A three-dimensional study on contribution of torso and throwing arm segments to ball velocity in baseball throwing motion. *Jpn. J. Phys. Educ.*, 41:23-37.

A KINETIC ANALYSIS OF LOWER EXTREMITIES DURING THE TENNIS FOREHAND STROKE

Y.Iino, N.Inoue¹⁾ and T.Kojima

The University of Tokyo, Tokyo, Japan and Aoyama-gakuin University, Kanagawa, Japan¹⁾

INTRODUCTION

The purpose of this study was to determine the joint forces and muscle moments of lower extremities and the torques around the superior-to-inferior (SI) axis of the pelvis in the tennis forehand stroke.

METHODS

Subjects were four male advanced tennis players. All subjects performed a forehand stroke three times in the closed stance with each foot on a separate force platform. Filming was conducted using two 16mm cameras at a nominal speed of 200fps. Data from the force platforms were recorded at 400Hz and synchronized with the two cameras. The 3D coordinates for necessary landmarks were determined using the DLT method and smoothed with a Butterworth digital filter. A standard link segment model was used to determine joint forces and muscle moments of the ankle, knee and hip joints. The torque around the SI axis of the pelvis, (T_{SI}), was determined from the right hip joint force in the pelvis anterior-to-posterior (AP) direction(F_{AP}^r), the right hip muscle moment in the pelvis SI direction(M_{SI}^r), the left hip joint force in the pelvis AP direction(F_{AP}^l) and the left hip muscle moment in the pelvis SI direction(M_{SI}^l) as follows.

$$T_{SI} = M_{SI}^r + M_{SI}^l + (F_{AP}^r - F_{AP}^l) \times IHC/2$$

where IHC represents the distance between the hip joint centers.

RESULTS and DISCUSSION

Results of the pelvis torque are discussed hereafter. Fig.1 shows a typical example of the torque. M_{SI}^r contributed to T_{SI} most. F_{AP}^r contributed to T_{SI} least. Knee extension and a consequent transfer of body weight forward are said to be among key elements of the forehand stroke. The results suggest that not only the transfer of body weight but also muscular actions around the right hip joint are important to rotate the pelvis around SI axis in the tennis stroke.

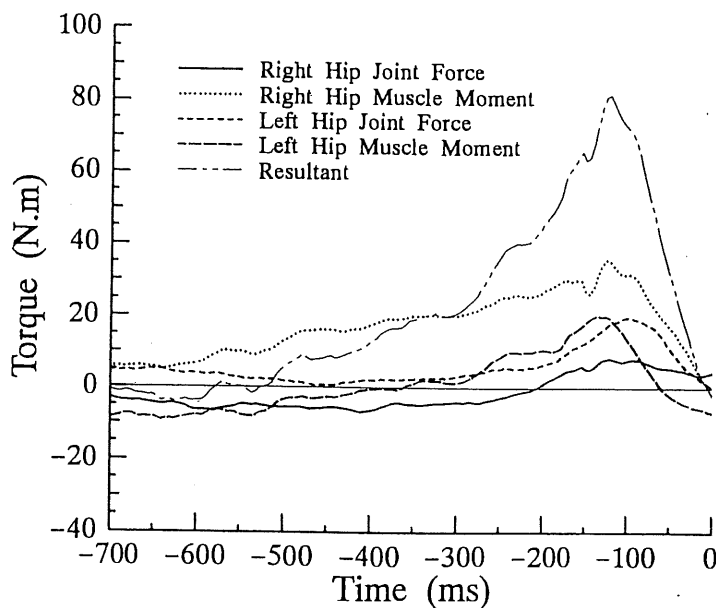


Fig.1 Resultant pelvis torque and its components

JOINT MOMENTS DURING TENNIS FOREHAND DRIVE: AN ANALYSIS OF ROTATIONAL MOVEMENTS ON A HORIZONTAL PLANE

T. Fujisawa, T. Fuchimoto, and M. Kaneko
Osaka University of Health and Sport Sciences, Osaka, Japan

INTRODUCTION

A powerful tennis drive can be achieved by the mechanism of a whip-like action (e.g., Kreambaum & Barthels, 1985)*. There is, however, few reports on tennis drive dealing with kinetic aspects such as the joint moments of the torso and upper limb movements. The present study was to examine how kinetic factors such as joint moments influence tennis ball speed after a forehand drive. Assuming that the rotational movements around the long axis of the body play an important role in executing the whip-like action mechanism during tennis stroke, the joint moments of the upper limb with torso rotation were determined two dimensionally on the horizontal plane (air-view analysis).

METHOD

The subjects were fifteen right-handed male tennis players, 20.1 ± 0.8 years of age, who were all college students. On each player 26 points were marked on the upper limb joints and the racket. In the calibrated reference area, the players were filmed at 100 fps using two 16mm movie cameras. The forehand drive was performed with every effort so that the player hit the bouncing ball toward the target area, 18.5 m² in the opposite court. The trials were repeated and that with the highest velocity for each player was selected for analysis. Using Winter's link model (Winter, 1979)**, the rotational moments were determined on the frontal plane for the joints of the right wrist elbow, shoulder and for torso rotation. The moment in the counterclockwise direction was defined as positive (+) and that in the clockwise direction as negative (-). A physical fitness test was also administered.

RESULT

According to the highest individual ball speeds, the subjects were divided into two groups; a fast group (n=8) and a slow group (n=7). Despite an apparent difference in the ball speeds between the two groups, no significant difference was found for any fitness item including the strength and power of torso rotation muscles.

During forehand drive in the fast group, the moments of torso rotation increased more than those in the upper limb joints. The wrist and elbow joint produced negative moments immediately prior to impact (Fig. 1). The integrated positive moments (IM+) during forward swing were found to be the greatest in torso rotation, followed by those in the shoulder and elbow joints. The IM+ values were significantly different between the fast and the slow groups in all movements except wrist joint moment (Fig. 2). Individual ball speeds by all subjects correlated positively with the IM+ of torso rotation and the shoulder joint. In the fast group, the ball speeds correlated negatively with IM+ values of the wrist and elbow joints, and correlated positively with IM- of the wrist joint. These results suggest that the moments of torso rotator and shoulder flexor muscles would contribute as a power generator to increase ball speed, while the wrist and elbow joints act as an energy transmitter controlling racket movement during the final stage of the tennis stroke.

* Kreambaum, E., & Barthels, K.M. (1985). Biomechanics. A qualitative approach for studying human movement. Minneapolis, Minnesota: Burgess Publishing Company.

** Winter, D.A. (1979). Biomechanics of human movement. New York: John Wiley & Sons.

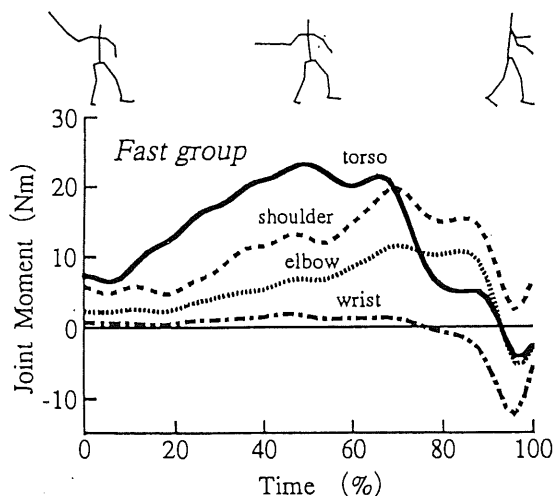


Figure 1. Joint moment changes during forehand drive in the fast group.

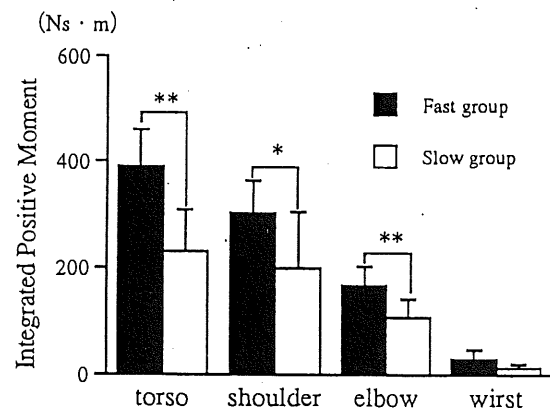


Figure 2. Integrated positive moments of the fast and slow groups. (** $p < .01$, * $p < .05$)

EVALUATION OF SMASH TECHNIQUE FROM THE VIEWPOINT OF CONSERVATION OF ANGULAR MOMENTUM

H. P. TANG, S. TOYOSHIMA

AICHI Prefectural University, Nagoya, Japan

INTRODUCTION

The purpose of this study is to evaluate the jumping smash technique of badminton from the viewpoint of Conservation of Angular Momentum. The angular momentum estimates of the whole body were calculated, and the application of the principle of conservation of angular momentum was discussed for the smash motion.

METHODS

A cinematographic technique was applied to determining the angular momentum of the human body about its mass center for general three-dimensional movements. The three orthogonal components of the angular momentum of 15 body segments composed of a transfer term and a local term were computed. The total angular momentum of the whole body was considered to be composed of the sum of the angular momentum of all body segments. The three-dimensional coordinates for determining the angular momentum were computed using Direct Linear Transformation Method from film data.

RESULTS AND CONCLUSION

According to the principle of conservation of angular momentum, the smash motion is supported by the cooperation of all the body segments. As the jumping smash movement is initiated, the lower limbs acquire some angular momentum on take-off phase from the ground (fig.1), and then transmit it to the smashing arm (fig.2). The torso and the left arm act as an intermediary in transmission of the initial angular momentum between the lower limbs and the swing arm. During the fore swing phase of the arm, a large angular momentum is generated by rapid arm rotation. The lower limbs and the head reacted upon the arm swing with an equal and opposite angular momentum to keep the angular momentum constant. This kind of counter rotation to the smash arm is useful to keep the body balance and reinforce the hitting arm. Key point for learning the jumping smash technique are to acquire an initial angular momentum on take-off phase and to make some counter rotation opposite to the swing arm.

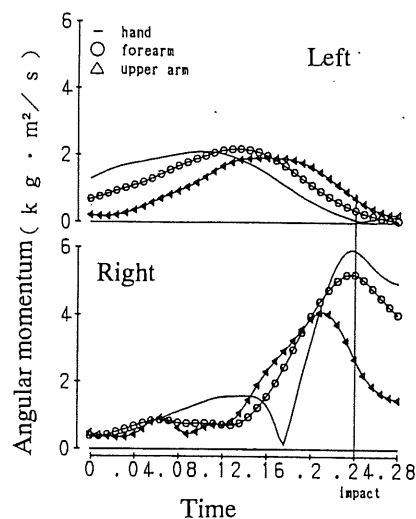


Fig. 1 Angular momentum of the hand, forearm and upper arm..

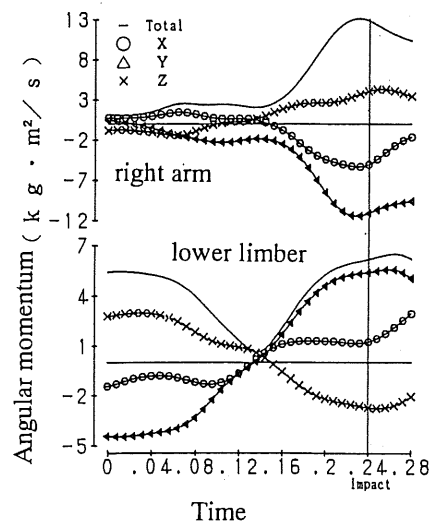


Fig.2 Angular momentum of the right arm and lower limb.

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- H. P. Tang, A method to determine the angular momentum of a human body, Japan J. Phys. Educ., Vol. 40, 1995, p. 161-69.

BIOMECHANICAL ANALYSIS OF DIFFERENT BADMINTON FOREHAND OVERHEAD STROKES OF TAIWAN ELITE PLAYERS

Chien-Lu Tsai
Chenfu Huang
Shyh-Ching Jyh

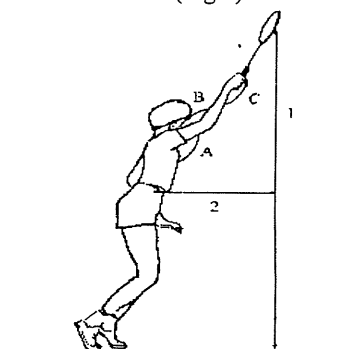
National Taiwan Ocean University, Keelung, Taiwan
National Taiwan Normal University, Taipei, Taiwan
National College of P.E. & Sports, Taoyuan, Taiwan

INTRODUCTION

Badminton is the most popular sport among the college students in Taiwan. Previous studies of badminton focused on the description of the smash and the jump smash. The purpose of this study is to compare the performance of four different forehand overhead strokes (jump smash, smash, clear and drop) by the seven Taiwan elite badminton players. We are interested in analyzing the phase of preparation position to the point of making contact with the shuttle.

METHODS

Two Peak Performance high speed video cameras operating at 120Hz were simultaneously used to record the 3-D data. A repeated one-way ANOVA and the Pearson product moment correlation were used to test the selected variables at .05 significant level. The selected variables included the shuttle speed, flying angle, contact height, the location of center of mass(CM), the angles and the angular velocities of the upper limb and the relative distance between CM and contact(Fig.1).



1. height of contact
2. horizontal distance from contact to CM
- A. shoulder B. elbow C. wrist angle

Fig.2 Relative Position of CM and Contact

RESULTS AND DISCUSSIONS

Table.1 The Comparison and the Correlation of Four Types of Badminton Strokes

Variables (unit)	Shuttle Speed (m/s)	Flying Angle (deg)	Movement Time (sec)	Contact Height (m)	Shoulder Angle (deg)	Elbow Ang V. (deg/s)	Wrist Ang V. (deg/s)
J. Smash	67.97	-13.50	0.511	2.78	164	1035	-1447
Smash	62.50	- 7.48	0.354	2.55	163	793	-1167
Clear	48.04	9.66	0.362	2.57	152	733	-1125
Drop	25.57	-2.14	0.371	2.53	150	257	-421
Sig Diff	*	*	*	*	*	*	*
r	—	*	*	*	*	*	*

* $p < .05$

The results as shown in Table 1 indicate the significant different among four types of strokes and the correlation between shuttle speed and the selected variables. The shuttle initial velocities of the smashes were from 55-70(m/s) averaging 62.50(m/s). The jump smashes were from 55-75(m/s) which has an average of 67.97(m/s). The clears were from 42-51(m/s) averaging 48.04(m/s), and the drops were from 22-29(m/s) averaging 25.57(m/s). The clear and the drop were significantly slower than the smash and the jump smash. The shuttle initial velocities of Taiwan elite players were comparable to the world class players. The following variables have significant positive correlation with the shuttle initial velocities, the movement time, the vertical displacement of C.M. , the horizontal distance between shuttle contact and C.M., the shoulder and the elbow angles at the contact, the elbow and the wrist angular velocities at the contact. The shuttle initial velocities of the smash and the jump smash were similar to the world class players. It is suggested that in order to increase the shuttle velocity, the player should lower the CM during the preparation phase and increase the angular velocities of the elbow and the wrist during the shuttle contact.

BIOMECHANICAL ANALYSIS OF CATCHING IN LACROSSE

Motoko YANASE-FUJIWARA, Eiko MARUYAMA, Aiko KIDO, Yoko YAOI
Department of Physical Education, Nara Women's University, Nara, Japan

INTRODUCTION

In this study we made biomechanical analysis of ball catching in lacrosse, particularly about how the shock to the crosse is absorbed.

METHODS

Thirteen female right-handed lacrosse players were used as subjects. Seven of them had 4 years experience (group E) and 6 were beginners (group B). The subject was asked to catch the ball of 3 different speeds (fast(F), 15.5m/s; middle (M), 13.5 m/s; slow(S), 10.8 m/s) using a crosse at a designated position. The force applied to the crosse was measured with two strain gages attached to the shaft of the crosse. The movement of the subject was recorded with a hi-speed VTR camera (SONY RSC 1150, 60fps) positioned at the subject's right side and analyzed by motion analyzer (OVAS-06).

RESULTS

Figure 1 shows a typical example of the force applied to the crosse during catching before and after the impact of the ball. Catching was divided into two phases: that is, **impact phase** which is the period of the first sharp peak of force and succeeding **moderate phase** with slow increase in force. Two factor of analysis of variance test (group $\times$ speed) was applied to three components (impulse, duration and mean force) in each phase, separately. In the impact phase, main effect for speed was significant in all components (impulse; $F(2, 33)=8.81, p<.001$, duration; $F(2, 33)=3.63, p<.05$, mean force; $F(2, 33)=14.95, p<.001$). Main effect for group was not significant in all components, nor was interaction between group and speed. In the moderate phase, the main effect for group was significant in duration ($F(1, 33)=6.66, p<.05$) and mean force ($F(1, 33)=66.62, p<.001$). The main effect for speed was found only in mean force ($F(2, 33)=39.23, p<.05$).

There was no significant difference in interaction between group and speed in all three components. The results of Scheffe's test were summarized in Table 1.

Table 1. Summary of the results of Scheffe's test

phase	impact phase			moderate phase		
component	impulse	duration	mean force	impulse	duration	mean force
group	NS	NS	NS	NS	E<B*	E>B***
speed	F>S***	F<S*	F, M>S**	NS	NS	F>S*

NS; non significant, *, $p<.05$, **, $p<.01$, ***, $p<.001$

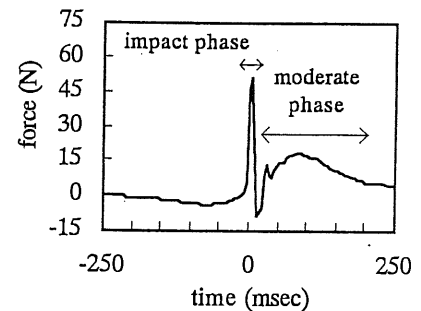


Fig.1. Typical example of force applied to the crosse. (time 0: impact)

Figure 2 shows typical stick picture of the crosse in every 50ms in group E (Fig. 2A) and group B (Fig. 2B,C). The moving pattern of the crosse pulling in group B was classified into "short pulling type" (Fig. 2B) and "over pulling type" (Fig. 2C). The subject who showed "short pulling type" caught the ball bounded from the net of the crosse. On the other hand, in the subject of "over pulling type", a ball bounded from the net repeatedly through the moderate phase.

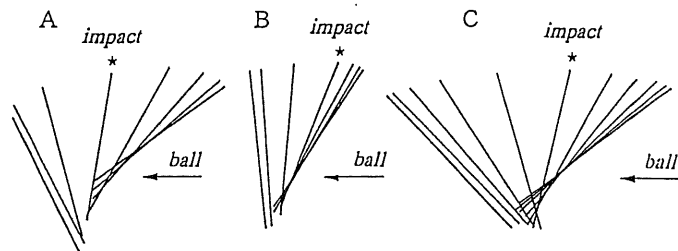


Fig.2. Typical stick picture of the crosse. (A; group E, B,C; group B)

DISCUSSION

During moderate phase, group E caught the ball with a shorter duration and larger mean force than group B. In actual ball games, ball catching should be done for a short period. The catching movement in group E is characterized by a quick and wide pulling before and after the impact and by a downward pulling of the end of the shaft after the impact. This pulling manner shown in group E may absorb the shock of the ball impact smoothly.

STICK ANGLES ON BALL CONTROL DURING THE PUSH-STROKE IN FIELD HOCKEY

A. Kuga, A. Nagata and K. Terada

Department of Human Sciences, Waseda University, Tokyo, Japan

INTRODUCTION

In field hockey, the push-stroke has been the fundamental skill and one of the most important stroke. The duration that the stick touches the ball during the push-stroke are longer than that of the hit-stroke, then the stick force is generated to three directions for the ball and would have a great influence on the ball control. Considering those directions of the stick force, the optimal stroke have been assumed the force operate to directly to the target during the push stroke. However, at the actual push-strokes, the striking planes are not always the same direction to the target. The purpose of this study was to designed the effect angles of the striking plane related to ball velocity and the direction during the push-stroke of field hockey.

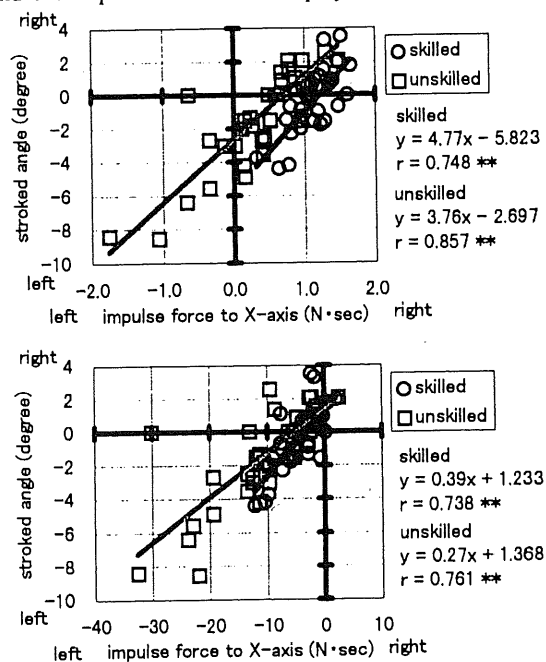
METHOD

On the artificial turf, two subjects (skilled or novice player) performed 30 trials of the push-strokes to the target at ten meters ahead with one step motion as a preliminary. All trials were recorded by two high speed video films (400 fps, shutter speed 1/1000). The 3D coordinates were reconstructed from 2D displacement data by the direct linear transformation method, and the angle displacement of the striking plane was calculated quantitatively. Additionally, the strain gauge was used to the stick, and the time-force waves of each trials were recorded continuously on the digital data recorder. The mass of the stick and the ball were 570g and 157g respectively.

RESULTS and DISCUSSION

The skilled player showed 16.65m/s of the mean ball velocity from the parallel direction to the Y-axis, and this velocity were compared with 12.53m/s of the novice player. The former velocity was much faster than one of the latter significantly ($p < 0.01$). However, the mean force impulse parallel to the Y-axis of the novice player obtained at 8.8Ns significantly greater value than value of 7.7Ns of the skilled player ($p < 0.01$). Difference between the ball velocity and the impulse force at both players calculated on the duration that the stick touches the ball. Duration of the novice player (0.102s) was significantly longer than that of the skilled player (0.059s), and therefore, the ball velocity should depend not on the impulse force but on the power.

Significant correlation between the stroked angle and the impulse force paralleled to the X-axis should be acquired with correct relationship between the stroked angle and the angle of striking plane. At the ball release, these striking angles were faced to the left direction, and also the impulse force to the right was larger than one to the left. These data would be demonstrated as following that two modified parameter values have some effects on the ball placement control.



A KINEMATIC ANALYSIS OF THE UPPER TORSO MOTION IN JAPANESE ELITE AMATEUR BASEBALL PITCHERS

K.SHIMADA¹⁾, M.AE²⁾, N.FUJII²⁾, Y.KUNUGI²⁾, M.YUKI¹⁾, T.KAWAMURA³⁾, and K.SHINOHARA⁴⁾

1) Doctoral Program in Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan.

2) Institute of Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan.

3) Hamatonbetsu High School, Esashi, Japan.

4) Master Program in Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan.

INTRODUCTION

Most of three dimensional analyses of baseball pitching motion done in recent years focus on the analysis of the throwing arm after stride foot contact with the ground (SFC). But few quantitative analysis on the upper torso motion has been done. The present study analyzed the motion of the upper torso in pitching, and tried to obtain suggestion for coaching baseball pitching.

METHODS

Pitching motions of six elite amateur baseball pitchers, included an underhand deliverer, were videotaped with two high speed video-cameras (200fps). Three dimensional coordinates of the segment endpoints were obtained by applying a DLT method.

The vector from the left shoulder to the right shoulder (LSRS) was defined. Angle of the upper torso direction (AUTD) was defined as an angle between fixed X axis and projected LSRS on horizontal plane (shown in fig. 1).

RESULTS AND DISCUSSION

Figure 2 show AUTD and the angle of the throwing shoulder external rotation (ATSER). Prior to SFC AUTD decreased while ATSER increased. This implies that the upper torso had already rotated toward pitching direction before SFC. At SFC AUTD was $82.1 \pm 7.0^\circ$. After SFC AUTD continued decreasing, and reached $-34.1 \pm 7.4^\circ$ as ball release.

While decreasing AUTD, the ball was held above the line of the shoulders (See stick pictures) regardless of the difference of delivery types and the trajectory of the throwing arm before this phase. This ball position relative to the upper torso is suitable for increasing ATSER (Feltner, 1986).

The results indicate that elite pitcher started to rotate the upper torso prior to SFC with holding the ball above the line of the shoulders so as to rotate the throwing shoulder externally. It is suggested that the motion of the upper torso and the ball position relative to the upper torso should be observed before SFC in coaching pitchers.

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Feltner, M. and Dapena, J. (1986) Dynamics of the Shoulder and Elbow Joints of the Throwing Arm During a Baseball Pitch. *Int. J. of Sport Biomech.* 2: 235-259

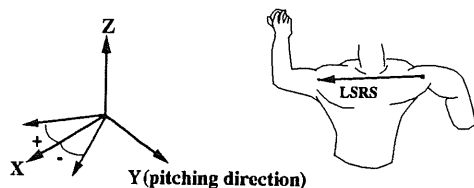


Fig.1 Definition of AUTD.

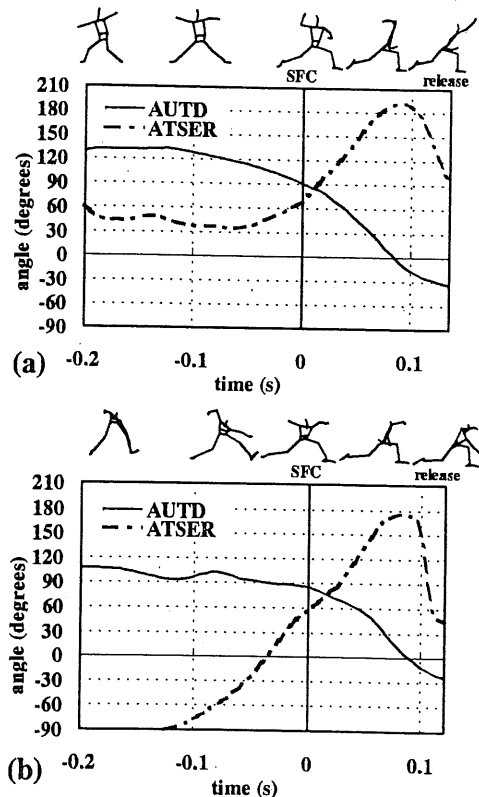


Fig.2 Changes of AUTD and ATSER. (a) overhand deliverer; (b) underhand deliverer. Decrease of AUTD means that the upper torso rotates toward pitching direction. Increase of ATSER means the throwing shoulder rotates externally. ATSER almost coincides with standard anatomic definition. Both pitchers held the ball above the shoulders' line while AUTD decreasing.

RELATIONSHIP BETWEEN ACCURACY OF PITCHED BALL AND PITCHING MOTION IN BASEBALL

K.Ishida and Y.Hirano

Department of Exercise, Sports and Health Sciences, Graduate School of Education, The University of Tokyo, Tokyo, Japan

INTRODUCTION

A baseball pitcher has to pitch not only fast but also accurately. But most biomechanical analyses have focused on the motion to throw fast ball.

In overarm throwing, because the throwing hand moves in arc, on being released the ball will move in a line which is the tangent to that arc (Hore et al. 1995). Kreighbaum and Barthels (1985) described that the more linear the throwing hand trajectory, the more time the performer has to project the ball accurately. And the rectilinear trajectory is achieved by a sequence of joint rotations including elbow extension, shoulder adduction and extension (Hore et al. 1995).

The purpose of this study was to measure the accuracy of pitched ball and to examine whether the pitching motion, especially the hand movement, affected its accuracy.

METHODS

Thirteen collegiate baseball players were asked to pitch 100 balls as accurately and fast as possible, aiming at the center of the target 18m away. Each hit point of the 100 pitched ball was recorded by using the xy-coordinates on the target, and the accuracy of the pitched ball of each player was defined as the standard deviation of the x and y value (SD_H , SD_V).

Their pitching motions were analyzed by using the direct linear transformation (DLT) method of three-dimensional cinematography (200 frames/s). Eight points were digitized per frame; hip, shoulder, elbow, wrist, hand of throwing arm side, hip, shoulder of opposite side and the ball. To smooth these position data the second order Butterworth-type low-pass filter was used (cut-off frequency was 15 Hz) (Winter 1979). The hand trajectory was estimated by the mean curvature of its projection on the horizontal plane and the sagittal plane during the acceleration phase (CH , Cv). Adduction/abduction, horizontal adduction/abduction, internal/external rotation at the shoulder joint and flexion/extension at the elbow joint were calculated. To examine a player's throwing pattern (overarm throw or sidearm throw) the angle formed by upper arm and vertical line at ball release (θ_{UA}) were calculated.

RESULTS AND DISCUSSION

Significant correlation was found between SD_V and Cv (Figure 1). Also correlations between θ_{UA} and CH as well as Cv were found significant. This indicated that an overarm throw player showed large Cv and SD_V . However, differences in Cv were shown among the players who had similar θ_{UA} . Curvilinear hand trajectory (large Cv) was possibly caused by the decrease of elbow horizontal velocity prior to ball release and wide range of forearm movement during acceleration phase. The potential cause of the former was that the trunk had rotated excessively prior to ball release, and that of the latter was that the elbow angle at the maximum external rotation of the shoulder was small.

Table 1 Maximum and minimum values for the variables.

	SD_H (cm)	SD_V (cm)	CH (m^{-1})	Cv (m^{-1})	θ_{UA} (deg.)
max.	31	47	1.59	1.57	94.3
min.	20	22	0.54	0.43	43.8

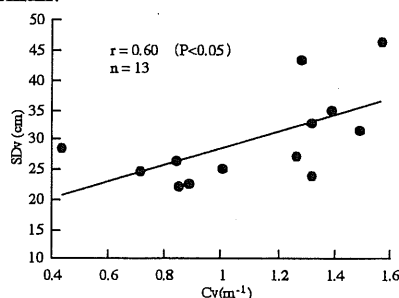


Figure 1 Relationship between Cv and SD_V .

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BIOMECHANICAL ANALYSIS ON TENPIN BOWLING TECHNIQUES

Youlian Hong and Li Weiping*

Dept.of Sports Science & Physical Education, The Chinese University of Hong Kong

*Shandong Sports Science Center, Jinan, Shandong, China

INTRODUCTION

The purpose of this study was to improve the individual technique of the Hong Kong elite tenpin bowling athletes. Due to the lack of biomechanical research papers in the international publications, this study established database for further investigation. To attain this purpose, a generalized biomechanical model of playing tenpin bowling was developed based on the kinematic data collected using video cameras from the elite athletes of Hong Kong.

METHOD

The approach and delivery movement of 11 Hong Kong Tenpin Bowling Team athletes were filmed in side and back directions during their training session. The taped materials were selected, digitized and calculated with a Peak Motion Measurement System. Only 4-step approach was shot due to limitation of view size of our cameras.

RESULTS AND DISCUSSION

Table 1. The kinematic parameters of the subject's movement

Kinematic parameter (M±SD)		Male	Female
Velocities of body mass center (m/s)	Step 1	0.81 (0.33)	0.97 (0.23)
	Step 2	1.40 (0.15)	1.54 (0.10)
	Step 3	1.95 (0.14)	1.90 (0.26)
	Step 4		
	-D ₁ (Pre-sliding distance)	2.43 (0.27)	2.55 (0.26)
	-D ₂ (Sliding distance)	2.78 (0.20)	2.89 (0.34)
Release velocity		8.00 (0.56)	7.70 (0.29)
Average sliding distance range of body mass center (cm)		29~66	27~61
Average release angle of the left knee (degree)		123	124
Average distance range of ball climax to the right shoulder (cm)		-9~40	1~29

Regarding the displacement and speed characteristics of the athletes' body mass center during the phases of approach and delivery, it revealed that the (1) most male bowlers adopted the "long by long" style, and all female employed the "short-long-short-long" approach style; (2) a long 1st-step led to a short sliding distance; (3) Most athletes accelerated during approach. The analysis found that the acceleration of the athletes were maintained evenly during their approach. But bowlers reduce their speed during the fourth step. This indicates the rhythmic sequence of approach was out of controlling. The sliding velocity of some bowlers was high but the ball released velocity was relative small. This shows that the continuity of motion from forward swing of ball to ball delivery was not perfect good. It is found that the left knee angles of male and female were high. However, if the angles could be smaller, the risk of injury can be reduced.

CONCLUSIONS

Some important conclusions and suggestions can be given as follows: (1) the foundation skill should be strengthened. "Four stride" technique should act as the model for unique Hong Kong style bowling; (2) the rhythm of approach should be paid attention in training and technique; (3) Strengthen the muscle power of lower extremities especially the ankle to prevent sports injury.

A BIOMECHANICAL COMPARISON OF THE SWING TECHNIQUES OF MEN AND WOMEN GOLFERS

J. Park

Seoul Women's University, Seoul, Korea

INTRODUCTION

Many golf professionals and teachers have thought that the women's swing is merely a slower version of the men's swing. However, other golf authorities (Flick 1984, Stockton 1992) have proposed that there are gender differences in swing mechanics. The purpose of this study was to compare the segmental actions to gender related physical differences during the swing.

METHODS

Ten men and ten women golfers with similar levels of skill (less than 10 handicaps) were chosen for the study. The subjects performed the swing with a number one club. They were videotaped from the side (sagittal view) and front (frontal view). Two NAC HSV-400 high speed cameras with a NAC high speed recording system were used to video tape the subjects at a rate of 200 fps on standard VHS tapes. The nine physical variables and the fourteen biomechanical variables were collected from all subjects.

RESULTS

The men and women golfers differed significantly ($p < 0.10$) in all physical variables except arm length. At the start of back swing, the women golfers had a significantly ($p < 0.10$) more upright swing plane than the men golfers. The men golfers were significantly ($p < 0.10$) faster at the swing time and the peak linear velocities than the women golfers. The men golfers also had a significantly greater width of stance than the women golfers. The women golfers, on the other hand, moved their hip joint significantly ($p < 0.10$) farther than the men golfers during the swing. No significant ($p > 0.10$) differences were found between the groups for the shoulder rotation and shoulder joint horizontal displacement.

DISCUSSION

The women golfers have a different swing style than the men golfers. The women golfers should use their lower body more in order to hit a ball well and far. The men golfers, on the other hand, should move their body quickly with the wide stance which provides the stability to swing.

KINEMATICAL ANALYSIS OF FOOT FORCE DURING THE GOLF SWING

K. KAWASHIMA

Laboratory of Biomechanics, College of Bioresource Science, Nihon University
1866 Kameino, Fujisawa-City, Kanagawa-Ken, Japan

Introduction: This study has the purpose to investigate sole pressure recorded from both feet of two groups during the golf swing. 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 sole of both feet. Foot pressure of the golf swing have been used by many golf players, who trained those patterns of the golf swing. Foot force could be appeared with subjective observation alone during the golf swing. Then this study has the purpose to investigate the feet forces of six area of both feet during the golf swing. It was became obvious that there was a difference between skilled golfers and unskilled golfers in the swing movement by compared the pressure the sole of the feet. But it is almost never made a study of foot movement during the golf swing. And the foot pressure movement performance is done unconsciously although the change of foot has chance to be the cause of wrong swing. Therefore, the purpose of this study was to develop a method for improving performance by motion of applied pressure.

Methods: Seven skilled golfers (Handicaps= 5.5 ± 1.8) and seven unskilled golfers (Handicaps= 28.0 ± 4.0) volunteered to be the subjects in this study. Subjects were all right handed person, and all of them used 5 iron shot. The force of six points of both feet were measured quantitatively with the seven small sensors. It was carried out using the small load cell on the insole. Calibration up to 100 N was performed individually for each sensor. Also, a VTR camera was used to analyze the phase of address, impact, and follow through. While impact was recorded with using a laser sensor (1ms). After these data were analyzed comparatively, following results were acquired statistically. The aim of this study was compare the foot pressures recorded for two groups of golf swings of different performance levels, in order to establish differences between skill levels for several foot pressure.

Results and Discussion: Maximal foot pressure values of phase, and while comparing skilled golfers and unskilled golfers. The results of this study would indicate the following conclusion maximal L.minimi area, showed that the intensity of pressure in skilled golfers ($68.57 \pm 31.36\text{N}$) than unskilled golfers ($31.14 \pm 5.87\text{N}$) ($p < .05$) during the swing. Peak force of the R.hallucal area was recorded $56.0 \pm 25.0\text{ N}$ for the skilled golfers, and $21.4 \pm 6.6\text{ N}$ for the unskilled golfers at impact. Force of the R.minimi digiti of skilled golfers was $43.7 \pm 21.8\text{ N}$, and those of unskilled golfers were $21.7 \pm 14.0\text{ N}$ at prior impact ($p < 0.05$). While the force of the both feet pressure of the skilled golfers were increased higher than that of the unskilled golfers. Both forces of the L.hallucal area L.heel area of the left foot did not get difference between skilled golfers and unskilled golfers. However, the maximal force of the right foot were appeared about 120-230 ms prior at impact. Then, formation of the golf swing pattern would be suggested to be important to make without drawing out four forces of the R,L minimi digiti, R.heel area, and R.hallucal area. It was found that there was difference between skilled golfers and unskilled golfers in the point of pressure pattern of foot, which are related to foot movement in swing time ($p < .05$). According to this study, it was recognized that the useful for golf swing, it is recommended that the feet are received the heavy pressure at impact.

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WAVE MOTION OF BUTTERFLY DOLPHIN KICK

S.Shimonagata and M.Taguchi and S.Taba

Department of Physical Education,Fukuoka University,Fukuoka,Japan

INTRODUCTION

Butterfly swimmers establish the kick underwater with a very quick and shallow dolphin's kick while streamlining the position after starting.The dolphin's kick is very important for all levels of butterfly swimmers. When doing a dolphin kick a swimmer's body parts are oscillated vertically throughout the stroke cycle.The oscillations transfer caudally along the body.The body oscillations have been similar to wave-like motions. Although the vertical oscillations of body parts have been identified,there has been little study to investigate the magnitude of these oscillations and the temporal relationships of body parts.

The purpose of this study was to estimate the feature of wave motions during a dolphin's kick with oscillation's and electromyographic analysis.

METHODS

The subjects were seven skilled and two unskilled butterfly swimmers.Each subjects swam dolphin kicks underwater with three subjective speeds(Fast,Middle,Slow) .Swimming motions were recorded by 60Hz video camera below the water.Electromyogram(EMG) was measured from seven muscles with surface electrodes.For synchronizing video data and EMG data,the pressure sensor fixed on the instep.EMG and pressure data was recorded on multitelemeter system by 500Hz.

RESULT

Wave motion.Swimming velocity ranged from 1.06m/sec to 1.65m/sec for skilled swimmers,and for unskilled swimmers,from 0.53m/sec to 0.70m/sec.Continuous displacement-time curves of the shoulder,hip,knee and ankle of the skilled subject is showed in Figure 1.This indicated a wave motion traveled from hand to ankle along the body.This wave motion was similar to the unskilled subjects.Phase angles of hand-ankle were 364.7° for skilled swimmers,and 345.8° for unskilled swimmers.Hand-shoulder phase angle was 89.1° for skilled swimmers and 4.92° for unskilled.The correlation of hand-shoulder phase angles between skilled swimmers and unskilled swimmers were significant difference($p<0.01$).Similarly shoulder-hip phase angles(skilled swimmers= 125.1° ;unskilled swimmers= 163.9°) were significant difference($p<0.01$).

EMG.The flexed and extended muscles activity of skilled swimmers indicated a coordinated wave along the body.The electromyographic patterns of antagonists of skilled swimmers alternated in time, however, that of unskilled swimmers showed the co-contraction(Figure 2).

DISCUSSION

In this study, skilled and unskilled swimmers' oscillation and EMG during swimming dolphin kick was investigated.The fact that dolphin kick of skilled swimmers traveled from hand to ankle coordinaly was indicated from phase angle of each body segments and muscle activity patterns of antagonists.These results showed truly harmonic or effective dolphin kick.On the other hand, phase angle of unskilled swimmers were few angles of hand-shoulder,too large angles of shoulder-hip.This consequence was caused by the law of conservation of momentum.Because these facts indicated that unskilled swimmers didn't transfer oscillation from hand to hip,swimming motion of unskilled swimmers were constrained.

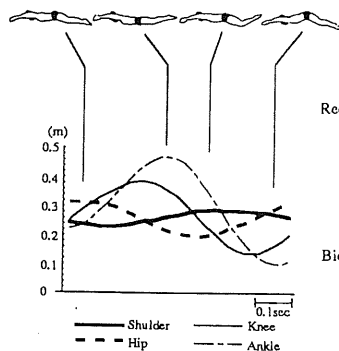


Figure 1. Continuous displacement-time curves

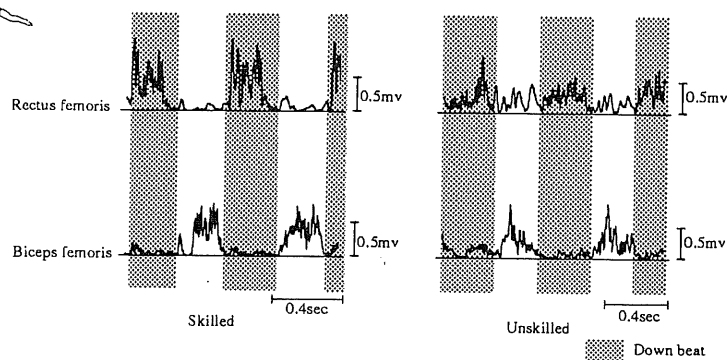


Figure 2. EMG of Muscles of the Thigh

SAFE TECHNIQUES FOR THE STARTING MOTION IN YOUNG SWIMMERS

T.Nomura *, Y.Mutoh †, M.F.Ohta †, H.Kamioka † and K.Mochizuki ‡

* Kyoto Institute of Technology, Kyoto, Japan

† The University of Tokyo, Tokyo, Japan

‡ Hongo Godo Law Office in Tokyo, Tokyo, Japan

INTRODUCTION

In swimming, starting from the platform carries a risk of collision against bottom of the pool. Among cervical spine injuries, diving injuries accounted for 60 to 80% of sports injuries (Mutoh and Ohta, 1992). Diving accidents are related to various factors such as physical condition, starting techniques, pool structure, and instructional management. This study focused on water depth and starting techniques, comparing the starting motions between at 1.5m pool depth and 0.8m pool depth.

METHODS

An original pseudo bottom made of polyvinyl chloride net was set in a swimming pool (1.5m depth). The pseudo bottom was floated at 0.8m from the water surface to create a safe shallow pool. Water depth of 1.5m was considered the normal situation, while a water depth of 0.8m was considered a shallow situation. Two video cameras were set to record the right side view from land and underwater. To compare the two situations, 18 elementary school students starting from a starting platform (0.4m height) were filmed. Video images were converted to digital images using a PC. Then, 2-D analysis was performed.

RESULTS

Hand entry occurred at a horizontal distance from the front edge of the platform was 2.40 ± 0.35 m in the normal situation and 2.40 ± 0.30 m in the shallow situation. There was no significant difference between these values. However, the deepest point of the head in the normal situation, 0.70 ± 0.12 m, was deeper than that in the shallow situation, 0.55 ± 0.10 m ($p < 0.01$). Furthermore, the deepest point of the foot in the normal situation, 0.82 ± 0.15 m, was deeper than that in the shallow situation, 0.71 ± 0.14 m ($p < 0.05$).

The table shows a comparison between the normal situation and the shallow situation. The tilting angles are indicating forward tilting. When a part of body tilted below the horizontal line, the tilting angle is shown as a minus value.

Table The tilting angles of trunk and arm in the vicinity of the entry phase for the starting motion.

Parts of body		Trunk			Arm		
Water depth		1.5m	0.8m	Signif. level	1.5m	0.8m	Signif. level
Phase							
Just before entry	M	-24.39	-24.25	NS	-42.99	-42.36	NS
	SD	9.21	5.50		8.39	8.28	
During entry	M	-24.63	-24.20	NS	-19.66	-8.26	**
	SD	9.96	7.11		10.12	7.01	
Just after entry	M	-10.67	-5.84	*	-8.55	1.62	**
	SD	8.83	9.07		6.45	6.87	
Deepest point of the head	M	-5.31	-0.98	*	0.23	4.63	*
	SD	8.84	4.35		6.49	6.16	

Unit; degree

Significant level; * $p < 0.05$, ** $p < 0.01$

Minus value means forward tilting.

DISCUSSION

In comparing the normal situation and the shallow situation, the flight distance and the entry angle were similar. However, the deepest point of the head in the normal situation was deeper than that in the shallow situation. Hence, the starting motion contains a control function for the vertical distance after the hand touches at the water surface.

During entry, the arm forward tilting angle in the shallow situation was drastically reduced. Just after entry, drastic reduce of the trunk forward tilting angle in the shallow situation followed the arm. Accordingly, it seems that the arm movement directs the underwater movement.

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EFFECTS OF THE FOOT FRONTAL AREA AND THE LOWER LIMB'S FLEXIBILITY ON THE MAXIMAL SWIMMING VELOCITY DURING LEG KICK AND WHOLE BODY STROKE OF FRONT CRAWL

H. Fujiwara and F. Ogita

National Institute of Fitness and Sports, Kanoya, Japan

INTRODUCTION

It has been considered that the foot frontal area is important for the ability to swim faster in leg kick and whole body swimming because an enlarged propulsive surface area induces an increase of the propelling efficiency. In addition, it has been generally thought that the flexibility of the lower limb is also a significant factor to produce a higher swimming speed during leg kick and whole stroke. However, there has been no current research which actually examined the interrelationships between these factors and the maximal swimming velocity. Therefore, the purpose of this study was to determine the foot frontal area and the lower limb's flexibility, and to examine the relationships between these physical factors and the maximal swimming velocity during leg kick and whole body stroke of front crawl.

METHODS

The subjects were 18 trained college male swimmers. Their mean age, height, and body mass were 21 ± 1 years old, 172.4 ± 5.7 cm, and 68.2 ± 5.9 kg, respectively. All experimental measurements were conducted in a short course swimming pool. The subjects swam 25.0 m three times in leg kick (L) and whole stroke (S) of front crawl swimming with a maximal effort. The stroke rate, the distance per stroke, and the swimming velocity as the stroke indices were determined by analyzing a videotape which was recorded at 60 fps. The data between 12.5 and 20.0 m were used for a video analysis and the values obtained from three trials were averaged. The right foot frontal area and the maximal joint angle for an active extension of the right knee and the right ankle were determined by using a squared paper and by a protractor, respectively.

RESULTS

The stroke rate, the distance per stroke, and the swimming velocity during L and S were 0.79 ± 0.04 and 0.92 ± 0.07 strokes $\cdot s^{-1}$, 1.39 ± 0.14 and 1.85 ± 0.11 m $\cdot stroke^{-1}$, and 1.10 ± 0.10 and 1.71 ± 0.10 m $\cdot s^{-1}$, respectively. A significant correlation was found in the swimming velocities between leg kick and whole body swimming. The foot frontal area significantly correlated with the distance per stroke and the swimming velocity for both L and S. On the other hand, the joint angle in the knee and the ankle extension had no significant correlations with the stroke rate, the distance per stroke, and the swimming velocity in L and S.

DISCUSSION

A significant positive correlation was found in the maximal swimming velocity between L and S. This suggests that the ability to swim faster in L would be a determinant of the maximal swimming speed in S. In both L and S, the foot frontal area significantly correlated to the distance per stroke and the swimming velocity. It can be expected that a larger foot frontal area (i.e. a larger propulsive surface area) would induce a higher propelling efficiency because a swimmer can push off against a larger mass of water during one stroke. Consequently, the higher propelling efficiency would be considered to induce a longer distance per stroke and might contribute to producing a higher swimming speed. On the other hand, the joint angle for the knee and the ankle extension had no relationships with any stroke indices in L and S. This result suggests that although it has been generally considered that the flexibility of the lower limb is an important factor to produce a faster swimming speed, it is quite questionable whether the flexibility of the lower limb can be an important determinant of the maximal swimming velocity for L and S, especially for a well trained swimmers. In conclusion, our results suggest that a larger foot frontal surface area which would induce a higher propelling efficiency is more important for a determinant of the maximal swimming velocity during L and S, when compared to the flexibility of the lower limb. However, since it is usually difficult to enlarge the foot frontal area by a training or by any voluntary effort, this may rather contribute as a helpful information to assess one of the talents or potential abilities of a swimmer.

THE INFLUENCE OF PADDLE GRIP POINT ON THE KINEMATICAL VARIABLES DURING STROKING IN CANADIAN CANOE

Ji-Seon Ryu, Yeon-Jong Lee & Chang-Kyu Park

Korean National University of Physical Education, Seoul, Korea

INTRODUCTION

The purpose of this study is to investigate the effects of paddle grip point on the kinematical variables during stroking in Canadian canoe. Though the effects of two paddle types on performance in North American Canoe racing (Nolan & Bates, 1982) and the world's best flatwater paddlers were analyzed (Plagenhoef, 1979), there are few quantitative analysis on the Canadian stroking.

METHODS

Six male Canadian Paddlers who have nearly the same reach span were used as subjects. The kinematical variables of the stroking limb, blade and boat from catch to finish phase were analyzed for three different grip patterns (grip A: 5cm above the center of paddle, grip B: center of paddle, grip C: 5cm below the center of paddle) through use of a three-dimensional film analysis technique. The length of paddle used in this study was 165.5cm, mass was 900g, and the length of blade was 58cm. The DLT method was used in the computation of three-dimensional coordinates of the digitized body points and boat. The digitized data were filtered at a cutoff frequency of 6 Hz using a fourth order low-pass Butterworth filter before computation of kinematical values.

RESULTS AND DISCUSSION

There were no significant differences in the displacement of the elbow, wrist joint and blade, but there were significant differences ($p < 0.05$) in that of the shoulder joint (type A: 2.10m, B: 2.42m, C: 2.66m) and boat (type A: 3.10m, B: 3.44m, C: 3.14m) among three grip patterns. In addition, there were significant differences ($p < 0.05$) in the velocity of the shoulder joint (type A: 3.79m/s, B: 3.80m/s, C: 4.14m/s) and boat (type A: 4.16m/s, B: 4.35m/s, C: 4.23m/s) among three different grip points. In conclusion, the different grip point during Canadian stroking had effects on the shoulder joint's kinematical variables and the boat of displacement. Grasping the center of a paddle during stroking in a singular Canadian was therefore confirmed as the best grip pattern from kinematical point of view.

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STATISTICAL ESTIMATION OF FORCE-VELOCITY RELATIONSHIP DURING DYNAMIC MUSCLE CONTRACTION

K. Itoh¹, K. Mita¹, K. Akataki¹ and A. Katoh²

¹ Institute for Developmental Research, Aichi Human Service Center, Kasugai, Japan

² Aichi Institute of Technology, Toyota, Japan

INTRODUCTION

Hill (1938) reported the relationship between force and velocity of shortening in frog muscle was represented in a rectangular hyperbola. Kaneko (1970) confirmed the hyperbolic relation in human elbow joint flexion. Furthermore, he showed an estimation method of the Hill's characteristic equation by transforming it to a linear form under the condition that the maximal isometric force was known, then by determining two parameters associated with heat and energy liberation with the regression analysis. The aim of the present study is to propose an advanced method for estimating the Hill's characteristic equation even if the maximal isometric force is unknown.

THEORETICAL DESCRIPTION

We denote a pair of velocity and force on the Hill's characteristic equation by v and P .

$$(P+a)(v+b)=b(P_0+a) \quad (1)$$

(a, b : constant, P_0 : maximal isometric force)

The above equation is changed into Eq.(2).

$$P=b\{(P_0-P)/v\}-a \quad (2)$$

It represents a straight line with the slope b and intercept $-a$ in the coordinate system $\{(P_0-P)/v, P\}$.

Consider that a dynamic force P' measured at a velocity v includes an error ΔP , where the expectation of ΔP is zero ($E[\Delta P]=0$). Furthermore, we denote the maximal isometric force estimated from a data set (v, P') by P'_0 . It has a difference ΔP_0 from the true value P_0 . The Hill's characteristic equation associated with P'_0 and (v, P') is expressed by

$$(P'+a)(v+b)=b(P'_0+a). \quad (3)$$

A similar equation to Eq.(2) is obtained by rearranging Eq.(3) as follows:

$$P'=b\{(P'_0-P')/v\}-a \quad (4)$$

or

$$P=b\{(P_0-P)/v\}-a+(b/v)\Delta P_0-(b/v+1)\Delta P. \quad (5)$$

After expectation operation, the final expression is derived.

$$P=b\{(P_0-P)/v\}-a+(b/v)\Delta P_0. \quad (6)$$

This indicates if the maximal isometric force can be estimated correctly with measured data ($P'_0=P_0$), the linear relation same to Eq.(2) is obtained. However in the case of $P'_0 \neq P_0$, this relation is curved due to the third term of Eq.(6).

If we calculate the correlation coefficient as a function of P'_0 in $\{(P'_0-P)/v, P\}$ coordinate system, the greatest value is determined at $P'_0=P_0$. Thus P'_0 associated with the greatest correlation coefficient will lead to an optimal estimation of the Hill's characteristic equation. In this sense, the method is referred to as the Maximal Correlation Method (MCM). In addition, the non-linear optimization (NLO) with the parameter values resulted from the MCM as an initial condition is applied for further improvement of the accuracy.

SIMULATION

Statistical simulation with Gaussian random number was performed for testing validity of the MCM. The standard deviation of Gaussian population was 3% of P_0 . Total number of samples was 10,000 and they contributed to 100 time trials with 100 data points. Three parameters a, b and P_0 were assumed to be 74N·m, 5.2rad/s and 245N·m, respectively. The velocity ranged from 0 to 8.7rad/s. Validity of this method was evaluated from the estimated values of these parameters.

RESULTS & DISCUSSION

The simulation results only by the MCM showed the estimation error of $119 \pm 42\%$ in a , $51 \pm 18\%$ in b and $3 \pm 3\%$ in P_0 . The combined method of the MCM and NLO improved the accuracy, i.e., $11 \pm 8\%$ in a , $7 \pm 5\%$ in b and $1 \pm 1\%$ in P_0 . Thus the use of only the MCM seems to be insufficient in determination of the constant parameters a and b , and the combined method with the NLO utilizing the results by the MCM as initial value may be better way for estimating the Hill's characteristic equation. In conclusion, the combined method appears to be a useful tool in investigating human dynamic movement.

EXPLANATION FOR SYSTEMATIC ERRORS IN DETERMINING THE POINT OF FORCE APPLICATION WITH FORCE PLATES

H.-B. Schmiedmayer¹ and J. Kastner²

¹Institut für Mechanik, Technische Universität Wien, Wien, Austria

²Forschungsinstitut für Orthopädiotechnik, Wien, Austria

INTRODUCTION

Force plates are broadly used measurement devices in biomechanics. Systematic errors in determining the point of force application have been reported in the literature (Bobbert 1990). In this paper an analytical solution for a 2D-model is given to show the reasons for these errors. Also the specific design parameters that influence these errors will be discussed.

METHOD

For 2D-investigations the force plate can be modelled by three beams, one representing the plate itself with a bending stiffness of EJ_1 and two vertical beams including the force transducers with a bending stiffness of EJ_2 (Fig. 1). The force plate is a statically overdetermined system, and therefore deformations must be taken into account to calculate the supporting forces and moments. To get a dimensionless description of the error and special design parameters the following variables are used to describe the system: ξ normalized distance of the point of force application; $\zeta = h/a$ normalized distance of the plane of measurement (i.e. the plane where the force components are measured by the piezoelectric transducers) to the point of force application; $\zeta_0 = h_0/a$ normalized height of the force transducer beams; $\varepsilon = (EJ_1)/(EJ_2)$ normalized stiffness. All deformations were supposed to be small and only the bending modes (no shear deformation) of the beams were taken into account.

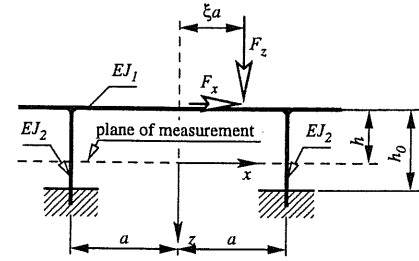


Figure 1: 2D-model for the force plate using beam elements of different stiffness EJ_i .

RESULTS

The formulas for the calculation of the point of force application

$$\hat{\xi} = \frac{B_{2z} - B_{1z}}{F_z} - \zeta \frac{F_x}{F_z} \quad \text{with} \quad F_x = B_{1x} + B_{2x} \quad \text{and} \quad F_z = B_{1z} + B_{2z} \quad (1)$$

given by Kistler (Kistler 1984) are based on the assumption that in the plane of measurement no moments are transmitted through the force transducers. This is not true for the current setup of the force platform. Based on the above model the supporting forces and moments can be calculated. Substituting the corresponding forces in equation (1) leads to the following expression describing the error for the point of force application:

$$\hat{\xi} - \xi = \frac{(1 - \xi^2) \xi}{2(3 \cdot \varepsilon \cdot \zeta_0 + 1)} + \left(\frac{3 \cdot \varepsilon \cdot \zeta_0}{2(3 \cdot \varepsilon \cdot \zeta_0 + 1)} - \zeta \right) \frac{F_x}{F_z} \quad (2)$$

DISCUSSION

The first term in equation (2) shows, that the error is a polynomial function of the position of the point of force application and is of the type $p(\xi) = K(1 - \xi^2)\xi$. It has zeros at the middle of the force plate ($\xi = 0$) and directly over the force transducers ($\xi = \pm 1$) (Fig. 2). The scaling factor K in this function depends mainly on the normalized stiffness ε . The error can be minimized by $\lim_{\varepsilon \rightarrow \infty} K = 0$ i.e. by making the plate much more stiffer than the measurement beams. The scaling factor becomes a maximum for $\varepsilon \rightarrow 0$ with $K = 0.5$.

The second term in equation (2) shows the influence of the crosstalk between F_x and F_z which can be minimized by selecting an appropriate distance of the plane of measurement with $\zeta_{\text{opt}} = 3 \cdot \varepsilon \cdot \zeta_0 / (6 \cdot \varepsilon \cdot \zeta_0 + 2)$ being independent of ξ and the applied forces.

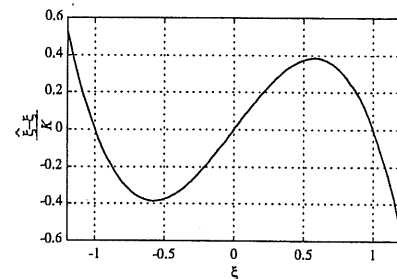


Figure 2: Normalized error $\frac{\hat{\xi} - \xi}{K}$ as a function of the normalized position ξ .

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COMPUTER INVESTIGATION OF INFLUENCE OF GEOMETRICAL AND MECHANICAL PROPERTIES OF JOINTS CONTACT SURFACES ON DYNAMICS OF HUMAN MOTION

A.A. Ivanov, V.A. Sholukha, A.V. Zinkovsky
St. Petersburg State Technical University, Russia

INTRODUCTION

The problem of the modelling of the human motion can be solved through a) description and prediction of the motion kinematics, b) determination of the internal forces in joints, bones, and muscles or c) determination of the external forces arising from the interaction with the environment during slow and fast motions. Dynamical model is most universal model which can help solve any of these tasks. However, there is needed detailed research on what influence have joints contact surfaces properties on motion modelling.

METHODS

The dynamical model of skeletal muscular apparatus developed by the authors is used within the frame of a universal anthropomorphic model (Zinkovsky 1995, Zinkovsky 1996) with the traditional model approach of a rigid body system with arbitrary connections. This traditional approach suggests joint modelling by one or two degrees of freedom cylindrical joints with fixed rotation axes. In the considered model we employ joints with a pre-set contact surface form. Detailed description of anthropometrical parameters is used.

RESULTS

We investigate influence of complicated joint contact surface form on the kinematics and dynamics of some typical motions of the leg, foot, arm, hand. Anthropomorphic data on positions and functions of muscles, tendons and some forms of joint contact surfaces were taken into account. Location of instant axes of rotation depends on the parameters of the contact surface and on the control action of the muscles. The influence of tribology property of joint is investigated too.

DISCUSSION

The developed model permits to simulate the limbs motion and estimate involved forces values and kinematic characteristics. Equally the model allows to investigate its sensitivity to anthropometric characteristics, such as geometrical and mass-inertial parameters as well as to the form and friction parameters of joint contact surface. After parametrical adjustment via detailed experiments, the model will predict the dynamics, kinematics and statics of the whole limb as well as its separate parts.

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COMPUTER SIMULATION ANALYSIS OF THE EFFECT OF A BALL'S MASS AND SHAPE TO ONE'S UPPER LIMB FROM THROWING

Yoshiyuki MOCHIZUKI*, Haruo AMANO*, Tsuyoshi MATSUMOTO**, Kazushi TEZUKA***, Shinichi YAMASHITA****, and Koichi OMURA*****

* Laboratories of Image Information Science and Technology, Osaka, Japan

** University of Tsukuba, Ibaraki, Japan

*** Beta-endorphin, Tokyo, Japan

**** Triaxis Co.,Ltd., Kanagawa, Japan

***** Takarazuka University of Art and Design, Hyougo, Japan

INTRODUCTION

Though it is easily assumed that a throwing motion is affected by a ball's mass and shape, it is very difficult to analyze the effects of the examination when we use an actual human. This paper serves as an analysis method for the effect of optimizing motion simulation for one's upper limb on the computer, and indicates some results of the simulation for several ball types.

METHODS

When we use the same initial motion and man's physical data, throwing motions generated by the simulation for some ball types are different from each other. These differences are reflected by the effect that is caused by differences from ball types. Our simulation method is based on this thinking and has following features: (1) a motion captured data as an initial motion, (2) a 3D-mathematical model of the upper limb and a ball described by Lagrange equations, (3) the quasi-Newton method with the BFGS formula and the secant method with Wolfe conditions, and (4) the objective function consists of terms for total torque quantity, total torque-change quantity, and penalties about the range of joint movement, the range of joint torque, and ball velocity (35 m/s). Table-1 indicates conditions of different balls using our simulation.

RESULTS

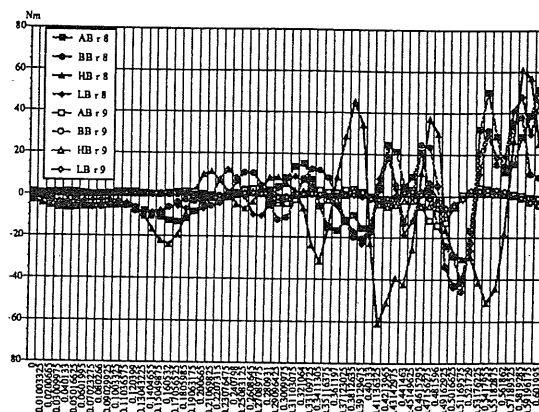
Obtained data from the simulation are time sequence data of each joint angle, torque, and velocity of each part of the body. For example, Graph-1 and Graph-2 shows time sequence data of elbow joint torque and velocity in the condition (AB), where No.8 corresponds to elbow flexion-extension, No.9 corresponds to pronation-supination.

DISCUSSION

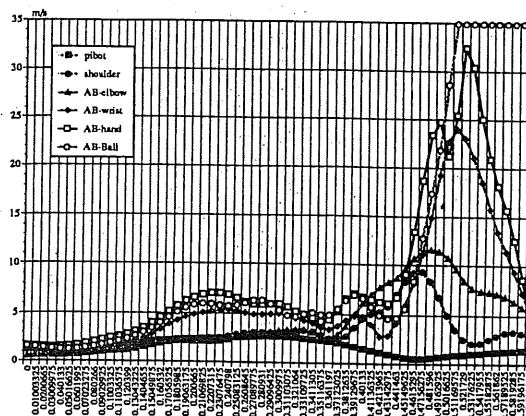
The remarkable alternation of motion has occurred between the late-cocking and the follow-through. In comparison of mass in the case of baseball, torque quantity naturally increase in turn of (LB), (BB), and (HB), and also time reached extremal value points are shifted, which is particularly notable on the elbow joint. It shows that there is a risk for a trainee to be injured because a torque peak comes at the different timing and the different quantity from the usual, especially in the case of using a heavy ball. Otherwise, though the mass of (AB) is same as (HB), the torque graph in the case of (AB) almost resembles (BB) on the shoulder and elbow pronation-supination. Another feature in the case of (AB) has appeared on the relation between the hand velocity and the ball velocity on Graph-2. These are caused from the ball shape of American football that the distance between the ball center and the hand is longer than in the case of baseball. It indicates that there exists a knack which adapts a feature of the ball shape, and it seems that a wrist motion is one of the most important knacks in the case of American football.

Table-1: Ball Condition

Condition Name	Mass (g)	Ball Shape
BB	141.8	Baseball
HB	425.0	Baseball
LB	70.9	Baseball
AB	425.0	American Football



Graph-1: Joint Torque (Elbow)



Graph-2: Velocity in the Case of (AB)

MEASUREMENT OF MUSCLE SHAPE USING MR IMAGES AND ACTIVE CONTOUR MODEL

Hiroyuki Tanaka*, Nobutoshi Yamazaki* and Takeo Nagura**

* Department of Biomedical Engineering, Faculty of Science and Technology, Keio University, Yokohama, Japan

**Department of Orthopedic Surgery, Keio University School of Medicine, Tokyo, Japan

INTRODUCTION

Muscle strength and location of individuals are fundamental data for motion analysis. Those data can be obtained from the three dimensional measurement of muscle shape in intact body. We developed in this study an automatic extraction method of muscle shape from magnetic resonance (MR) images.

METHODS

The idea of active contour model is proposed by Kass et al. (1988) as "Snakes". The active contour model is an elastic ring composed by nodes with a torque spring which are connected by a spring each other. We constructed a three-dimensional muscle model using the elastic rings as shown in Fig.1.

The model shape was made initially by tracing the outline of muscle in MR images of a subject. Then a few cross-sections of the model were placed in a object MR images by operator, the other cross-sections were settled automatically according with the connection of the nodes in the initial model shape (Fig.2). After the setting of the muscle model, each cross-sectional active contour was shortened in order to be involved in the object region (Fig.3(a)). The shortened active contour was expanded again to minimizing the summation of following energy functions; (1)Distance from nodes to outline: The brightness around the outline pixels changes suddenly on the border of a region. Distance energy was defined as a value which was less if nodes were closer to the outline pixels. (2)Distribution of brightness around nodes: Skeletal muscles are often closely placed. In this case, the distribution of brightness in the region of the object muscle is useful to distinguish the region from neighbor muscles. (3)Area of the active contour; (4)Distance between cross-sectional nodes: Each muscle has an adequate range of anatomical size. These two energy functions were defined as values

which increased if the size of active contour was not in adequate range of anatomical size. (5)Distance between longitudinal nodes: The interval between cross-sections do not change in this modification. However distance between longitudinal nodes changes. According to the node position this energy function was defined as a value which increased if the distance was longer than initial distance. (6)Angle between cross-sectional links; (7)Angle between longitudinal links: These two energy functions were defined as values which increased if the shape of active contour was far different from the shape of anatomical knowledge.

The extraction process were repeated from the top cross-section to the bottom cross-section.

RESULTS AND DISCUSSION

Calculation was performed by a workstation (HP-model 712/80), and it took 15 minutes on average per muscle. Fig.3(b) shows that the active contour model was well fitted to the cross-section of psoas major. The shape agrees with the traced line by an orthopedist. The developed method can apply to other muscles with remarkably different shape, and clarify the differences of individual musculo-skeletal system (Fig.4).

CONCLUSION

Image processing system for measurement of the individual muscle shapes was developed using active contour model and anatomical knowledge. Tendon part could not be estimated yet from MR images because of indistinctness.

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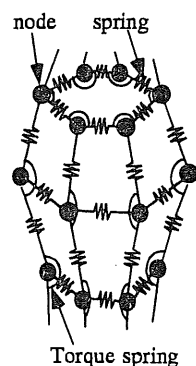


Fig.1 3D model of active contour

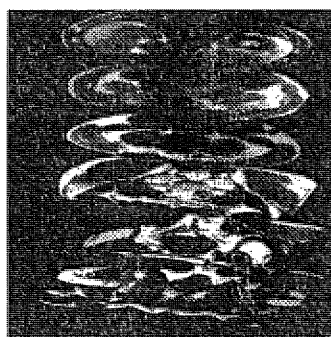


Fig.2 Piled MR images and located standard shape model (Psoas major)

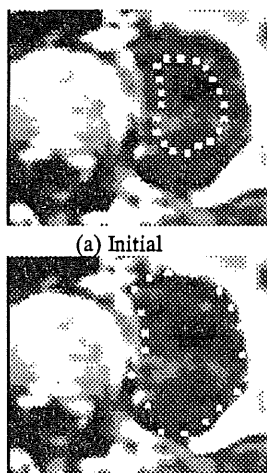


Fig.3 Model fitting (Psoas major)

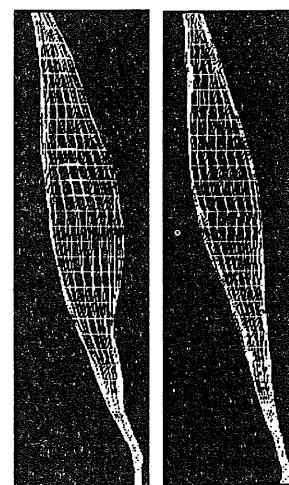


Fig.4 Individual difference

Establishment of an equation for VO₂ estimation from HR

Tsutomu Sasaki, Kazuhiko Tsunoda, Isao Katoh

Biomechanics Lab. Hokusei Jr. College, South -4, West -17, Chuo-ku, Sapporo, 064, Japan
Canon Trading Co. Tokyu-mita Bill., Mita, Minato-ku, Tokyo, Japan

INTRODUCTION

The linear regression could not been applied to estimate energy expenditure for various activities. Establishment of a accurate model for VO₂ estimation could not be only applied have benefit for sports training but also various medical situations, for example the diagnosis of diabetes. If the VO₂-HR relationship could be clarified with significant correlation, HR could be not only an indicator of exercise intensity but also would be able to give evidence of energy expenditure in physical activity. In our mathematical model for VO₂ estimation, HR is an independent variable.

The purpose of this study is to establish mathematical models for calculating energy consumption by heart rate.

ESTABLISHMENT OF EQUATIONS

Typical metabolic response patterns during a period of post exercise have different time delays. VO₂ and HR can be determined using equations (1) and (2).

The equations are:

$$VO_2 = V_{rest} + (V_{max} - V_{rest}) \cdot e^{-\frac{1}{V_c} \cdot t} \quad (1)$$

$$HR = H_{rest} + (H_{max} - H_{rest}) \cdot e^{-\frac{1}{H_c} \cdot t} \quad (2)$$

V_{rest} , $V_{reserve}$, H_{rest} and $H_{reserve}$ as a constant values are obtained by metabolic tests. V_c and H_c are time constants.

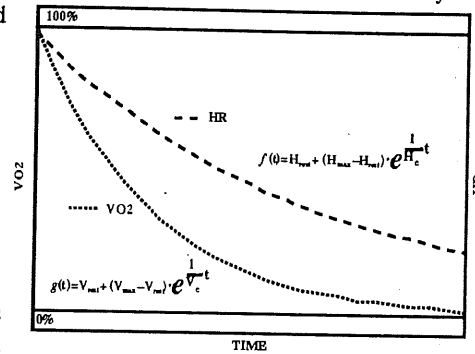


Fig.1 Attainment of VO₂ and HR to resting level.

RESULTS

VO₂ and HR converge to a different steady state level. Exponential equations sufficiently characterize the VO₂ response. The time constant for VO₂ is constant across work intensities during exercise. Metabolic responses can be delayed at the period of post exercise. Responses of VO₂ and HR were delayed in the attainment of a resting metabolic level after exercise. Equations (1) and (2) that have VO₂ and HR as dependent variables, are expressed as time series exponential functions. In this case, time (t) is an independent variable in each equation. Each time constant is different in that decrement pattern. Responses of VO₂ and HR have respectively different time delays (See Fig.1). Thus, the attainment of each resting level was dependent on the time constant. Equation (3) is defined by follows:

$$\dot{VO}_2 = h(HR) = V_{rest} + V_{reserve} \left(\frac{HR - H_{rest}}{H_{reserve}} \right)^{\frac{H_c}{V_c}} \quad (3)$$

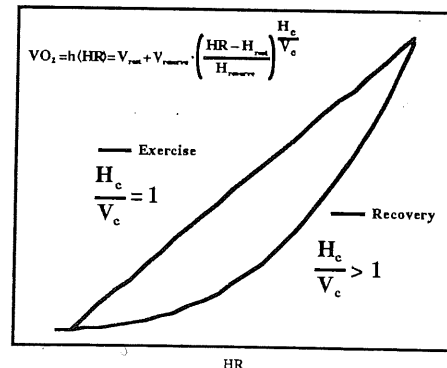


Fig.2 Results of estimation equation (3)

The power component in equation (3) is obtained from equation (1) and (2) which describe data during the post exercise. Therefore, we can use equation (3) for VO₂ estimation in various human activities.

NEW METHOD OF THE ENERGY LOSSES MEASURING IN SPORT SHOES SOLES' AND ARTIFICIAL SURFACES MATERIALS

Dychko B., Moscow, Russia

The energy losses in the cycle "compress-stretching" is one of the main characteristics of the sports shoes soles (SS) and artificial surfaces (AS) materials.

According to results (Denoth, 1978) test-object begins to take-off from the tested material when the sample is deformed. It's possible there are both residual deformation and local destructions and damp take place in the contact zone. But "drop-test" method can't to estimate at the mechanical works were spent on these processes separately.

PURPOSE

The purpose of the our research was the elaboration of the method to determine of the "clear" energy losses without the mechanical work has be done on the local destruction and deformation in tested SS\AS materials and increase the precision of the estimation.

METHODS

We assumed that the total mechanical losses' work may be presented as the sum two independent works: work which has be done on the local destruction and deformation (WLDD) and work which has be done on the "clear" damp energy losses (CEL). WLDD is depended on the mass and form of the test-object. CEL may be calculated if we assume that the test-object's mass will be equal "zero". It's necessary thus SS\AS materials have non-linear characteristics and their energy absorbed standard properties must be found in this condition. This given theoretical assumptions permit to offer such technique of processing CEL in SS\AS materials using drop-test. It's necessary to use several test-objects with the different masses thrown down from the several heights. After that we can find the relation between take-off velocity and test-object mass. Extrapolating the found dependence on the "zero" test-object mass it's possible to find the value of the take-off velocity corresponding to very small WLDD in the contact zone and then to find CEL.

RESULTS

This method was tested on two AS samples - rubber and tartan (thickness -20 mm) mounted on the force platform's cover. There are the metallic shots (3,5,7 kg) were used in the experiment. The shots were thrown down from the heights 9, 12 and 15 cm. CEL was calculated over the energy losses factor.

The analysis of the experiment' results showed next:

1. The real energy losses in tested samples are more then CEL.
2. With the rising test-object's landing velocity from 1.33 to 1.72 m/s CEL increase both in the rubber as in the tartan.
3. With the test-object's landing velocities 1.33 - 1.72 m/s CEL in the rubber is more than in the tartan. It's necessary to note that with the landing velocity corresponding to the foot landing velocity in the sport walk and jogging the rubber's CEL are more than in the tartan. With the rising of the test-object' landing velocity CEL in the rubber/tartan are compared.

CONCLUSIONS

1. The new method of the determining of the energy losses in the SS/AS materials without WLDD was elaborated.
2. The elaborated method may be used both in the laboratories and in the real using.
3. The CEL may be used as the standard for SS/AS materials.

IMPULSE FORCE OF THE SHOES SOLE DURING LANDING MOTION OF HUMAN

A. Nagata, K. Terada, A. Kuga and Y. Sakurai*

Department of Human Sciences, Waseda University, Tokyo, Japan

*Midori Anzen Kogyo (Ltd.)

INTRODUCTION

The purpose of this study was to measure the impulse force of the shoes sole during landing motion of human from some height. Then, human's absorbable function would be analyzed for physic parameters of the shock and acceleration to avoid some damage or infirmities safely.

METHODS

In order to evaluate correctly the shock impulse force at the foot and the worn shoes sole, the new designed measurement apparatus has been manufactured (referred Figure). Using this measurement apparatus, the shock force and oscillation during the landing motion from 10cm, 20cm and 30cm heights were measured with four piezo-type sensors and one piezo-type accelerometer. Comparing with falling and landing motion of human and the load (30kg). These shock differences were evaluated about the force, the acceleration, and the rising slope degree and then ANOVA were applied for these measurements to estimate the clinical value to become a dangerous motion. These kinds of the sole materials (A,B,C) were used at these measurement of the fall-landing. A and C materials are harder and softer.

RESULTS and DISCUSSION

Shock force 34KN, acceleration 83G, and rising slope degree 14.5 (KN/ms) were acquired at the experiment of the load 20kg-10cm fall at the case of material A, while C sole material showed five-half values of 7KN, 20G and 2.1 KN/ms respectively. Those measurement depended upon various kinds of materials from elasticity and viscosity (absorbability) contents, significantly.

In the human's fall-landing experiment, the sole shock force 4.0KN, (2.5) acceleration 2.5G, (1.4) and rising slope degree 0.14KN/ms (0.07) were obtained at the shoe condition of A material or C (refer ()). Then human fall-landing motion showed about twenty half, values, respectively. Those human value would show some shock-absorbable characteristics of the joint action of the lower legs and reaction motion the upper extremities.

CONCLUSION

Polyurethane material of the shoe sole showed the lowest shock force, and oscillation frequency among the other material, statistically. Reaction motion of the human's extremities estimated the minimum values at the fall-landing experiment. To avoid the damage for the body, the sole material and the fall-landing motion have become the more important factors at the shock absorbability.

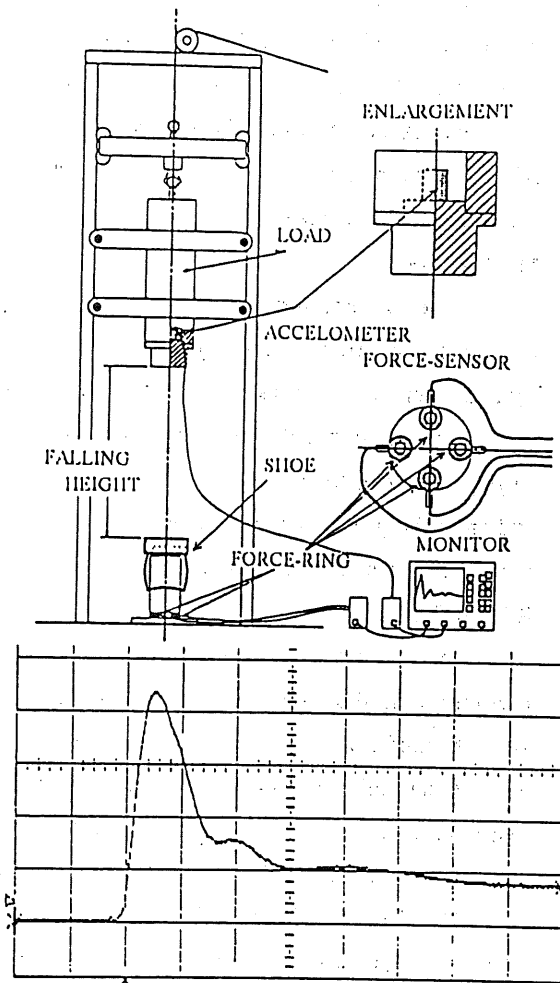


Fig. AN APARATUS OF FALL-LANDING FORCE

EFFECTS OF THE HALLUX SEPARATED SHOES DURING WALKING

T. Kawai¹, Y. Hasegawa¹ and H. Ihara²

¹Sports Sciences Laboratory, Mizuno Corporation, Osaka 559, Japan

²Department of Orthopaedic Surgery, Kyushu Rosai Hospital, Kitakyushu 800-02, Japan

INTRODUCTION

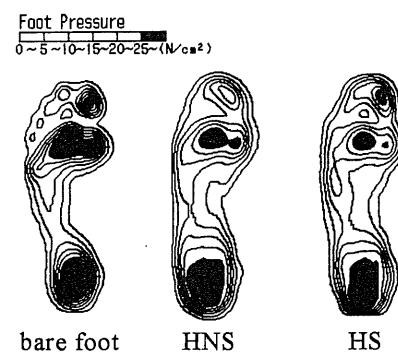
Human foot is endowed by nature with the plantar arch of elastic shock absorbing system. During barefoot walking, the plantar arch was elevated by the plantar fascia that was drawn for the hallux in phalangeal dorsiflexion before the take-off. This system was called "windlass action" (Hicks:1961). Human foot with shoes, this system doesn't seem to be worked well. From this point, it was suggested the hallux separated shoes that was separated by materials (hanao) between the hallux and others. However, there are a very few reports on effects of the hallux separated shoes. The purpose of this study was to investigate effects of hallux separated shoes during walking.

METHODS

The subjects were 10 healthy males (age 23-47). They were walked five times at each following condition 1) bare foot, 2) hallux non-separated shoes + normal socks (HNS), and 3) hallux separated shoes + toe separated socks (HS). The difference between #2 and #3 conditions was only hallux support. Foot pressures per 1 cm² during walking were measured by EMED System (Novelgmbh) at a sampling frequency of 100 Hz, which performed interpolation and smoothing. Maximum pressure at each pressure point and ratio of duration over 20 N/cm² around the hallucal region to contact time (D/C) were calculated.

RESULTS

An example of maximum pressure at each pressure point drawn per 5 N/cm² of equal pressure line at each condition is shown Figure 1. Larger pressure around the hallucal region was observed on bare foot and HS than on HNS. Foot pressure of HNS distributed to the hallux and second toe. D/C on bare foot was $2.2 \pm 0.34\%$, $1.6 \pm 0.31\%$ on HNS, $2.1 \pm 0.45\%$ on HS. The significant differences were obtained between bare foot and HNS (6/10 subjects) and between HS and HNS (5/10).



DISCUSSION

Larger pressure around the hallucal region observed on bare foot and HS indicates ground kick of the hallux before the take-off. However, distributed pressure between the hallux and second toe before the take-off was observed on HNS. The "windlass action" by the plantar fascia was performed enough to kick the ground by the hallux before the take-off. This study showed that the "windlass action" obtained on bare foot was similar to that on HS.

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NEW ATHLETIC SPIKED SHOES CONSIDERING THE POWER TRANSMISSIVE FUNCTIONS

Yoshikuni Takamoto, Akira Kataoka and Masanobu Inohara
Research Institute of Sports Science, ASICS Corporation, Kobe, Japan

INTRODUCTION

The athletic spiked shoes can be regarded as the media that transmits the power exerted by kicking to the surface. The purpose of this study is to clarify its efficiency of the power transmission in the sprinting performance. According to our concept, decreasing loss in kicking power during contact of spiked shoes with the surface can contribute to accelerate the sprinting speed.

Figure 1 shows power loss factors caused during contact phase. These factors can be divided into three types, slip of the shoes on the surface, deformation of the shoes and slip of the foot inside shoes.

The 1st factor is derived from slip of the shoes on the surface. The slip mainly depends upon shape and disposition of the spikes and the projections. Almost previous study concerning the spiked shoes focused on the 1st factor. This study discusses other two power loss factors.

The 2nd factor is derived from deformation of the shoes. During contact phase, loads act on the shoes in various directions. Especially in turns and/or hurdling, torque due to the rotatory movement about the body axis (e.g. swing of legs and arms) is 8-10 times as much as that of straight sprinting. This indicates that high stiffness sole is required in order to prevent deformation.

The 3rd factor is derived from slip of the foot inside shoes. It is considered that this factor depends upon stiffness of the uppers. To add to the above torque, in turns, the lateral load is 30-40 times as much as that of straight sprinting. Standard soft uppers cannot firmly hold the foot, and let the foot move sideward and rotate. Prevention of unnecessary foot slip also subdues injuries due to the excessive torque about the joints.

METHODS

Normal spiked shoes and structural improved spiked shoes (Figure 2) to decrease the 2nd and 3rd power loss factors were used in this study. In-plane flexural stiffness of the improved shoes' soles was reinforced 4 times the normal shoes' to decrease deformation against torque. Further, the improved shoes' uppers used high-elastic resin straps to hold foot slip inside shoes.

Five male sprinters (Ave.P.R.100m:10.8s) were instructed to accelerate 50m to reach the measuring point at maximum speed. Each subject used in both the shoes sprinted 3 times. Speed at 5m intervals, ground reaction force were recorded at the point.

In addition, sagittal and frontal movement of the shoes were filmed by two high-speed videos to detect deformation of the soles and the uppers.

RESULTS

Comparing speeds in both the shoes, the improved shoes were significantly 2% faster than the normal ones (9.32m/s vs. 9.50m/s, $p < 0.01$). This improvement corresponds to 0.202s shortening in 100m race. Figure 3 shows that the improved shoes decreased braking impulse by 17%, and then increased the forward impulse (propulsion-braking). Contact time of the improved shoes were shorter than that of the normal shoes by 2%. In the improved shoes, the sharp peak in torque appeared, which proves improvement of power transmission. Moreover, it was observed that the improved shoes restrained deformation of the soles and the uppers.

DISCUSSION

These results suggest that the decreases in the 2nd and 3rd power loss factors consequently increase the forward propulsion. In other words, the high efficient spiked shoes of power transmission contribute to accelerate the sprinting speed.

P.S. Koji Ito (FUJITSU) drastically broke the national record in 200m with the improved shoes (20.44s->20.29s).

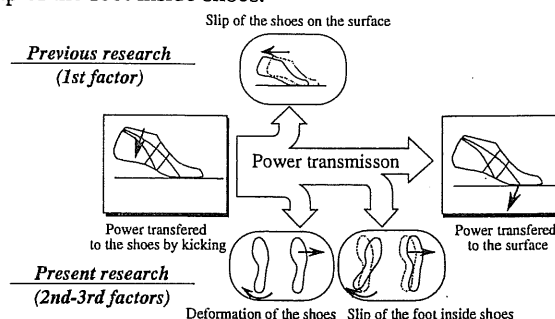


Fig.1 Power loss factors during contact phase



Fig.2 The structural improved spiked shoes

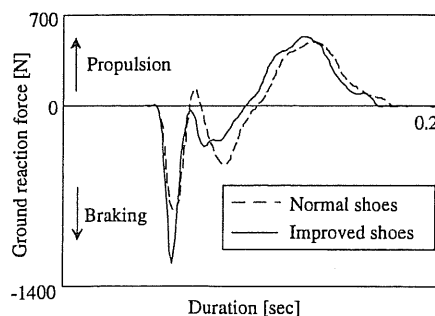


Fig.3 Ground reaction force for different spiked shoes

SUBJECT AND MATERIAL ANALYSIS OF TENNIS RACKET POWER

Tzyy-Yuang Shiang, *Ron-Li Su

National College of Physical Education and Sports, *Da-Yeh Engineering College, Taiwan

INTRODUCTION

The power of tennis racket can be represented by the coefficient of restitution (COR) or rebound ball speed during the ball racket impact. There are lots of kinetic and elastic potential energy transfer during the impact. Therefore, the frame stiffness and string tension of tennis racket are the major factors for racket power. Brody [1979] and Groppe [1987] pointed out that the relationship between string tension and COR is not linear. Shiang [1995] found out that the stiffer racket has better power. General and extensive discussions of frame stiffness and string tension can be found in numerous studies [Brody, 1979, Elliott, 1982] But more detail about both frame stiffness and string tension should be studied together to thoroughly understand the complex relationship among power and racket parameters. Furthermore, all of these work have been done in the laboratory to simulate the real ball and racket impact phenomenon. The discrepancy between laboratory test and realistic situation were still not clear. The goal of current work is to provide a more extensive evaluation of tennis racket power. Therefore, this study conducted both static material test and dynamic subject test to determine the power of racket in the laboratory and on the court.

METHODS

Tennis rackets with four levels of frame stiffness and various string tension were tested in this study. Frame stiffness were measured using Babolat measuring machine which scaled the stiffness from 0 (flexible) to 100 (stiff). The subjects for this study were 3 male experienced tennis athletes, with a mean age of 22.33 years (S.D. 0.47), and mean tennis age of 9 years (S.D. 2.94). All subjects were currently the members of National Tennis Team during the test. Each subject was asked to smash serve 30 times using randomly chosen rackets with various frame stiffness and string tension combination. The serving ball speed was recorded by the speed detector to determine the power of the rackets. The high speed video camera (1000 frames/sec) and motion analysis system (PEAK Performance System) were used to record the impact situation and calculate COR in material test. Briefly, a tennis racket was rigidly clamped on the handle part, and a tennis ball was controlled to impact the center of the racket head vertically at approximate 20 m/sec velocity. The ball speed before and after the impact were determined using video camera and motion analysis system. CORs of various rackets were then calculated to determine the power.

RESULTS

The COR and ball speed for various racket are shown in Table 1. Two-way ANOVA was used to analyze the effects of frame stiffness and string tension on racket power. Figure 1 displays the contour plots of COR and ball speed.

Table 1: Rebound ball speed [km/hr] and (COR) of rackets with various stiffness and string tension

		String Tension (lb)			
		40	50	60	70
Stiffness	50	176.67 (0.482)	174.18 (0.508)	178.27 (0.498)	167.46 (0.494)
	55	173.19 (0.502)	174.83 (0.538)	179.35 (0.520)	168.68 (0.522)
	60	172.14 (0.512)	174.05 (0.558)	176.77 (0.540)	165.67 (0.536)
	70	170.40 (0.536)	176.74 (0.574)	179.15 (0.564)	168.55 (0.560)

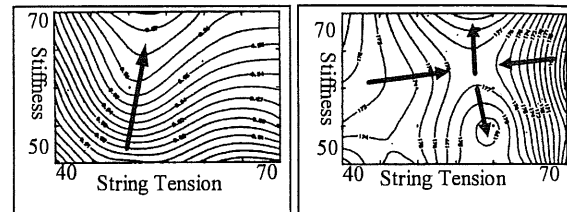


Figure 1: Contour plots of COR and rebound ball speed

DISCUSSION

The results show that the stiffer racket acquire higher COR. These results give a clear indication that frame stiffness is the major factor for the racket power in static material test. But frame stiffness become less dominant compared to string tension in dynamic subject test. Another interesting result is that the COR of 50-lb string tension is significantly higher than those of others. The ball speed of 60-lb string tension is significantly higher than others. It seems more likely that the best power region changed from 50-lb string tension in material test to 60-lb in subject test as displayed in Figure 1. It is not too surprising that the ball speed contour has more variation since subject test results will be affected by the subject physiological and psychological conditions. Therefore, the athletes with better skill and more stable performance were preferred in this kind of study. Based on the results obtained, further attention should be paid to dynamic subject analysis of tennis racket to clarify the realistic ball-racket impact phenomenon.

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INFLUENCE OF BRUSHING ON CURLING STONES

J. Gavin Reid, School of Physical and Health Education and Department of Anatomy and Cell Biology
Queen's University, Kingston, Ontario, K7L 3N6, Canada.

INTRODUCTION

The sport of curling requires that stones be delivered towards the house at the opposite end of the ice sheet with accuracy. Brushes are used to clean the ice in front of the moving stone to ensure that the desired trajectory is maintained and to guarantee that the stone travels close to the required distance.

Brushing produces a minute amount of heat due to friction, which is enough to melt the ice and lower the coefficient of friction for a brief period. Force production by brushing has been recorded on a force platform and measurement of power and friction on an ice surface on a force platform (Day and Reid, 1993). These findings are of interest, however, it is important to study the effects of brushing on the curling stone on the full playing surface.

The purpose of this study was to assess the effect of brushing with different brushes on the absolute distance traveled by a stone when the stone was launched reliably at known initial velocities.

METHOD

The time a stone takes to travel from the near hog line to the T-line is commonly used as an indication of the condition of the ice. The longer the time period the faster the ice. This time reflects both the condition of the running surface of the stone and the ice. The time commonly used for slow or heavy ice is 22 seconds, fast or keen ice is 24 seconds. Times of 25 to 26 seconds would reflect exceptionally fast ice while 21 seconds or less exceptionally slow ice. Three conditions of the ice/stone relationship were carefully prepared and used for testing the effects of using different brushes on the distance the stones traveled given the same initial velocity. One hog hair brush and three different synthetic brushes were tested. Student t-test was used to determine statistically differences between brushes and effects of brushing compared with no brushing.

A reliable prototype launcher was utilized to propel the curling stones down the ice with desired rotation and initial velocity. To check the reliability of the launcher, the initial velocity of the curling stone was recorded using an infrared timing device and a microcomputer.

For all trials, two curlers were instructed to brush as hard as possible from the near hog line until the stone stopped. There were five trials for each condition of brushing and five trials of no-sweeping. The order in which the brushes were tested was randomized to alleviate any systematic bias.

RESULTS AND DISCUSSION

Under the medium ice condition the range of values between the brushes was no more than 0.5m. No significant differences were observed between the brushes. Under the slow ice condition the control variable (no brushing) produced slightly lower values. The total range between the brushes and the control was no more than 1m (statistically significant at $p < 0.10$).

Under fast ice condition the overall distance traveled by the stone increased by approximately 1m over that observed under slow ice condition. Under this condition, the range between brushes was 2.1m, with no significant difference between the brushes. On average, the brushing increased the distance traveled by the stone by 2.5m over the control variable of no brushing (statistical significance at $p < 0.001$).

The effect of brushing varies with the stone/ice condition. On slow ice the velocity of the stone is greater and the time period for sweeping is less than on fast ice. Therefore, the sweepers can carry the stone further on fast ice and with experience can place the stone close to the desired position in the house.

CONCLUSIONS

1. With the same initial velocity brushing resulted in the stone gliding further than no brushing.
2. The combined average distance the stone traveled while being brushed, beyond that of no brushing, was 0.7 m (2.3 ft), 1.1m (3.6 ft), and 2.5m (8.2 ft) for the slow, medium and fast ice condition.
3. Brushing was most effective on fast ice (24 seconds from hog to T-line) and least effective on slow ice (22 seconds), however the coefficients of variation were greatest under fast ice and much lower under slow and medium ice condition.
4. Under each of the ice conditions there was no significant difference between the brushes tested. Each brush performed the task similarly.

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COMPARISON OF DYNAMIC AND ISOMETRIC KNEE EXTENSOR MUSCLES STRENGTH AND POWER OUTPUT IN NORDIC COMBINED ATHLETES

M. Pääsuke, J. Ereline, H. Gapeyeva

Institute of Exercise Biology, Faculty of Exercise and Sport Sciences,
University of Tartu, Tartu, Estonia

INTRODUCTION

Muscular strength and power are parameters that are recognized as influencing performance in a great number of sporting activities. Study of both these muscular characteristics has acquired even more importance in the functional evaluation and profiling of athletes, in training control, and in prevention of injuries. However, differences exist in the literature concerning the relationships between "strength" measured by various methods and especially the relationship between isometric and dynamic parameters of strength.

The purpose of this study was to determine the relationships between three methods of measuring of knee extensor muscles strength and power output by means of isometric and isokinetic dynamometry and force platform system.

METHODS

Seven male sportsmen participated in this study. All of them are members of the Estonian team of nordic combined. Their age, height and body mass were (mean $\pm$ SE) 20.7 ± 1.4 years, 181.0 ± 0.9 cm and 74.4 ± 1.6 kg, respectively. After 10 minutes warming up the subjects performed next exercises.

Isometric dynamometry. Maximal isometric force (MVC) and rate of force development (RFD) of knee extensor muscles were measured using an electromechanical dynamometric chair. The subjects were seated with the knee and angles equal to 90° and 110° , respectively.

Isokinetic dynamometry. Peak torque (PT) and maximal power output gravity-corrected values at three different angular velocities (1.53, 3.06 and 4.59 rad/s) were recorded in the same extremity and position which was tested isometrically, using Cybex II isokinetic dynamometer.

Vertical jumps. Three types of vertical jumps were performed on the force-platform (75 x 75 cm): squat jumps without countermovement (SJ), countermovement jumps without (CMJ1) and with the help of arms (CMJ2). The height of rise of center of gravity (H) and maximal power output of leg extensors muscles (W_m) were calculated.

RESULTS

Isokinetic PT at all of three angular velocities (213.9 ± 11.7 , 114.9 ± 11.2 and 38.6 ± 11.0 N·m resp.) correlated strongly with each other ($r = .91$ -. $.94$) and with isometric RFD (2478 ± 207.2 N/s, $r = .78$ -. $.96$). Significant correlation was also found between isometric RFD and height of CMJ1 (0.397 ± 0.021 m, $r = .82$). Isokinetic PT at angular velocity of 4.59 rad/s correlated with height of SJ (0.374 ± 0.019 m) and CMJ1 ($r = .75$ and $.88$, resp.). High correlation was found between W_m of SJ (879.1 ± 67.8 W) and isokinetic PT and maximal power output at angular velocity of 1.53 rad/s ($r = .80$ and $.75$ resp.). The mean value of isometric MVC (809.3 ± 39.7 N) had strong correlation with RFD ($r = .93$). All these correlations were significant at level $p < 0.05$.

DISCUSSION

Isometric and isokinetic dynamometry as well as vertical jumps (Bosco C et al., 1983) are three methods to measure force and power in athletes. Results of present study support that isometric RFD, isokinetic PT and height of CMJ without the help of arms are intercorrelated. However, no significant correlations between isometric MVC and characteristics of dynamic force were found. These results support the concept that there exists a some general muscle strength component (Knapik, Ramos, 1980; Hortobagyi et al., 1989). The differences observed between these three methods to measure force and power output of knee extensor muscles in well trained athletes may be discussed in terms of the participation of other muscle groups in the movement, the relative utilization of elastic and reactive components of dynamic force, and other characteristic of ballistic movements.

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EFFECT OF BALANCE OF ISOKINETIC LEG STRENGTH ON THE RELATIONSHIP BETWEEN THE 100M RECORDS AND THE LEG STRENGTH IN MALE SPURTERS

T.Tagawa and M.Fukuoka

Research Institute of Sports Science, ASICS Corporation, Kobe, JAPAN

INTRODUCTION

It is generally accepted that leg strength is an important component of sprinting speed and there have been some studies which investigated the relationship between the muscle strength around the knee joint and the 100M records in sprinters³⁾, however few previous studies have focused on the relationship between the muscle strength around the hip joint and the performance¹⁾. None of them has ever tried to examine the relationship from the viewpoint of the balance among the muscle strength around the knee or hip joint.

The purpose of this study was twofold; 1) to classify Japanese male sprinters based on the balance of isokinetic leg strength in 4 major muscle groups (hip flexors, hip extensors, knee flexors, knee extensors) which mainly produce force in sprinting, 2) to investigate the relationship between the isokinetic leg strength and the 100M records in each classified group.

METHODS

The subjects consisted of 72 Japanese male sprinters (Age:19.2±2.0yrs, Ht:1.75±0.04m, Wt:65.7±4.6kg, PR/100m:10.94±0.29sec(10.19~11.50sec)) and performed hip flexion and extension movements in a supine position, and knee flexion and extension movements in a sitting position using an electro-dynamometer (MYORET,Kawasaki heavy industry Co.Ltd.). The testing velocities were 180deg/sec for the hip joint and 300deg/sec for the knee joint movement. The peak torque divided by the subjects' body weight was used to analyze.

RESULTS

Significant negative correlation was obtained only between the 100M records and the knee extensors in all the subject(p<0.05).

In order to classify the subjects, principal component analysis was applied to the strength of 4 muscle groups and extracted 3 factors (Table1 shows the eigenvectors). Individual factor scores of the second and third factor were calculated to draw a scattergram. The classification of the subjects was executed with the 4 areas of the scattergram and each areas could be interpreted as representing "knee flexors type", "hip extensors type", "hip flexors type", "knee extensors type" respectively (Figure1).

There was no significant difference in the 100M records among the classified groups. It was found that the hip flexors (p<0.05) in the "hip extensors type" and the knee extensors (p<0.01) and hip extensors(p<0.01) in the "hip flexors type" had significant negative correlations with the 100M records respectively.

DISCUSSION

There has been fairly general agreement that the knee flexors is the most important and significant to the performance in Japanese male sprinters³⁾. However, the results of this study in all the subjects showed a significant correlation only between the knee extensors and the performance¹⁾. It is suggested that the subjects in this study were more homogeneous; i.e. elite sprinters, whose knee flexors have been already well-trained²⁾

We applied principal component analysis to classify the Japanese male sprinters into 4 groups by the balance among the strength of 4 major muscle groups. There was no significant difference in the 100M records among the classified groups. On the other hand, there were significant differences among the classified groups when the relationship between the strength of 4 muscle groups and the 100M records were examined. Especially for the hip flexors in the "hip extensors type" and the hip extensors in the "hip flexors type" could not be observed significantly without the classification of the subjects. These results might suggest a new finding that there is not a specific classified group superior to the performance, however the muscle strength which makes the greatest contribution to the 100M record is changed according to the balance among the muscle strength around the knee or hip joint.

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Table1 The eigenvectors of the principal components

	Factor1		Factor2		Factor3
knee ext.	0.453	hip ext.	-0.555	knee ext.	-0.496
knee flex.	0.493	hip flex.	-0.385	hip flex.	-0.470
hip ext.	0.493	knee flex.	0.456	hip ext.	0.338
hip flex.	0.556	knee ext.	0.580	knee flex.	0.647

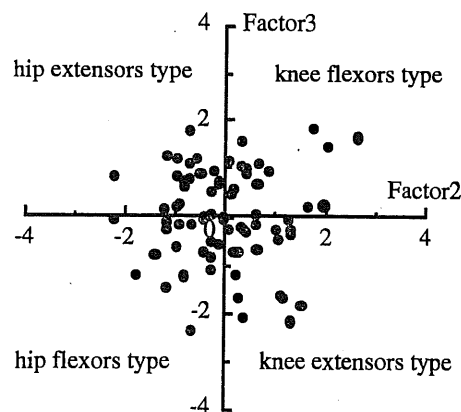


Figure1 : The classification of the sprinters by the second and third factor scores

RELATIONSHIP BETWEEN THIGH MUSCLE COMPOSITION AND SPRINTING PERFORMANCE IN JUNIOR ELITE SPRINTERS : 4-YR FOLLOW-UP STUDY

Y. Kano, S. Kuno, S. Katsuta and M. Miyamaru

Inst. of Health and Sport Sci., Univ. of Tsukuba, Tsukuba, Japan.

INTRODUCTION

Few studies have reported on the relationship between sprinting performance and morphological features of thigh muscles in junior elite sprinters. The aim of this study was to clarify the characteristics of thigh muscle composition and sprinting performance in junior elite sprinters by examining longitudinal changes of the thigh muscles cross-sectional area over a four-year period.

METHODS

Seven adolescent boys who had shown high ability of sprint running, participated as subjects over a four-year period between their 12th and 16th year. Axial images of the thigh muscle were taken by magnetic resonance imaging (MRI; Signa, General Electric; 1.5T) at upper (70%), middle (50%) and lower (30%) position between major trochanter and latera lintercondylar tuberculum in femur. From these images, the cross-sectional areas (CSA) were measured for the quadriceps femoris (QUAD), the hamstring+adductor group (HAM+ADD). Longitudinal changes of these parameters were examined over 4 years.

RESULTS AND DISCUSSION

The results of the regression analysis showed significant correlations between 100 m sprint-time and both CSA of QUAD and HAM+ADD at 70% position ($r=-0.69$ and $r=-0.83$, respectively). Also there were significant correlations found between 100 m sprint-time and CSA of QUAD or HAM+ADD at 50% position ($r=-0.58$ and $r=-0.55$, respectively). No significant correlations, however, were observed between sprint-time and CSA of QUAD or HAM+ADD at 30% position. These results suggest that greater muscle volume at upper position of the thigh affect sprinting performance. Of particular interest was the fact that a much higher correlation coefficient was observed between changes in sprint-times and CSA of HAM+ADD at 70% position over 4 years. We have previously reported that there were significant correlations between 100 m sprint-time and CSA of adductor and hamstring muscles at 70% position, but none regarding quadriceps femoris muscles at 70 or 50% positions in adult sprinter. This fact means that greater muscle volume of hamstring and adductor at upper position affect sprinting performance. Therefore, hamstring +adductor muscle area may be an important determinant of sprint performance in elite junior sprinters.

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Table Correlation coefficients between CSA of thigh muscle and 100 m sprint-time.

	70% position	50% position	30% position
QUAD	-0.69 **	-0.58 **	-0.07
HAM+ADD	-0.83 **	-0.55 **	-0.27

** $p < 0.01$

CROSS-SECTIONAL AREA OF COXAL PSOAS MAJOR REGULATES SPRINT TIME

S. Kuno, Y. Enomoto*, K. Miyashita*, H. Akima*, Y. Kano*, K. Yamanaka*, M. Okada and S. Katsuta*
Center for Tsukuba Advanced Research Alliance (TARA), *Institute of Health and Sport Sciences,
University of Tsukuba, Tsukuba, Japan

PURPOSE

The purpose of this study is to discuss a hypothesis that the cross-sectional area of coxal psoas major be critical as a factor to regulate the sprint capability.

METHODS

The subjects used were 14 male university track and field athletic club students majoring in sprints (with an averaged 100m-dash time of 10.7+/-0.29 sec and 200m's of 21.9+/-0.72 sec) and 19 male soccer club member students. Measured were psoas major areas by MRI and 50m-sprint times.

RESULTS

Fig. 1 shows the relationship between psoas major area and sprint time obtained from each of sprinter and soccer groups. This indicates the presence of a positive correlation between the two measurements in sprinters ($r=0.79$, $p<0.01$). In contrast, there was no relationship observed in the soccer group ($r=0.01$, n.s.). Review of such a relationship for both groups combined as shown in Fig. 2 found its significant presence but the correlation factor was lower than that in the case of sprinters alone ($r=0.47$, $p<0.01$).

DISCUSSION

These results support the hypothesis that the psoas major volume be critical as a factor to regulate the sprint time. A high degree of correlation observed in sprinters who specialize in sprinting despite nothing like that given from soccer suggests the presence of event-specific attributes in this hypothesis.

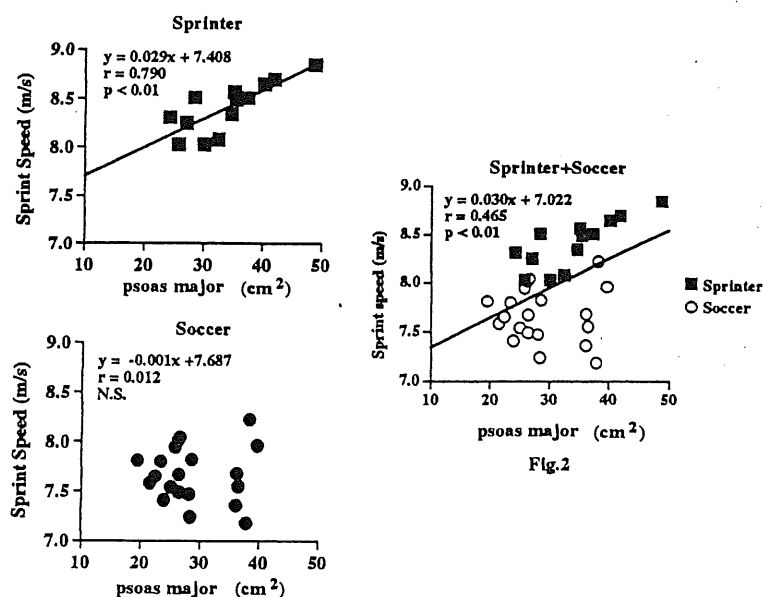


Fig.1

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Difference of pedaling performance estimated by electromyogram between cyclists and noncyclists

Tetsuo TAKAISHI¹ and Toshio MORITANI²

¹Institute of Natural Sciences, Nagoya City University, Nagoya Japan, ²Laboratory of Applied Physiology, Graduate School of Human and Environmental Studies, Kyoto University, Kyoto Japan

INTRODUCTION

Many studies have reported that mechanical efficiency during cycling exercise shows maximum efficiency between 40 and 70 rpm for both cyclists and noncyclists. However, it is well known that cyclists prefer and adopt comparatively a higher pedal cadence (more than 90 rpm) in actual cycling races. We have already shown that a preferred cadence in individuals is related to neuromuscular fatigue of working muscles rather than oxygen uptake, and that the cadence at which minimum neuromuscular fatigue obtained were 70 rpm for noncyclists and 80 or 90 rpm for cyclists (Takaishi et al. 1994, Takaishi et al. 1996). Then, we hypothesized that difference of a preferred cadence between noncyclists and cyclists derives from the difference of skill to transmit the force of legs to a pedal, and that the performance has a great effect on a cadence at which the minimum neuromuscular fatigue will be obtained. The purpose of this study is to elaborate on the cause for the difference in preferred cadence between cyclists and noncyclists in terms of pedal force, neuromuscular activity in lower extremities and oxygen consumption.

METHODS

Seven noncyclists (NON) and seven cyclists (TR) performed five bouts of pedaling work for sampling myoelectric signals and respiratory data (VO₂) on an electromagnetically braked ergometer at five different cadences (45, 60, 75, 90 and 105 rpm). Myoelectric signals were recorded by the surface EMG technique from lower extremities (vastus lateralis, vastus medialis, biceps femoris and gastrocnemius). Pedal force was measured by the force transducer equipped inside the pedal. The signals were amplified and digitized at 1 kHz to obtain iEMG and the peak pedal force. VO₂ for each bout was estimated from respiratory measurements during last 2 minutes of each exercise. The normalized iEMG at 60, 75, 90 and 105 rpm were determined as a percentage to the individual data obtained at 45 rpm. Paired and grouped t-tests were performed for statistical treatment. Differences were considered significant at $P < 0.05$.

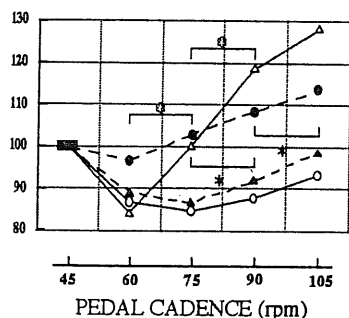


Fig. 1

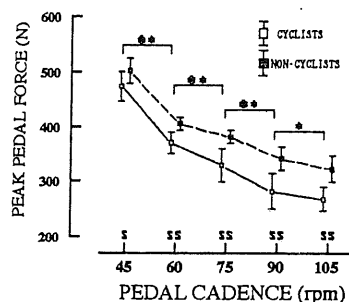


Fig. 2

Fig. 1 Relationship between cadence and normalized iEMG for cyclists (open symbols) and noncyclists (closed symbols). Circles are for m. vastus medialis and triangles are for m. biceps femoris, respectively.

Fig. 2 Relationship between means of peak pedal force and cadence for cyclists (open squares) and noncyclists (closed squares).

RESULTS

Fig. 1 shows the normalized iEMG for m. vastus medialis and m. biceps femoris against the cadence. As for m. vastus medialis, the iEMG for NON demonstrated a quadratic curve with a bottom at 60 rpm and the values significantly increased with increase of the cadence. As for m. biceps femoris, the iEMG for TR showed a significant increase with increase of the cadence. Fig. 2 shows the differences of the peak pedal force. The values for NON were larger than those for TR at all cadences. VO₂ for NON were significantly higher than those for TR at 75, 90 and 105 rpm.

DISCUSSION

The iEMG indicates that noncyclists have to increase muscle activity of knee extensors with increase of the cadence. Ericson et al. (1986) reported that EMG for knee flexors has a small peak during the backstroke, between the bottom and top pedal positions, for positive knee flexion with cyclists. Our data for the iEMGs and the peak pedal forces suggest that cyclists have a certain pedaling skill to alleviate the load for pushing down a pedal by one leg with using positive pedal lifting by the other leg. We speculated that cyclists prefer a higher cadence because they can pedal with lower muscle stress and higher mechanical efficiency with increase of the cadence according to pedaling skill.

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VELOCITY DEPENDENCE OF THE WORK DONE AT THE ELBOW JOINT DURING CYCLICAL MOVEMENTS

D. L. Benoit & J. J. Dowling

Department of Kinesiology, McMaster University, Hamilton, Ontario, Canada

INTRODUCTION

The amount of work produced by the flexors of the human elbow joint were evaluated over a range of velocities during repeated cyclical flexion-extension movements. It has been shown that a resonant frequency can be determined during controlled repeated movements at the elbow joint (Fellows and Rack, 1987). The purpose of this study was to evaluate the amount of work done by the elbow flexors during imposed stretch-shortening flexion and extension at different velocities in order to determine if frequency of movement contributes to changes in positive and negative work output of the elbow flexors.

PROCEDURES

Seven healthy male subjects (from 23 to 38 years of age) took part in this study. The subjects were seated in a custom made device which secured the upper body, while allowing the right arm to be abducted to 90 degrees. The subject's arm was then secured in the testing apparatus, which allowed flexion and extension about the elbow joint. An external torque motor drove the arm apparatus at four frequencies (0.58, 1.5, 2.4 and 3.3 Hz.) for 6 seconds over a constant range (100°-160° extension). During each trial the elbow flexors were stimulated at 50 Hz with a voltage that elicited a torque equivalent to 60% MVC. An additional six second trial was collected for each velocity in which the arm was moved passively through the movement. During all trials, elbow joint angle and torque, as well as biceps and triceps EMG data were collected using a CODAS acquisition system. Work about the elbow was calculated for both the active and passive trials. Passive work was then removed from the stimulated trial, leaving the work produced by the elbow flexors.

RESULTS

The average measured work values can be seen in figure 1. The total, positive and negative work values did not significantly change with velocity, although these work values did tend to be lower at 2.4 Hz. The torque-displacement data were very repeatable throughout each six second trial. The area within the torque-displacement graph (see figure 2) describes the work absorbed during each cycle of movement.

DISCUSSION

It can be seen in figure 1. that measured work did not significantly change as the frequency of the movement increased. Since muscle activation remained constant throughout the trials, the amount of work provided by the elbow flexors should have been reduced by at least 60% based on the force-velocity relationship. Conversely, the negative work produced during the eccentric portion of the movement should have provided an increase of at least 25%. The reasons for the lack of positive work decrease may be due to muscle potentiation or storage of elastic energy provided by the active lengthening phase of the movement. Also, the slope characteristics of the torque-displacement data for this in-vivo study (fig. 2) did not seem consistent with the data found by Rack & Westbury (1974) during a similar investigation of tetanized cat soleus and gastrocnemius muscles.

Figure 1. Muscle Work

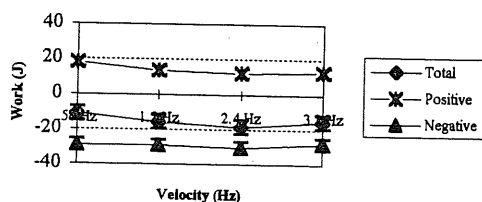
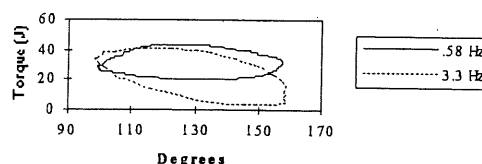


Figure 2. Torque-displacement



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MAINTENANCE OF PERTURBED STANDING POSTURE IN THE MEDIO-LATERAL PLANE --A KINEMATIC AND KINETIC ANALYSIS

S. E. Walt

Department of Safety Science, University of New South Wales, Sydney, Australia

INTRODUCTION

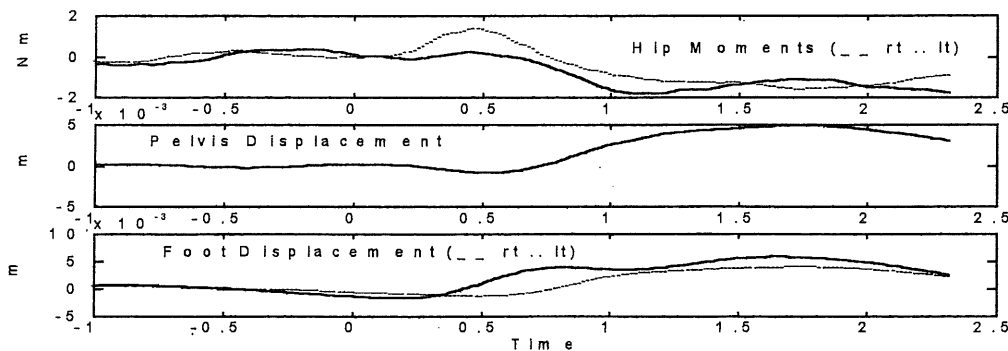
Posture is thought to be controlled through a combination of vestibular, visual and proprioceptive afferent signals. In order to develop a model of postural control, some indication of the relative influence of each input on the generation of the joint torques causing a movement response is needed. The perturbation used to challenge the control system is galvanic vestibular stimulation, a procedure which early literature reports sends a signal along the vestibular nerve resulting in medio-lateral sway. The purpose of the experiment is to analyse the differences observed in kinematic and kinetic variables due to varying combinations of vestibular, visual and proprioceptive inputs.

METHODS

Eight normal young subjects participated in the study. A three-dimensional mathematical model of the body incorporating nine segments was devised. Active infra-red markers were placed on body landmarks to obtain 3-D positional data which was subsequently filtered and differentiated to provide linear and angular displacement, velocity and acceleration. Two forceplatforms were used and the resulting ground reaction forces (GRF) were used with the kinematic information in an inverse dynamics procedure which predicted joint forces and moments. The stimulating electrodes for the galvanic vestibular perturbation were placed behind each ear. Five random trials of three independent variables were tested against the movement response: stimulus duration (3 sec. vs 500 ms), stimulus direction (left vs right), and vision (eyes open vs closed), resulting in a total of 40 trials per subject. Each subject was required to stand quietly with feet close together, one on each forceplatform. Measured dependent variables included: magnitude, duration, and onset latency of linear and angular segmental displacement; ground reaction forces; and onset latency and magnitude of joint moments. Trials were synchronised to perturbation onset and averaged.

RESULTS

The perturbation caused the head, trunk, and pelvis to move in a medial/lateral direction, in the manner of an inverse pendulum rotating about the centre of mass of the pelvis. This response was consistent across all subjects and all conditions. Vision, stimulus duration and direction conditions did not significantly effect displacement magnitude or onset latency of movement and longer perturbations resulted in responses of longer duration. Analysis of vertical GRF revealed a shift in body weight from one side to the other as a result of the changing displacement of the body. Kinetic analysis revealed that moments generated at the hips show a pattern of activity characterised by increasing abduction on one side and decreasing abduction on the other side at a shorter latency of 250 ms after perturbation onset. This is followed by a reversal of moment activity at a longer latency of 500 ms which appears to cause the movement of the pelvis (see figure). This result was consistent across conditions and subjects. Analysis of moments generated at the ankles showed consistent activity at longer latency on the left which may act to stabilise the ankle. No consistent pattern of activity was observed at the right ankle, regardless of the direction of movement.



DISCUSSION

A galvanic vestibular perturbation creates a conflict among the input signals to the balance control system which must be resolved in order to restore balance. The pattern of hip moments generated in response suggests that maintenance of balance is initially under vestibular control until movement, presumably about the ankles, triggers a proprioceptive response which overrides the vestibular input. This resulting hip moment causes pelvis movement which acts to restore balance. Visual input had no significant effect.

SLIDE OF COF ON THE FOOT ACCORDING TO THE CHANGE OF THE STANDING CONDITION AND POSTURAL SWAY

Ryuji Kawamoto and Koji Watarai
The University of Tokyo, Tokyo, Japan

INTRODUCTION

A lot of studies have been carried out on Center of Force (COF) and on postural sway in various fields, such as physiology, physical therapy and biomechanics. But studies which investigated the relationship between COF on foot and postural sway were limited. We conducted standing balance test to investigate 1: the slide of COF on the foot according to the change in the standing condition 2: the relationship between COF on the foot and postural sway.

METHODS

Fifty young healthy men (age: 20.5 ± 3.7) participated in the standing balance test. This test included 1: BO(Both-feet with Eyes-Open, Test time 30sec) 2: BC(Both-feet with Eyes-Closed, 30sec), 3: SO(Single-foot with Eyes-open, 10sec) conditions. In this test, Using the Tactile Sensor System, Fscan (Tekscan. Inc.), which can present COF on the foot on real time, we calculated COF in the anterior-posterior direction on the foot (fCOF), (See Figure1). Then we analyzed how fCOF slid when standing condition was changed. We also measured COF displacement (sway length) by the stabilometer (Anima. Inc.).

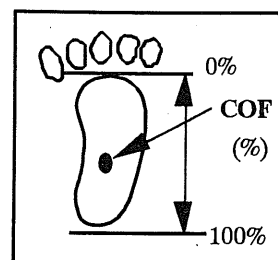


Figure 1. COF on the foot (A-P direction)

RESULTS

Table1. COF on the foot (%) in each standing condition. GroupP (COF posterior, N=20) slid COF more anteriorly than GroupA (COF anterior, N=20) according to the change of the standing condition.

	BO	BC	SO
Group A	36.5	36.0	36.7
Group P	49.8	45.8	42.9

GroupP: BO to BC ($p < 0.05$), BO to SO ($p < 0.01$)

Table 2. Sway length (cm) in each test condition. GroupP (COF posterior) was more unstable than GroupA (COF anterior) in every condition.

	BO	BC	SO
Group A	31.4 (± 5.9)	55.0 (± 18.0)	37.2 (± 9.8)
Group P	46.1 (± 18.5)	78.2 (± 37.0)	46.1 (± 13.9)

BO and BC: GroupA < GroupP ($p < 0.01, 0.05$)

DISCUSSION

During our investigation, clarification of the slide patterns of COF became obvious according to the change of the standing condition. The more difficult standing condition became, the more anteriorly subjects slid their COF. Next dividing subjects into three groups by COF position on quiet standing, we found that for keeping their balance, subjects who maintained their COF posterior during quiet standing were forced to slide their COF more anteriorly when the standing condition became more difficult. On the contrary, subjects with their COF anterior on the foot might not need to slide COF even on the single stance. In addition, we found that sway length of subjects with their COF posterior on quiet standing were longer than that of subjects with their COF anterior. According to this result, we can say that a person with COF anterior at quiet standing tends to be stable not only in the easy standing condition but also in the difficult one.

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EFFECTS OF STRENGTH TRAINING ON FORCE-TIME CHARACTERISTICS IN VOLUNTARY AND ELECTROSTIMULATION EVOKED ACTIONS IN MIDDLE-AGED MEN AND WOMEN

K. Häkkinen¹, U-M. Pastinen¹, R. Karsikas², V. Linnamo¹, W.J. Kraemer³ and R.U. Newton⁴

¹Dept. of Biology of Physical Activity, University of Jyväskylä and ²Diter Electronics, Turku, Finland

³Center for Sports Medicine, the Pennsylvania State University, USA

⁴Centre for Exercise Science and Sport Management, Southern Cross University, Australia

INTRODUCTION

Increases in strength during the first weeks of strength training in previously untrained subjects may be accounted for largely by changes in the quantity and/or quality of neural activation so that 1) voluntary activation of the prime movers is increased and/or 2) there is an improved coactivation of the synergists and/or 3) a reduction in coactivation of the antagonists (Moritani&DeVries 1979, Komi 1986, Sale 1991, Häkkinen 1994). Later strength development takes place primarily due to a gradually increasing contribution of muscular hypertrophy of trained muscles. Strength training leads to shifts primarily on high force portions of the force-time and/or force-velocity curves of trained muscles, whereas the changes will be minor in the early parts of the curves with higher velocities and/or shorter times of force production (Häkkinen 1994). The purpose of this study was to further examine the effects of strength training on muscular performance in men and women at their middle-age by recording force-time parameters of the trained muscles in both voluntary and electrostimulation evoked actions.

METHODS

Twelve healthy middle-aged men (49.6±1.3 yrs; M) and eleven middle-aged women (47.6±1.3 yrs; W) volunteered as subjects and gave written informed consent to participate in the investigation approved by the Ethics Committee of the University of Jyväskylä, Finland. Strength training took place two times a week for 12 weeks using the loads of 40-50%, 60-80% and 70-90% of 1RM. The exercises were performed with slow to medium movement velocities. A David 200 was used to measure maximal voluntary unilateral isometric force-time variables of the knee extensors at week -2, 0 and 12. Electrical stimulation (ES) was given by Diter Stim- stimulator (through two tin electrodes (12x8 cm) by placing the upper electrode over the motor point of the RF muscle. The carrier frequency of stimuli was 2000 Hz (sine wave current) which was modulated by a square pulse at a pulse rate of 50 Hz with a duty cycle of 1:1. Force-time variables were calculated from isometric actions evoked by ES with the highest intensity tolerated (Häkkinen et al. 1995).

RESULTS

Maximal voluntary forces remained unchanged during the 2-week control period (from -2 to 0) with no training but increased during the 12-week training in MW from 435± 25 to 492± 29 N (p<0.001) and from 443±29 to 474±26 N (p<0.05) for the right and left leg, respectively. The average increase of 14% in maximal force of the two legs in M did not differ from that of 9% recorded for W. The shifts on the higher force portions of the force-time curves were significant (p<0.05) in MW, while no changes took place in the earlier portions of the curve. No changes were recorded in electrostimulation evoked forces during the control period, while significant (p<0.05) increases occurred during the training in M for the left leg and in W for the right leg. The times taken to reach the submaximal force levels did not change during the training.

DISCUSSION

The average increase of 12 % observed for isometric strength gain suggests that progressive strength training can lead to substantial increases in maximal voluntary strength of the trained muscles in both men and women at their middle-age. The intensity and duration of the training are of great importance for both middle-aged and elderly people (Häkkinen 1994). Although great caution needs to be exercised with the interpretations, the increase of 9% recorded for both legs in electrostimulation evoked force suggests that considerable training-induced muscular hypertrophy may have taken place in both M and W. The finding that no correlation existed between the increases in voluntary and electrostimulation evoked forces further suggests that adaptations in the voluntary activation of the muscles may have also contributed to the strength gains. No changes took place either in the earlier portions of the force-time curve of the voluntary actions or in the times taken to reach the submaximal force levels by electrostimulation evoked actions. This supports well the concept of the specificity of heavy resistance strength training (Häkkinen 1994). In conclusion, the findings suggest that both neural adaptations and muscle hypertrophy contribute to voluntary strength development in men and women at their middle-age and that specific effects of heavy resistance training on the shape of the force-time curve are observable in both the voluntary and electrostimulation evoked actions.

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FUNCTIONAL AND MORPHOLOGICAL EFFECTS OF DIFFERENT RESISTANCE TRAINING WITH AND WITHOUT A COUNTER MOVEMENT IN ELBOW FLEXORS

Junichi Okada¹ and Senshi Fukashiro²

1: Waseda University, Saitama, Japan. 2: University of Tokyo, Tokyo, Japan

INTRODUCTION

The purpose of this study was to identify the effects on functional and morphological characteristics of elbow flexors in two types of resistance training, called SSC (Stretch-shortening cycle) and NOS (Normal style) respectively, with and without a counter movement in elbow flexion.

METHODS

Six subjects (31 ± 7 yr., 176 ± 4 cm, 74 ± 6 kgw) were asked to train the elbow flexor muscles by dumbbell curl with SSC and NOS randomly assigned for each right/left hand in one subject. The training ($60\%1RM \times 10$ reps $\times 5$ sets) was carried out 3 days/wk for 6 weeks. The static force tests at 90 (F90) and 120 (F120) deg of elbow angles and the dynamic elbow flexion tests (load; 2, 4, 7, 9 and 11 kgw) using the special arm ergometer were executed before and immediately after training period. The joint angle was started from 120 deg in the dynamic test. Force, angular displacement were recorded with 1000 Hz sampling time. The maximum rate of force production (Rmax) and the maximum force (Fmax) were obtained from force-time curve of both tests. Also, the magnetic resonance image (MRI) procedure was used to estimate anatomical cross sectional area (ACSA) and volume (MV) of elbow flexors (biceps brachii and brachialis muscles).

RESULTS & DISCUSSION

The static force of F120 was significantly increased only in SSC (+20%), although those of F90 were significantly increased both in SSC(+18%) and NOS(+10%). This result could be considered as the influence of the specificity of the training, because greater force was generated in the extended position of elbow angle in SSC than that in NOS. The same tendency was observed in the Rmax of the static test. In the dynamic test, the Rmax (+20 to 25%) and Fmax (+15 to 19%) in SSC were significantly improved in most of the loads (2 to 9 kgw except for 11 kgw), but those in NOS were not significantly improved (Figure 1). It was suggested that the training method using counter movement showed positive effectiveness for force and power production in spite of the low intensity of the training load. On the other hand, MV was increased only in NOS (+13%, see Figure 2). It was concluded from these results that training condition using the counter movement were more effective for force production and conventional isotonic training was more effective for muscular hypertrophy. The characteristic of SSC training could be considered as resulting from the large contribution of the tendon tissue.

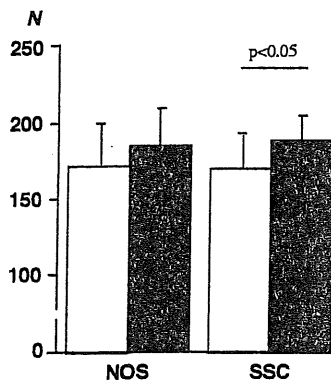


Fig.1 Fmax of 7kgw during dynamic elbow flexion test

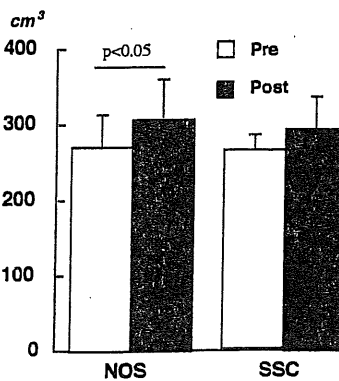


Fig. 2 Muscle volume of each training

VELOCITY SPECIFICITY TRAINING IN ELBOW FLEXORS

Pousson M, Amiridis IG, Cometti G, Morlon B and J. Van Hoecke
Groupe Analyse du Mouvement, Université de Bourgogne, Dijon, France.

INTRODUCTION

Fast velocity training has resulted in some controversy regarding training adaptations (Bell and Wenger 1992). Possible reasons to explain the velocity specificity principle could be associated with the selective recruitment of fast motor units in agonist muscles and with increase in recruited motor units number during the activity experienced in training (Coyle et al. 1981). The purpose of this study was to evaluate the possible specificity of elbow flexion moment gains following fast velocity training by using the Torque/Angular Velocity (T/AV) relationship and the level of activation of agonist and antagonist muscles.

METHODS

Subjects : twenty healthy males were randomly placed into 2 groups, one trained concentrically ($n = 12$) and the other serving as a control ($n = 8$).

Training : subjects performed 6 sets of 8 concentric elbow flexions, three times a week for 7 weeks. They were asked to execute ballistic movements at maximal velocity with a load corresponding to $35 \pm 5\%$ of each subject's one Repetition Maximum. The angular velocity of the movement was recording and calculated (Elite system), with the mean maximal value being $302 \pm 32^\circ \cdot s^{-1}$.

Biomechanical tests were carried out using the BIODEX (Biodex Corporation, Shirley, NY, USA) isokinetic dynamometer. Following a standardized warm-up, subjects were asked to perform three maximal voluntary elbow flexions at nine randomly selected angular velocities from $-60^\circ \cdot s^{-1}$ to $300^\circ \cdot s^{-1}$ and to hold three maximal isometric contractions at 90° of elbow flexion. RMS activities of the Biceps Brachialis (BB) and of the lateral head of Triceps Brachii (TB) muscles were calculated.

RESULTS AND DISCUSSION

Following training, isokinetic torque was not significantly ($p > 0.05$) different from that obtained before training in isometric conditions and at eccentric, slow and intermediate concentric angular velocities. However, torques obtained at $240^\circ \cdot s^{-1}$ and $300^\circ \cdot s^{-1}$ significantly ($p < 0.05$) increased (fig. 1A). Irrespective of angular velocity, the RMS amplitude of the BB agonist muscle was not significantly modified during concentric and eccentric actions (fig. 1A). Normalized RMS values of the TB antagonist muscle obtained after training were inferior to those observed before training (see fig. 1B), but only significantly so at $300^\circ \cdot s^{-1}$.

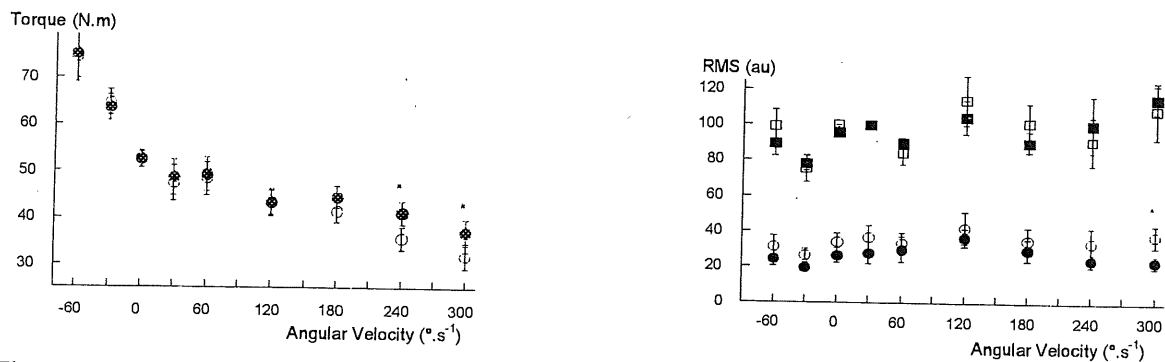


Figure 1A : T/AV relationships before (open circles) and after training (shaded circles). 1B : Normalized Root Mean Square (RMS) amplitude values of BB (rectangles) and TB brachialis (circles).
* represents values significantly different from those observed before training ($p < 0.05$).

In conclusion, the net moment gains at $240^\circ \cdot s^{-1}$ and $300^\circ \cdot s^{-1}$ appeared partly to be dependent upon a drop in the level of coactivation of antagonist muscles. In fact, training may influence directly this level of coactivation giving greater net torque in the intended movement direction (Westing et al. 1991). Nevertheless there was shown to be a dissociation between increases in maximal torque developed at high velocities caused by training and changes in the RMS coactivation. This is in need of further examination.

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CHANGES OF FIRING AND CONTRACTILE PROPERTY OF HUMAN NEUROMUSCULAR SYSTEM AFTER SIX-WEEK RESISTANCE TRAINING

Y. Taniguchi, K. Seki and M. Narusawa
International Budo University, Chiba, Japan

INTRODUCTION

Strength gain after resistance training is known to be induced by the two factors; neural and muscular adaptation. While many reports indicated the hypertrophy of a trained muscle after muscular adaptation, nature of neural adaptation is still unknown. We previously found the decrease of motoneuronal firing frequency during submaximal contraction after 6-weeks resistance training (Taniguchi, et al., 1995), and these changes were considered as a direct sign of neural adaptation. To clarify the underlying mechanism of this, we examined the relation between these neural adaptation and the contractile properties in the first dorsal interosseous (FDI).

METHODS

Five male subjects were volunteered to the experiment. In each training period, subjects exerted a maximal voluntary isometric force with adduction of left index finger and maintained the force for 5 seconds. They performed this training exercise 5 times per set with 25 seconds rest intervals in between, 3 sets per a day with 3 minutes rest intervals in between, 3 times per a week, for 6 weeks. Motor unit activities and contractile properties of muscle were tested on separate days, before, 3 and 6 weeks after the beginning of the training program.

Mean firing rates of motor units were recorded by means of tungsten microelectrodes during voluntary isometric adduction at the level of 20, 40, 60, and 80 % of the maximal voluntary force (MVC) (Seki and Narusawa, 1996). Then, voluntary force - mean firing rate relations were obtained for each subject. Contractile properties was examined by the precutaneous stimulation on FDI (200 - 350 V, 300 μ s, 2 seconds) at nine different frequencies (10, 15, 20, 25, 30, 35, 40, 50 and 100 Hz). Then, tetanic force - stimulation frequency curves were obtained for each subject. Finally, we compared the changes of the voluntary force - mean firing rate relations with that of tetanic force - stimulation frequency curves in each subject.

RESULTS & DISCUSSION

MVC increased from 33.2 ± 4.5 N (mean $\pm$ SEM) to 39.9 ± 4.9 N after 3 weeks and 40.7 ± 7.0 N after 6 weeks ($p < 0.05$). Maximal tetanic force also tended to increase.

In total, 1416 spike trains were analysed; result indicated that the slopes, but intercepts, of voluntary force - mean firing rate relations decreased ($p < 0.01$) after the resistance training (Fig. 1). This tendency was found in four of the subjects. No change was found in one subject.

The changes of the slope and that of maximal tetanic tension were corresponded when comparison was made between subjects; the more the slope decreased, the more the maximal tetanic tension increased; the subject who showed no decrease in the slope did not show the increase of maximal tetanic force.

The increase of maximal tetanic force, not only the MVC, suggest the hypertrophy of the FDI. The decrease of the slope without any changes of the intercepts imply that firing frequency of motor unit decreased especially when a subject exerted larger muscle force. Therefore, corresponding changes of the maximal tetanic force and the slope may indicate that, after resistance training, the adaptation of the muscular size and that of the neural mechanism to generate larger force are not independent.

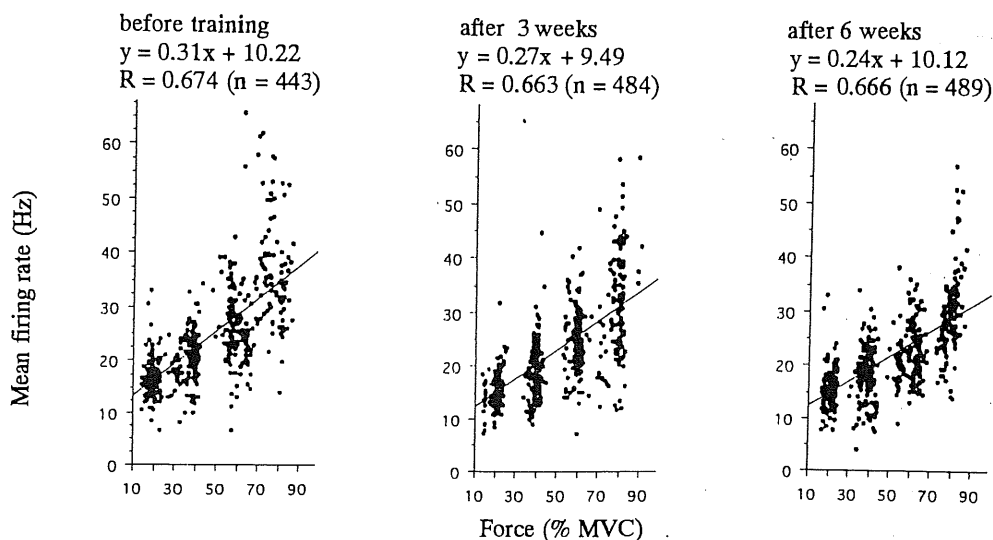


Fig. 1. The relationships of voluntary force and mean firing rate before and after (3 and 6 weeks) resistance training.

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EFFECTS OF ECCENTRIC TRAINING OF THE ELBOW FLEXOR MUSCLES UPON TORQUE-ANGULAR VELOCITY (T/AV) RELATIONSHIPS AND ACTIVATION

Colson, S., Martin, A., Pousson, M., Morlon, B., Van Hoecke, J.
Groupe Analyse du Mouvement, Université de Bourgogne, Dijon, France.

INTRODUCTION

The aim of this study was to examine the effects of eccentric training on the torque/angular velocity relationship with special attention being paid to the myoelectrical activity of agonist and antagonist muscles during maximal voluntary isokinetic elbow flexion. It has been shown that differences obtained in torque output during maximal isokinetic exercise between high and low-strength subjects and between highly-skilled and sedentary subjects could have been the result of an adaptation of both the co-activation level of the antagonist muscles (Carolan and Cafarelli 1992) and the tension-regulating mechanism (Westing et al. 1990). It was of interest therefore to observe these adaptations after training.

METHODS

Subjects : 8 volunteer male students ($X \pm SD$: 20.6 ± 2.2 years , 178.4 ± 7.2 cm, 76.8 ± 6.3 kg) made up a sedentary group whereas 9 others ($X \pm SD$: 21.4 ± 1.1 years, 180.6 ± 6.9 cm, 74.3 ± 10.1 kg) performed eccentric exercise training (EG).

Experimental conditions : Isokinetic tests were carried out using a Biodex dynamometer. Subjects were asked to perform elbow flexions under maximal isokinetic eccentric and concentric conditions (range of motion was 120° ; starting position was full extension at 0°) at 8 angular velocities (from -120° to $240^\circ.s^{-1}$). The myoelectrical activities of biceps and triceps brachii were detected. The Root Mean Square (RMS) was calculated. Normalized RMS data was obtained using the percentage of the RMS value of the maximal concentric action at $30^\circ.s^{-1}$ when the muscle acts as agonist.

Training was performed 3 times a week for 7 weeks consisting of 5 sets of 6 eccentric actions at 100-120% of each individual's 1 maximal concentric repetition.

For the comparison of results, Wilcoxon non-parametric statistical test was used.

RESULTS and DISCUSSION

The torque developed under concentric and eccentric conditions decreased as angular velocity increased, whereas for the EG after training, torque was always greater than those obtained before training and seemed not to be affected by eccentric angular velocities (Fig. 1a). This latter result could reflect an adaptation of the tension-regulating mechanism under high-tension eccentric loading conditions with training, as previously suggested (Amiridis et al. 1996). The gain of torque after training were the highest at both eccentric and the highest concentric angular velocities.

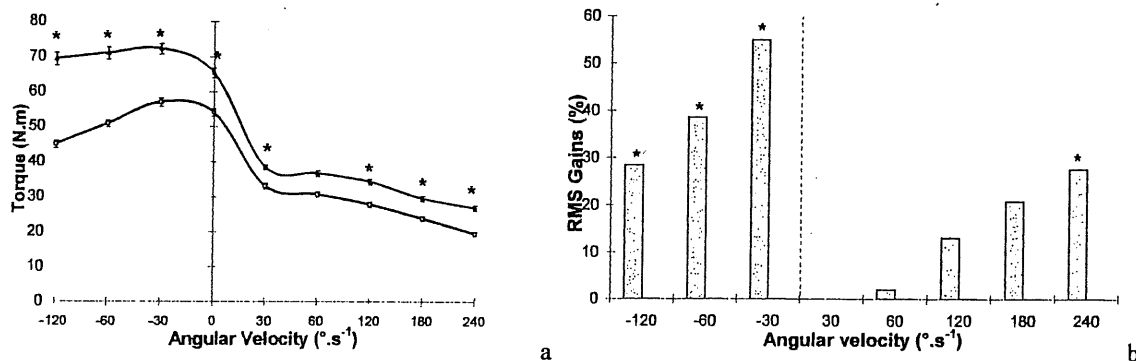


Figure 1: (a) T/AV relationship before and after training ; (b) RMS gains of agonist muscle

For the group EG, and after training, the activity of agonist muscle could explain variations of torque obtained under eccentric conditions and at the highest concentric angular velocity (Fig 1b). For the two groups, the RMS of antagonist muscle was not affected by angular velocity value in concentric as well as in eccentric conditions. However, irrespective of the angular velocity, there was a trend towards a lower antagonist activity after training. In conclusion, it is suggested that the eccentric training consists of a particular process to increase the participation of fast motor units, by increasing the recruited number, inducing a larger torque for the action velocities implicating these motor units, after training.

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NO TORQUE GAIN DURING SLOW ECCENTRIC ACTION IN KNEE EXTENSORS

H.NAGAREDA, K.FUNATO and T.FUKUNAGA

Laboratory of Sports Sciences, Department of Life Sciences, The University of Tokyo, Japan

INTRODUCTION

It is well known that joint torque increases during eccentric action in human skeletal muscle. Most of these findings were recognized at the velocity above 30 or 60 $\text{deg}\cdot\text{s}^{-1}$ by using an isokinetic dynamometer. However, it is not yet clear whether joint torque increases in cases of very slow movement of the joint under 10 $\text{deg}\cdot\text{s}^{-1}$. The purpose of this study is to represent the gains of joint torque during slow eccentric actions in knee extensors.

MATERIAL and METHODS

Five male subjects performed maximum voluntary sustained isometric knee extensions at a knee angle of 70 deg (0 deg : full extension) followed by maximum voluntary eccentric and concentric knee extensions for 5 s. Eccentric and concentric knee extensions were performed with an isokinetic dynamometer (Cybex 770, Cybex JAPAN) from 1 to 10 $\text{deg}\cdot\text{s}^{-1}$. The range of motions encompassing knee angles were from 70 to 90 deg for eccentric and from 70 to 50 deg for concentric conditions, respectively. Measured torque was averaged for 0.5 s before (isometric) and 2-3 deg angle after (concentric and eccentric) the beginning of the lever arm movement of the dynamometer.

RESULTS

Figure 1 shows changes in knee extension torques (TQ) against angular velocities from 1 to 10 $\text{deg}\cdot\text{s}^{-1}$. In case of very slow movements, i.e. under 4 $\text{deg}\cdot\text{s}^{-1}$, the eccentric TQ (eccTQ) indicated no significant increase compared to the isometric TQ, while the TQ of concentric action decreased significantly.

DISCUSSION

The interesting point in the present results is to observe no significant increment of eccTQ in slow velocities of eccentric action less than 5 $\text{deg}\cdot\text{s}^{-1}$. Muscle-tendon complex consists of muscle fiber and tendinous tissue. Tendinous tissue should be stretched without the change of muscle fiber length when the knee extensors perform eccentric action at a very slow velocity, resulting no gains in torque output.

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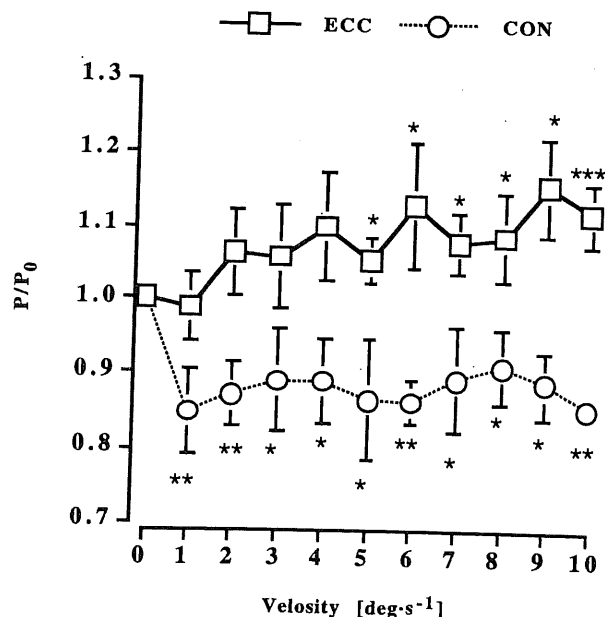


Fig. 1

STRENGTH GAINS AND FATIGUE ARE UNRELATED IN THE RECOVERY OF MUSCULAR STRENGTH AFTER IMMOBILIZATION

T. Hortobágyi, W. Dempsey, D. Fraser, and P. DeVita. Biomechanics Laboratory, East Carolina University, Greenville, NC, USA

INTRODUCTION

It is still unclear what type of mechanical loading is the most effective to rapidly recover muscular strength and size after immobilization. In addition to mechanical loading it has been also proposed that fatigue during exercise may be a stimulus for muscular strength and hypertrophy (Rooney *et al.*, 1994). However, fatigue and the nature of mechanical loading were not assessed simultaneously in prior studies. The purpose of the present work was to address the question whether fatigue has to occur during resistive exercise training using different mechanical loadings in restoring muscular strength following limb immobilization.

METHODS

The left quadriceps of 24 healthy young men and 24 women were tested for concentric, eccentric, and isometric torque on a dynamometer. Testing occurred prior to a 3-week period of immobilization, after the cast was removed, and after 12 weeks of training. Six men and 6 women were randomly assigned to one of 4 groups. All exercising subjects trained the left quadriceps muscle either with concentric (Group 1, $n = 12$), eccentric (Group 2, $n = 12$), or in a combination of concentric and eccentric contractions (Group 3, $n = 12$). Group 4 was a non-exercising control ($n = 12$). Subjects exercised 3 times per week for 12 weeks and performed 4 to 6 sets of 6 to 12 repetitions of maximal isokinetic contractions at 60 °/s.

RESULTS

A group (4) by contraction mode (3) by time (3) factorial ANOVA with repeated measures on the last two factors (followed by a Tukey's post-hoc contrast) revealed that eccentric training improved eccentric strength 43 %, concentric training improved concentric strength 27 %, and mixed training improved eccentric strength 51% and concentric strength 46 % (all $p < .05$). In addition during training, the eccentric group and the eccentric portion of the mixed group showed ~20-30% fatigue, i.e. torque scores actually increased by the end of sessions while the concentric training and the concentric portion of the mixed group showed ~25% fatigue ($p < .05$). Fatigue correlated with strength gains $r = 0.00$. Full recovery from immobilization occurred between weeks 3-4 with eccentric training, week 4 with mixed training, and week 6 with concentric training, the rate of recovery being significantly ($P < .05$) different between eccentric and concentric training.

DISCUSSION

Strength gains occurred without fatigue during eccentric training indicating that mechanical loading must be a primary factor in development of muscle strength and muscle size. Incorporation of eccentric loading in training resulted in greater muscular strength gains, faster rate of recovery, and less or no fatigue. At the time of the study, only 4 of 17 surveyed area physical therapy clinics incorporated eccentric actions in strength recovery programs. The present results in conjunction with recent reports that submaximal eccentric loading is similarly effective in inducing significant strength gains (Hortobágyi *et al.*, 1996) underscore the importance of incorporating eccentric contractions in rehabilitation and training paradigms. Because larger strength gains occurred without fatigue in conjunction with eccentric training, the current findings are particularly relevant for conditions where fatigue during exercise may be a limiting factor in the ensuing strength gains and hypertrophy.

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Supported in part by NICHD and ECU's Research / Creative Activity grants.

SPECTRUM ANALYSIS AT THE SURFACE OSCILLATION AND M-WAVE ABOUT THE LOWER LIMB ATROPHY MUSCLES

T. Miura and A. Nagata *

Graduate School of Human Science, Waseda University, Tokyo, Japan.

* Faculty of Human Science, Waseda University, Tokyo, Japan

INTRODUCTION

The purpose of this study was designed to estimate a compositive ratio of various muscular fibers types using the power spectrum of the surface oscillation signals and that of M-wave, that were generated by the electric current stimulus. Those estimation about the compositive ratio were characterized with the frequency analysis at electromyogram and oscillatory wave signals about the atrophy muscle noninvasively.

METHODS

Oscillatory wave signals were recorded on the triceps surae with the mini-accelerometer, and M-wave signals were measured on the gastrocnemius with two surface electrodes at the same time. Those surface oscillatory wave and M-wave signals were analyzed at both healthy (uninjured) and atrophy (injured) legs of nine subjects wore a plaster cast and nonmoved for about one month (refer Fig.1). Maximum entropy method (MEM) was applied for those signals to calculate frequency bands of Mean Power Frequency (MPF) and Peak Power Frequency (PPF) that were calculated with sampling time of ten milliseconds and on each signal.

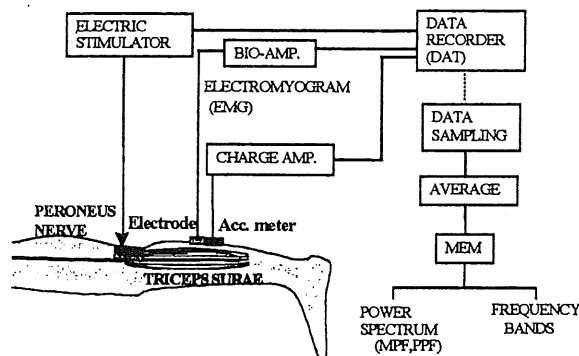


Fig.1 Experimental chart

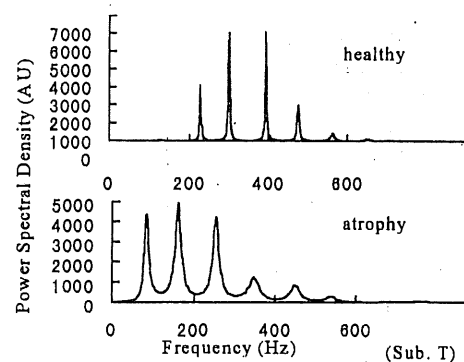


Fig.2 Power spectrum of M-wave

RESULTS

These experiments elicited distinctive different power spectra of the surface oscillatory wave and M-wave signals about healthy or atrophy leg, respectively. In the case of the atrophy muscle, the frequency bands of M-wave spectra were obtained averaged 80Hz (MPF), while those MPF of the healthy M-wave were gotten about 120Hz. Those frequency difference between both MPF were demonstrated by Student's *t*-test method, significantly ($p < 0.01$). Additionally the power spectrum of the surface oscillatory wave signals were appeared at averaged 16.9Hz on the atrophy signals, and about 19.7Hz were elicited on the healthy ones, significantly. Then, at the atrophy case, longer duration of vibration signals were characterized remarkably as compared with the healthy case.

DISCUSSION

Those frequency analysis suggested that high frequency bands of the healthy muscle (MPF, PPF) would be depended upon main firing recruitment of the fast-twitch type of the motor units and upon the muscle fiber type II during evoked oscillation and muscle contraction. On the other hand, the atrophy muscles, lower frequency band would be related to the slow twitch type of the motor units recruitment and to the alteration of the muscle fiber composition to type I.

3D ANGULAR VELOCITIES MOST RELATED TO BALL RELEASE VELOCITY IN THROWING

Christopher P. Sherwood, Richard N. Hinrichs, and Gary T. Yamaguchi*

Departments of Exercise Science and *Chemical, Bio, and Materials Engineering, Arizona State University, Tempe, Arizona, USA

INTRODUCTION

The purpose of this study was to determine the three-dimensional (3D) angular velocities most related to release velocity in baseball throwing using subjects with a wide range of throwing abilities. In the past, 3D analyses of throwing have been limited to reporting mean data obtained from elite throwers (e.g., Feltner and Dapena, 1986). Many of these studies have implicated rapid shoulder internal rotation (SIR) of the throwing arm near release as one of the most important body movements in throwing (Dillman et al., 1993; Feltner and Dapena, 1986; Pappas et al, 1985). SIR angular velocities in these studies were reported to be greater than 100 rad/s at ball release in elite throwers. However, these studies calculated internal rotation angle using only the joint centers to represent a stick figure of the arm. Typically the cross product of the upper arm vector (shoulder joint to elbow joint) and the forearm vector (elbow joint to wrist joint) is used to get an elbow flexion/extension axis. This axis vector is then compared with the trunk's reference frame to determine the shoulder int/ext rotation angle. The major problem with this model (hereinafter referred to as the "stick figure model") is that the cross product becomes unstable as the elbow approaches full extension near release leading to erroneous values of SIR angle and angular velocity. Contributing to this instability are the "carrying angle" of the forearm relative to the upper arm at full elbow extension and how it changes with supination/pronation of the forearm. Therefore, a secondary purpose of this study was to compare the accuracy of the stick figure model for measuring SIR angles with a complete rigid body model using three markers per body segment to determine Cardan angles. This rigid body model did not use the forearm points at all to determine SIR and hence avoided the problems associated with carrying angle and elbow extension near release.

METHODS

Twenty-one right-handed college-aged males were recruited as subjects. After appropriate warm-up, subjects performed three maximal baseball throws while being filmed with two 120 Hz video cameras (Peak Performance Technologies, Inc.). A third camera (60 Hz) was placed 29 m away perpendicular to the throwing direction to select the one trial per subject with the highest ball velocity for full analysis. 3D coordinates of the ball and 11 points on the body (joint centers as well as two additional markers attached each to the forearm and upper arm) were computed using the DLT algorithm. Raw position data were padded prior to being smoothed using a Butterworth digital filter with cutoff frequencies ranging from 9-15 Hz depending on the point and coordinate. Cardan rotations were used to compute joint angles relative to a predefined neutral position. These included left/right rotation, lateral flexion, and flexion/extension for the trunk; ab/adduction, horizontal ab/adduction, and internal/external rotation for the shoulder; and flexion/extension, valgus/varus, and supination/pronation for the elbow/radioulnar joint. For comparison, SIR angles were also computed with the stick figure model as described above. Velocities were computed using finite difference differentiation.

RESULTS

Ball velocities at release ranged from 21.9 to 37.7 m/s. The external markers attached to the upper arm and forearm were found to have negligible effect on ball velocity; the average ball release velocity (BRV) for all subjects was 29.2 m/s with markers in place and 30.1 m/s with markers removed. For the fastest seven subjects (with an average BRV of 34.7 m/s, comparable to BRVs measured on elite throwers in past research) the rigid body model calculated an average SIR angular velocity at release of 48.3 rad/s, while the stick figure model calculated an average value of 79.1 rad/s. Hence, the stick figure model overestimated the "true" SIR angular velocity (as measured with the rigid body model) by more than 60%. Only two angular velocities (measured at ball release) were found to be significantly correlated with BVR ($p < 0.01$). These were SIR angular velocity and elbow extension (EE) angular velocity. As expected SIR angular velocity was positively correlated with BVR ($r = 0.751$), but EE angular velocity was found to be *negatively* correlated with BVR ($r = -0.675$). In other words, the fastest throwers had the lowest EE angular velocities at release. In addition, all subjects had EE angular velocities that peaked at or prior to release. Although the magnitude of this peak value was not correlated with BVR ($r = -0.098$), the timing of this peak was significantly correlated with BVR ($r = 0.738$) such that the fastest throwers had their EE angular velocities peak *earlier* with respect to ball release.

DISCUSSION

The generation of large SIR angular velocities at release appears to be important for throwing a baseball at fast speeds. In addition, there seems to be a link between elbow extension and shoulder internal rotation such that the fastest subjects sacrifice elbow extension angular velocity at release in order to increase shoulder internal rotation angular velocity for an overall increase in ball velocity. The stick figure model appears to overestimate SIR angular velocity in throwing. Therefore it is likely that the large SIR angular velocities reported in previous studies are inaccurate. More research is needed using the rigid body model to analyze throwing performance.

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COMPENSATIONAL COORDINATION OF RELEASE VARIABLES IN BALL-THROWING MOVEMENTS

Kazutoshi Kudo (Graduate School of Arts and Sciences, The University of Tokyo), Seiji Tsutsui (Department of Physical Education, Aichi University of Education), Tadao Ishikura (Department of Physical Education, Chukyo University), Tomoki Ito (Department of Physical Education, Fukuoka University), Yuji Yamamoto (Research Center of Health, Physical Fitness, and Sports, Nagoya University), Tokyo, Japan

INTRODUCTION

In a ball-throwing task aimed at a stationary target, consistent movements produce consistent performance. However, it should be noticed that varying movements also produce consistent outcomes as far as the variables at the point of release are coordinated with one another so that the trajectories of a ball pass the target. Therefore, we propose a new measure called the *ICRV* (index of coordination of the release variables) calculated by the "shuffle" method to distinguish the effects of consistency and coordination of the release variables on consistency of performance.

METHODS

Eight adult subjects (age 20-24) performed a ball-throwing task for 150 trials. The target was on the floor 5 m away. Throwing movements were videotaped using a charge-coupled device camera (60 fps) with a timer counter. The release point was defined as the coordinates at which the distance between the center of the ball and the left metacarpale radiale exceeded 10 cm. The *ICRV* in a given block was calculated as the ratio of variabilities of the ranges covered by the ball ($\sigma R_{after} / \sigma R_{before}$) by the following procedure. 1) Calculation of variability of the range (σR_{before}). The range for the i th trial was calculated by the following equation:

$$R_{before i} = \frac{V_i^2 \cos \theta_i}{g} \left(\sin \theta_i + \sqrt{\sin^2 \theta_i + \frac{2gY_i}{V_i^2}} \right) + X_i, \quad (1)$$

where V indicates the initial velocity, θ the initial angle, X and Y the release point in horizontal and vertical coordinates for the i th trial, and g is the acceleration of gravity. 2) "Shuffle". This step involves the randomization of the release variables in the block so that the variables for the *different* trials were combined with one another. The compensational coordination among the release variables, if any, should be broken up by this step. 3) Recalculation of variability of the range from the release variables after "shuffle" (σR_{after}). If the variables are not compensated with one another, σR_{after} should be equal to σR_{before} , that is, the *ICRV* equal to 1. We repeated this procedure 1,000 times and took the mean as the score of the *ICRV* to confirm complete randomness.

RESULTS AND DISCUSSION

Variability of the range calculated from the digitized release variables was significantly decreased with practice, $F(4,28)=4.15$, $p<.01$, that means the performance improved with practice for 150 trials in terms of consistency (Figure 1). Among release variables, only the initial velocity became consistent with practice, $F(4,28)=4.03$, $p<.05$. The *ICRV* averaged across subjects was not increased with practice, $F(4,28)=0.82$, $p>.05$ (Figure 2). However, the *ICRV* averaged across subjects and blocks was 1.072, and the sign test showed the score was significantly greater than 1, $N=40$, $p<.01$ (two-tailed). This *ICRV* score means the variability of the range increases by 7.2 % without coordination of the release variable. In addition, variabilities of the range before shuffle were well predicted by the multiple regression equation including *ICRV* as predictor variable (Figure 3). The adjusted multiple correlation coefficient was $r^2=.95$ ($df=38$, $p<.0001$). In contrast the correlation between the observed and predicted variabilities was $r^2=.77$, when the multiple regression model included only the release variables (i. e., without *ICRV*). These results suggest that variability of the release variables can, at least in part, be considered as an emergence of compensational coordination, while the coordination was not improved with practice for 150 trials. In conclusion, the *ICRV* can be considered as a useful measure to quantify coordination of movements.

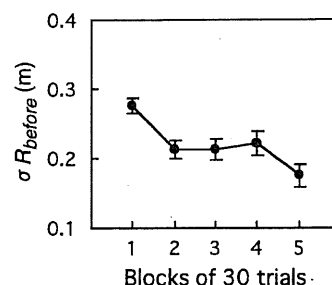


Figure 1. Averaged within-subject variability of the range before "shuffle" for each block. Error bars represent between-subject standard error.

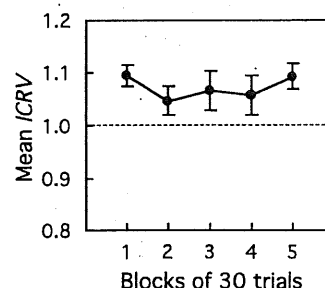


Figure 2. Averaged within-subject *ICRV* for each block. Error bars represent between-subject standard error.

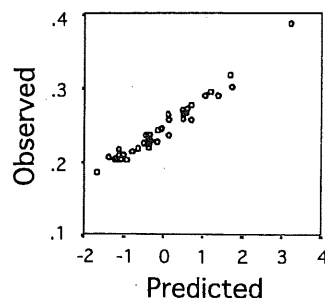


Figure 3. Observed and predicted variabilities of the range for each subject and block.

How and why does a forkball break downward ?

- Trajectory of a pitched baseball. -

Shinji SAKURAI, Yasuo IKEGAMI, Yuji YAMAMOTO, Atsushi OKAMOTO*,
Tohru TERASHIMA**, Hiroyuki NUNOME, and Kyonosuke YABE

Nagoya University, Nagoya, Japan

* Nagoya Women's Junior College of Commerce, Japan

** Chukyo University, Japan

KEY WORDS

Aerodynamics, Lift, Drag, DLT

INTRODUCTION

Many types of pitches are thrown in baseball games, such as fastball (FB), curveball (CB), slider (SL), change-up, and so on. Recently, many pitchers use the forkball (FK) and knuckleball as their main weapon. Characteristics of these types of pitches are generally thought as erratic motion with a sudden drop. The purposes of this study were to investigate the three-dimensional trajectory of the FK pitches and to compare with those of the other types of pitches.

METHODS

Subjects were highly skilled professional and amateur baseball pitchers. Flight of the pitched ball was recorded with two synchronized video cameras (60 Hz, exposure time = 1/10000 s) and analyzed using DLT procedures. A second order polynomial function using the least square method was used to derive time-displacement data of ball trajectory for each pitch. Initial and final velocity, vertical acceleration, and Coefficient of drag force was obtained from the regression equations.

RESULTS

Figure shows the comparison of several parameters among four types of pitches. (FB, CB, SL, and FK) of a professional pitcher. FK (32.7 m/s) was slower than FB (37.0) and SL (36.1), and faster than CB (30.0). The speed reduction ratio from initial to final ball speed was larger in FK (0.850) than any other types of pitches (0.864 - 0.882). The drag coefficient was also largest for the FK pitch, and smallest for FB and SL pitches. Estimated vertical acceleration during the flight was -4.5 (FB), -12.9 (CB), -7.4 (SL), and -9.6 m/s² (FK), respectively. Within the accuracy of the measurement, the erratic motion of the ball in the lateral direction was not found for any FK pitch in this study.

DISCUSSION

The average vertical acceleration value during the flight of FK pitches was very similar to the gravitational one ($g = -9.8 \text{ m/s}^2$), which was smaller than the acceleration of CB. It means that the ball does not drop more steeply in FK pitch than CB pitch. Moreover the acceleration in the latter half of the flight was not larger in any pitches including FK, when compared to the first half of the flight. These results suggest that the breaking straight down of forkball should be an optical illusion of the batter. The remarkable characteristic of the forkball was horizontal deceleration rather than a vertical break with larger vertical acceleration. The reason for the larger speed reduction of the forkball in flight was not obvious, but might be related to the low spin rate of the ball.

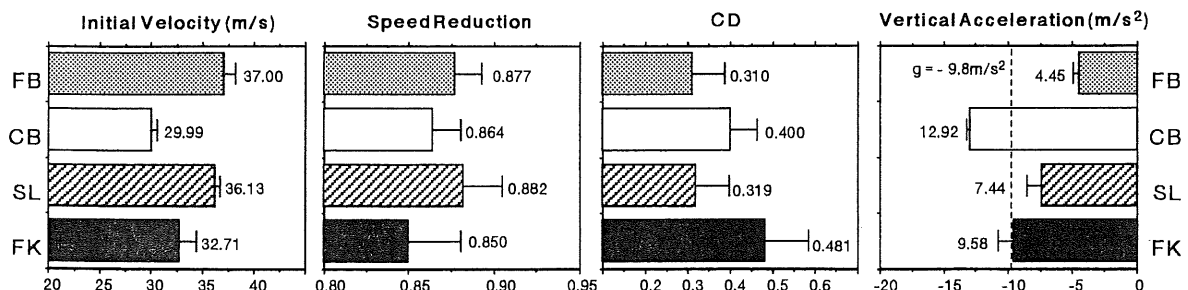


Figure : Comparison of four types of pitches. Speed reduction = final/initial speed, CD = Coefficient of drag. Values are mean of 8 pitches of each type of pitch thrown by a professional pitcher.

KINEMATIC AND ELECTROMYOGRAPHIC CHANGES WITH FATIGUE IN BASEBALL PITCHING

Y. Takada, S.W. Barrentine, G.S. Fleisig, N. Zheng, J.R. Andrews
American Sports Medicine Institute, Birmingham, U.S.A.
Kobe University, Kobe, Japan

INTRODUCTION

One of the main responsibilities of a baseball pitching coach is to determine when a pitcher has become fatigued, as a fatigued pitcher is believed to be less effective and also at higher risk of injury. Sports medicine clinicians share this concern about fatigue in pitching, as the vast majority of pitching injuries are attributed to overuse (Fleisig et al., 1995). Although most people agree that pitching while fatigued is dangerous, the changes in muscular activity have not been studied. Furthermore, kinematic changes associated with fatigue have not been quantified, leaving coaches with no scientifically-based criteria for identifying fatigue. The purpose of this study was to measure kinematic and electromyographic changes in baseball pitching during a simulated game.

METHODS

Ten collegiate pitchers served as subjects in this study. Subjects were tested in pairs in an indoor laboratory. After providing informed consent, history, and physical information, reflective markers were attached to fourteen bony landmarks (Fleisig et al., 1995, 1996). Bipolar surface electrodes were attached to the mid-deltoid, biceps brachii, triceps brachii, and flexor carpi radialis muscles of the throwing arm (Delagi and Perotto, 1980). After stretching and warming up, the pair of subjects alternated pitching for a simulated complete game (18 half-innings), or until unable to continue. In each half-inning, a subject threw up to 7 warm-up pitches and exactly 15 game pitches from a portable pitching mound to a catcher behind homeplate 18.4 m away. As in actual games, the catcher signalled for a pitch type (fastball, curveball, etc.) and location before each pitch.

After each pitch, a researcher announced a result (called strike, homerun, etc.) according to a predetermined randomized chart and the pitch location. Data were collected for three fastballs thrown from a windup position during each half-inning. Using methods described by Fleisig et al. (1996), three-dimensional kinematics were calculated from digitized data collected with a four-camera 200 Hz Motion Analysis system. Electromyographic (EMG) data were collected with a four-channel 1000-Hz Noraxon telemetry system. Motion analysis and EMG data collection were electronically synchronized.

RESULTS

Several kinematic variables showed consistent, gradual changes as the innings progressed. Comparing the ninth inning to the first inning, the following changes were observed. A pitcher had greater shoulder abduction (+5 change), horizontal adduction (+5), and external rotation (+8) at the instant the front foot contacted the mound. During the arm cocking and arm acceleration phases, a pitcher had less shoulder external rotation (-4), upper torso angular velocity (-90/s), and elbow extension velocity (-200/s). At the instant of ball release, a pitcher had less shoulder abduction(-4) and front knee flexion (-9). Ball velocity decreased by 1m/s. Gradual EMG changes were seen for certain subjects, however no consistent changes were seen for the entire group.

DISCUSSION

As expected, ball velocity decreased as the game progressed. This decrease was partially due to reduced velocity generated in the upper torso and the throwing elbow. Decreased velocity corresponded with two kinematic changes often observed by coaches: a lower elbow (reduced shoulder abduction) and a straighter front leg (reduced knee flexion) at ball release. The pitcher tried unsuccessfully to prevent this decreased shoulder abduction by increasing abduction at foot contact. Similarly, the pitcher tried unsuccessfully to prevent decreased shoulder external rotation during arm cocking by increasing external rotation at foot contact. Results from this study help explain how a pitcher alters kinematics during a game. Further research on muscle activity and joint kinetics may provide insight about the mechanisms of overuse injury.

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DYNAMIC ANALYSIS OF THE TENNIS SERVE

Giovanni Legnani[♥] Bruno Zappa[♥] Renza Perini[✱]

Università di Brescia, [♥]Dip. Ing. Meccanica, Via Branze 38, 25123 Brescia, Italy. Legnani@bsing.ing.unibs.it

[✱]Dip. Scienze Biomediche e Biotecnologie, Via Valsabbina 19, 25123 Brescia, Italy

METHOD

This paper presents the results of a dynamic analysis of the tennis serve based on film analysis. Aim of the study was the identification of the dynamic peculiarities of the tennis serve and their correlation to sport performances and various injuries in the back and in the elbow. Following previous work by the authors [4,5], eleven athletes have been filmed while performing a flat first serve. The motion of the arm and of the racket has been recorded by three camera working at a frame rate of 200 Hz. Thirteen markers were put on the subjects (9 on the arm, 1 on the shoulder, 1 on the hip and 2 on the racket). Frames have been attached to the markers to represent the segment motion (upper arm, forearm, hand with racket). The motion of the markers has been evaluated using the Direct Linear Transformation algorithm. The motion of the joints arm has been evaluated by the analysis of the marker movements and considering a 7 revolute degree-of-freedom model for the arm.

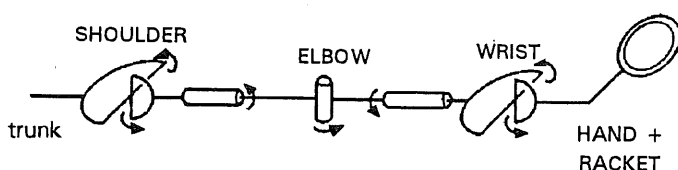


Figure 1. The 7 degree of freedom model

The motion of the arm has been represented by the time history of the 7 angles of the model. The joint displacements have been smoothed and differentiated by optimally regularised quintic splines [6].

The joint velocities and accelerations have been used for evaluating the absolute motion of each segment. This finally has given an

estimation of the actions (forces and torques) applied on the arm. The kinematic and dynamic quantities have been projected on the segment frames to give them an anatomical meaning.

MOTION ANALYSIS

The analysis of the marker trajectories has been performed using special algorithms to increase the accuracy of the joint angle estimation. The position of the markers has been normalised taking into account the redundancy of the marker system and the peculiarities of the anatomical constraints. For instance, the vector defined by the markers placed on the medial and lateral epicondyles has been forced to be normal to the upper arm and the forearm. The racket has been considered to be embedded in the hand, however some relative motion has been observed between some markers. To overcome this problem the relative position of the involved markers has been frozen by a solidification approach [2]. These procedures reduce the effects of errors like: skin motion artefacts, inaccuracy in the trajectories estimation, camera switching. The angles values have been analysed in two steps. First of all their trend has been elaborated to remove solutions jumps near singularities [4,5]. Finally they have been smoothed and differentiated. Simulation tests showed that the adopted algorithms reduce significantly the influence of many errors.

DISCUSSION

The velocity in many joints showed sudden peaks just before the time of the ball-racket impact. This has been previously described by other authors [3]. The adoption of a rigid body model predicts sudden variations or peaks in the joint forces and torques during the impact phase. Torques in the elbow can be greater than 100 Nm [5]. They are suspected to be one of the causes of the so called *tennis elbow*. These peaks are due to the sudden deceleration of the segments at the ball impact. Since the impact phase is not included in the model this deceleration is interpreted as produced by the arm giving the wrong value for forces and torques. An analysis of the ball impact [1] showed that the ball itself slows down the racket velocity of about 50%. A methodology has been applied to take into account this effect which is significant for some joints.

Acknowledgment. Some of the ideas reported in the paper have been developed during co-operative work with B. Elliot, G. Noffal and R. Marshall who collected the data at University of W. Australia.

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THE RELATIONSHIP OF SEGMENT ROTATION WITH RACKET VELOCITY AT IMPACT IN TENNIS GROUNDSTROKE

K. Takahashi¹, B. Elliott² G. Noffal³

¹The Faculty of Education, Saitama University, Urawa Japan.

²The Department of Human Movement, The University of Western Australia, Perth, Australia

³The Department of Kinesiology and Health Promotion, California State University, Fullerton, U.S.A

INTRODUCTION

Generating high racket velocity and hence ball velocity, is a critical factor in successful tennis performance. Research on the tennis forehand showed that segment rotations, sequencing from proximal-to-distal are exploited for this purpose(1). However, the question is as to whether the racket velocity depends on these segments equally is still to be answered. In this study correlations between contributions of individual segment rotations and resultant racket velocities at impact in tennis groundstroke were examined.

METHODOLOGY

The direct linear transformation method was used to calculate the three-dimensional displacement histories of selected landmarks on the nine high performance tennis players recorded on film at 200Hz hitting a flat forehand groundstrokes.

To compute angular velocities (ω) of the segments, orthogonal coordinates (x, y and z) with unit vectors(i, j and k) were constructed. If angular velocity of vectors of i, j and k about perpendicular axes to them were defined as ω_i, ω_j and ω_k , then we can express ω_i as $\omega_i = \omega_y j + \omega_z k$, where ω_x, ω_y and ω_z are x, y and z components of angular velocity vector ω . Similarly Vector ω_j and ω_k were expressed. From these, we had ω :

$$\omega = \frac{1}{2}(\omega_i + \omega_j + \omega_k),$$

where ω_i, ω_j and ω_k were obtained as follow. The Linear velocity of the tip of i is expressed as: $V_i = \omega_i \times i$. By taking a cross product of this equation with i and noting that the scalar product of i and ω is zero since i is perpendicular to ω , then ω_i was approximated as

$$\omega_i = \frac{i_{n+1} \times i_n}{t_{n+1} - t_n},$$

where t and n show the frame rate and the frame sequence respectively. Angular velocities ω_j and ω_k were then calculated using the same procedure. From these values the contributions of each segment were calculated and correlation coefficients between the segment contributions to the racket velocity and the impact value were computed.

RESULT & DISCUSSION

Results showed high correlations between contributions of the upper limb segment rotations and racket velocities at impact(Table 1: $r = .834$ for x-direction, $r = .898$ for y-direction and $r = .898$ for z-direction). However, both the contribution at impact and the maximum contribution of upper and lower trunk segments didn't correlate significantly with racket velocity at impact except for y-direction of hip rotation($r = .543$). These results indicate that the contribution of upper limb rotations, which occur in the latter part of the stroke, is primarily responsible for the derivation of the racket velocity at impact in tennis forehand groundstroke. These results support research completed by Elliott, Takahashi and Noffal(2) on flat, Topspin and topspin lob forehand groundstrokes.

Table 1: Midhip, hip, upper trunk and upper limb contributions to impact x-direction(forward), y-direction(sideways) and z-direction(upward) racket-head velocities(ms⁻¹) in tennis forehand ground strokes and correlation coefficient between contributions of segments and racket velocities at impact(n=9)

		Midhip Linear Movement	Hip Rotation	Upper Trunk Rotation	Upper Limb Rotation
X-direction	Mean ($\pm$ S.D.)	.18(.35)	5.14(2.48)	2.21(2.20)	8.73(3.31)
	Correlation Coefficient	.247	.366	.173	.834**
Y-direction	Mean ($\pm$ S.D.)	-.11(.42)	1.56(.77)	-.20(1.09)	.54(2.35)
	Correlation Coefficient	.180	.543*	.161	.898**
Z-direction	Mean ($\pm$ S.D.)	.14(.20)	.00(.90)	2.50(1.5)	6.32(2.44)
	Correlation Coefficient	.066	.060	.161	.898**

**and * Significant correlation with racket velocity at impact($P < .05$) and ($P < .001$)

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A COMPARISON OF UPPER LIMB MOVEMENT BETWEEN TENNIS AND SOFT-TENNIS PLAYER

Lin-Hwa Wang, Hong-Wen Wu*, Fong-Chin Su*

Physical Education Office & *Institute of Biomedical Engineering
National Cheng Kung University, Tainan, TAIWAN

INTRODUCTION

Except tennis, soft-tennis is also popular in oriental countries. Due to using different rackets and balls, some distinct techniques exist between tennis and soft-tennis. Although there are many studies published regarding the biomechanics in tennis [1,2], the biomechanical behavior in soft-tennis has not been well investigated. What are the correct upper limb swing techniques? What are the differences of techniques of forehand drive between tennis and soft-tennis? Thus, the purpose of this study is to investigate the upper limb movement during forehand drive playing in tennis and soft-tennis and to compare their differences. Hopefully, the results would provide correct swing techniques for the reference of tennis players and improve the technique of forehand drive.

METHODS

Six tennis players and six soft-tennis players were recruited for this study. An Expert Vision motion analysis system with six video cameras (Motion Analysis Corp., Santa Rosa, CA, USA) was used to collect the images of sixteen retroreflective markers on the upper extremities and trunk movement, while the subjects took close stance to do topspin forehand drive. Ten trials were sampled for each subjects at 60 Hz. Sixteen markers included one each on processus xiphoideus, incisura jugularis, 7th cervical vertebra, acromion, medialis and lateralis epicondylus, ulnar and radial styloid process, 2nd and 4th metacarpal heads, two anterior superior iliac spine and posterior superior iliac spine, and a triad markers on the upper limb. The three-dimensional joint angles and their angular velocities were computed from the trajectories of landmarks placed on the upper extremity and trunk. A nonparametric statistical method, Mann-Whitney U-test, was used to analyze the differences of angular arc, minimum and maximum angles of the wrist, elbow, and shoulder joints between tennis and soft-tennis players.

RESULTS

The movement patterns of the shoulder and elbow joints, and the trunk in forehand drive for tennis and soft-tennis were similar. For the range of joint motion, there existed significant differences between soft-tennis and tennis in the flexion/extension angles of shoulder joint and elbow joint. Compared with tennis players, soft-tennis players had smaller shoulder joint flexion/extension ($p = 0.001$) and larger elbow joint flexion/extension ($p < 0.001$). The angular velocities of the elbow and wrist joints were greater for soft-tennis than tennis. For the wrist, soft-tennis players had significant angular movement changes in terminal forward swing and impact compared to tennis players (Figure 1). It extended pronounced in terminal forward swing and flexed rapidly at impact.

DISCUSSION

For soft-tennis, the flexion velocity of the shoulder and velocities of extension and internal rotation of the trunk reached the maximum values at the moment of impact. The momentum was increased and transferred to the shoulder and trunk from the beginning of the forehand drive and reached the peak value in the forward direction at the moment of impact. To the contrast, the velocities of extension and internal rotation of the trunk reached the maximum values after the impact for tennis. We concluded that the elbow flexion plays a very important role to generate momentum for soft-tennis. For tennis, the internal rotation of the forearm is the key function to control the ball.

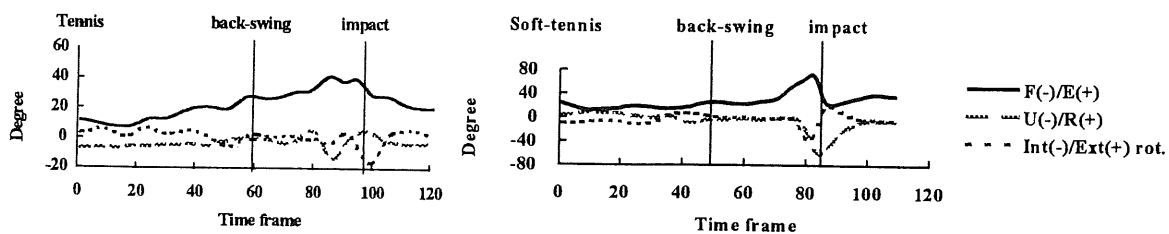


Figure 1: The angular movement of wrist joint

ACKNOWLEDGEMENT: National Science Council, TAIWAN, ROC, NSC85-2331-B-006-074 M08.

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MICRO-FRACTURE SIMULATION IN INDIAN OLYMPIC FIELD HOCKEY PLAYERS

H. S. Ranu, M. S. Hasan*, and J. E. Lander

Life College, School of Graduate Studies, Marietta, GA 30060.

*Indiana State University, School of Technology, Terre Haute, IN 47809.

INTRODUCTION

Injuries due to jumping or falling are common among the participants in sports, such as basketball or skiing. It has been reported that 20% of all sports related injuries are due to jumping in games. Injury secondary to falls results in approximately 18,000 deaths in the USA. Thus, the necessity to provide quantitative and qualitative analysis of impact behaviour of ankles and knees due to jumping is strongly urged. The response of fracture or injury due to jumping is also important in designing of safety equipment related to sports. A linear elastic fracture mechanics approach has been used for the analysis.

METHODS

A standing jump was made by two Indian field hockey players on a force plate under standardized conditions. They weighed 757N, 735N and 182 cm, 180 cm in height. The vertical ground reaction force for the study ranged from 3.7-3.9 times the body weight. Number of cycles to cause the failure of bone due to the repetitive number of jumps were obtained by using the following equation:

$$N_f = \int_{a_i}^{a_f} \frac{da}{C \Delta K^m} \quad (1) \text{ where, } N_f \text{ is the number of cycles of loading, } a \text{ is the crack length,}$$

ΔK is the cyclic range of stress intensity factor, and C and m are constants. where a is the crack length, N_f is the number of cycles of loading. K is the cyclic range of stress intensity factor, and C and m are the constants. Using $a_i = 10.0 \times 10^{-8} \text{ m}$, $C = 1.0 \times 10^{-8} \text{ m} = 1.25$, and $\Delta K = 1.12 \text{ (F/A)}$, equation (1) is integrated numerically with an increment of $10 \times 10^{-6} \text{ m}$ to the final value of the crack length (a_f) $10 \times 10^{-6} \text{ m}$. The results are then used to obtain the rate of change of crack length with respect to the number of cycles (da/dN).

RESULTS AND CONCLUSIONS

Typical behaviour of crack growth rate characteristics for a force of $F = 2897 \text{ N}$ and m being 1.25, reveals that 11 jumps are sufficient to cause micro-fracture of bone. This study also indicates that the number of jumps for micro-fracture of human bone decreases as the value of m increases. It shows that shoes, and other supports can also reduce the injury or propagation of bone crack. The study further reveals that micro-fracture due to jumping is sensitive to number of jumps. This analysis may further help in understanding of fracture fixation of bone. Model is being extended to evaluate accidents in other areas where vertical falls are involved, e. g. parachuting.

SIMULATION OF MICRO-FRACTURE IN THE HUMAN VERTEBRAL BONE DURING WEIGHT LIFTING

J.E. Lander*, H. S. Ranu*, and M.S. Hasan

*Life College, School of Graduate Studies, Marietta, GA 30060. Indiana State University, School of Technology, Terre Haute, IN 47809.

INTRODUCTION

Human lumbar vertebrae are a heavily loaded component of the human body and are subjected to repetitive loading in daily activities. It has been estimated that there are over 500,000 vertebral fractures and over 250,000 hip fractures each year in the USA alone and the associated annual cost exceeds \$ 10 billion. In the past several years, a substantial number of studies have been conducted on human vertebrae. Results from experimental and clinical studies provide considerable variation and it is often very difficult to interpret the data. However, only limited information on failure mechanism of lumbar vertebrae are available to date. Thus, there is a need to develop an analytical model to predict the fracture characteristics of the vertebral body.

METHODS

A linear elastic fracture mechanics approach has been considered and a mathematical model has been proposed so that the predictions can be made more easily related to the occurrence of the injury. Number of repetitive loading necessary to cause the fracture has been obtained by using the following equation:

$$N_f = \int \frac{a_i}{C \Delta K^m} da \quad (1) \text{ where, } N_f \text{ is the number of cycles of loading, } a \text{ is the crack length, } \Delta K \text{ is}$$

the cyclic range of stress intensity factor, and C and m are constants. The resultant compressive force applied in extreme weight lifting was derived from Lander et al. In this study it is reasonably accurate to use $\Delta K = 1.12 \sqrt{F_c/A}$, where F_c is the total resultant applied load on the vertebrae and A is the average cross-sectional area of lumbar spinal disc. Using $a_i = 10.0 \times 10^{-8} \text{ m}$, $C = 1.0 \times 10^{-8}$, $m = 3$, $A = 465 \times 10^{-4} \text{ m}^2$ and the final value of crack length (a_f) $10 \times 10^{-6} \text{ m}$, equation (1) is numerically integrated with an increment of $10 \times 10^{-6} \text{ m}$, the number of repetitive loading (N_f) to cause fracture of vertebrae have been calculated.

RESULTS AND CONCLUSIONS

The study reveals that for a person weighing 1334 N and lifting a weight of 345 kg during squat exercise causes vertebral micro-fracture at 12 repetitive standing lifting. While the same load at lowest position (hip angle = 49.9 degrees) yields micro-fracture at less than 3 lifting. The numerical study also shows that for the change of position from standing to the lowest position the resultant compressive force acting on the spine increases by two times whereas the possibility of micro-fracture increases by five times. Similarly at deadlift exercise, lifting 325 kg from standing to lowest position increases the resultant compressive forces on the vertebrae by 2.5 times. This however, increases the micro-fracture ratio by six times. Hence, a similarity of micro-fracture characteristics for different exercises have been observed. The model presented here is being extended to include a complex geometry of the spinal disc, force and torque will be applied to simulate the micro-fracture of the vertebrae.

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THREE DIMENSIONAL NUMERICAL MODELING AND OPTIMIZATION OF AN INTERVERTEBRAL DISC PROSTHESIS

A. TEMPLIER*, W. SKALLI*, J.P. LEMAIRE†, F. MENDES, A. DIOP*, F. LAVASTE*

*Laboratoire de Biomécanique, ENSAM, Paris, France

†Point Médical, Dijon, France.

INTRODUCTION

Difficulties of in vitro experiments for implant evaluation are well known : human cadaver specimens supply, dispersion among individuals, experimental setup and protocols. With regard to that, numerical simulation based on the Finite Element Method becomes a key tool necessary to complement experiments. The experimental approach often consists in a kinematic study of the implant behavior, whereas numerical modeling allows calculation of intra-articular loads and stresses both in organic and artificial materials. The present study combines these two approaches, through the biomechanical analysis and the optimization of the SB Charité intervertebral disc prosthesis.

METHODS

Six fresh human cadaver L3-L4 specimens were used. Age was comprised between 35 and 71 years (mean : 57 y.). Specimens were frozen at -30° C until use. A specific vertebral testing device was used (Lavaste *et al.*, 1990), the lower vertebra was fixed while the upper one was loaded using masses and pulleys to apply flexion, extension, lateral bending and torsion moments up to 10 Nm. In each case, load-displacement curves of the upper vertebra were recorded. Specimens were tested first intact, then instrumented using artificial discs. An existing tridimensional L3-L4 spinal segment model (Lavaste, Skalli, Robin *et al.*, 1992-1993), based on the ANSYS® Finite Element Method software was used. Simulations were performed in flexion and torsion on intact and instrumented segments. The insertion of the prosthesis was simulated by removing the disc except the posterior part, and by replacing it using a model of the prosthesis (fig.1), taking into account contact and sliding between components. The numerical results were first compared to experimental ones. Then the model was used to obtain complementary data, such as posterior intra-articular loads. Eventually, modifications of the implant were considered and their effects were analyzed using the model.

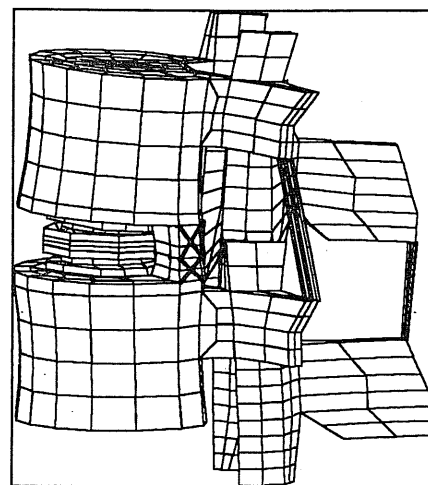


figure 1: instrumented segment model

RESULTS

Experimentally, mean values of main rotation at 10 Nm on intact segments were 6.7° and 2.3° respectively in flexion and torsion. In flexion, the discal replacement yielded a load-displacement curve which had the same aspect as the intact one, with an mean increase of the amplitude of +30%. This increase was much larger in torsion: +217%, with also an increase of sagittal rotation coupling. Numerical results are of the same order of magnitude as experimental ones for intact segments and instrumented ones. Complementary data yielded by the model indicated a large increase of posterior articular facet loading in torsion : +147%. Simulations of alternative designs highlighted improved solutions yielding more physiological ranges of motion, and intra-articular loading. The articular load increase was lowered to +81%. Moreover, synthesis between experimental and numerical results allowed to clarify interactions between the intervertebral disc stiffnesses and the posterior articular loading.

DISCUSSION AND CONCLUSION

The biomechanical evaluation of an implant through such in vitro experiments and numerical simulations only gives an rough estimate of how the implant behaves in vivo, since we still do not master the exact role of the musculature to regulate and control the mobility of each vertebral segment. However, such in vitro experiments allow to better understand and to objectivize the implant behavior for elementary load cases. The finite element model, not only complements very usefully experimental results, but also constitutes a powerful tool to test various modifications without increasing the need for experiments. The combination of both approaches constitutes the way of the future for implants biomechanical evaluation.

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ANI-MAN3D : A 3D MOTION ANALYSIS SYSTEM USING A NUMERICAL HUMAN MODEL. APPLICATION TO CROSS COUNTRY SKIING.

M. TAVERNIER, P. COSSERAT, S. EMMENDOERFFER, A. RUBY. FFS, Cellule Recherche, Annecy (France).

J.P. VERRIEST, INRETS LESCO, Bron (France).

J.M. LAVEST, M. DHOME, Université Blaise Pascal, LASMEA, Clermont-Ferrand (France).

The performance analysis in cross country skiing during high level international sport events needs the use of non invasive method such as video 3D motion analysis without the possibility of setting some active or passive markers on the sportsmen. Up to now, the precision of 3D analysis systems using two video recorders (VCRs) allows to discriminate some behavior differences between expert and non expert performers. In order to display some characteristic differences between high level ski racers, the video motion analysis system must undergo some improvements dealing with : picture resolution, space calibration and morphological human model.

The use of two synchronized Beta SP VCRs provides the best possible resolution today (572 lines). The video sequences (25 images per second or 50 fields per second) are digitized without distortion on a PC as a series of bitmap images by a PVR card from Silicon Co. Thus, the two images recorded simultaneously by both VCRs can be displayed and analysed in parallel on the screen of a regular PC. This method of display, which is inevitable for ANI-MAN3D space reconstruction method, allows to reduce the uncertainty on the estimated positions of some particular hidden points (end of segments, joints) when a classical 3D motion analysis method is used. ANI-MAN3D uses the numerical human model MAN3D (Verriest, 1991) the position and posture of which can be adjusted. Then, MAN3D is projected on both synchronized images. When the projections of MAN3D are superposed to the images of the skier, the position and posture of the skier are the same as that of MAN3D. In order to get the best possible precision, the morphological properties of MAN3D are deduced from measurements on the subject. Posture and position adjustments are optimized by locating on both synchronized images the 3D positions of some particular points of the skier which must coincide with some equivalent points of MAN3D (Verriest, 1997). This reconstruction method requests a precise space calibration. The calibration method must give the parameters of VCRs (positions, directions of optical axes and focal lengths) and the numerical compression factors which are not provided by the DLT method. The algorithm used for space calibration is based on a photogrammetric method (Li, Lavest, 1996). This method needs a reasonable experimental estimation of the camera's parameters and a very accurate measurement of the positions, in space and on screen, of some fixed markers. In order to get a precise space calibration a dozen or more of fixed markers is required, the positions of which must be determined with a very good accuracy (theodolite) and must fill the greatest part of the common space seen by both VCRs. MAN3D can be equipped with some accessories such as skis or ski sticks, the positions of which can be determined by constraints with respect to some particular points of MAN3D (Figure 1)

The improvements provided by ANI-MAN3D motion analysis system are the following:

- the space calibration is very accurate
- the measurement of the subject position realized with the help of the outlines of his projections is more accurate than with the use of 10 or 20 particular points
- the measured lengths of segments (legs, arms) are necessarily constant
- the inertial properties of the subject (center of mass, moments of inertia) are better defined.

These improvements are particularly useful for the studies of the factorial analysis of motions (A. Ruby, 1995) and for the estimation of the mechanical energy produced during the motion.

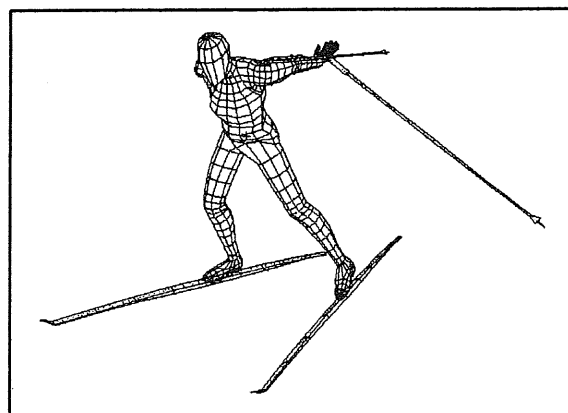


Figure 1 : Plane projection of the volumic human model MAN3D, equipped with skis and ski sticks

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THREE DIMENSIONAL SIMULATION OF RUNNING

I.C. Wright, R.R. Neptune, A.J. van den Bogert
Human Performance Laboratory, The University of Calgary, Alberta, Canada

INTRODUCTION

Running is a vastly popular activity, and is an activity central to many sports. However, injuries due to running are common (van Mechelen, 1992). Forces on and in the leg during running, and the mechanisms by which they may be related to these injuries are not well understood.

The determination of internal forces during running from kinematic data is made difficult by inaccuracies of inverse dynamic methods. Simulations have been used to investigate the kinetics of running, and have contributed to the understanding of some mechanisms involved in loading the leg during running (Cole *et al.*, 1996). Most simulation models have considered only movement in the sagittal and frontal plane. However, during running, movements of the foot and ankle, muscles geometries, and forces at the joints all have three dimensional components. Therefore, the goal of this study was to develop a model capable of simulating running that includes three dimensional movements and forces.

METHODS

A three dimensional model of the lower extremity was developed from the geometrical data of Delp (1990), and inertial properties from literature (Figure 1). The model was dimensioned to represent a 180 cm tall, 75 kg person. The model had rigid segments representing the foot, talus, shank, patella, and thigh of the support leg, and a rest-of-body segment. The joints of the ankle had their axes aligned as described in Delp (1990). The knee joint had coupled translation and rotation in the sagittal plane, as well as limited ad/abduction, and internal/external rotation. Patellar kinematics were coupled to knee flexion. The hip was modelled as a ball and socket joint. Internal/external rotation and ab/adduction of the hip and the knee were limited by passive joint torque elements.

The contact between the foot and the ground was represented by 66 contact points distributed across the underside of the foot. Simulations were performed using the DADS software (CADSI, Coralville, IA)

Position, orientation, linear and angular

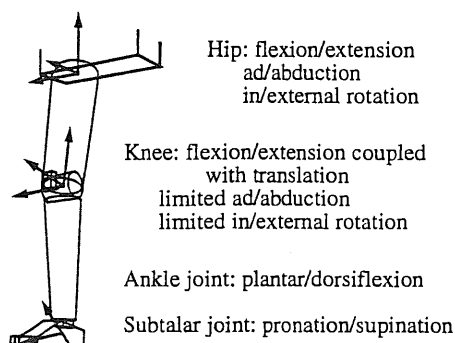


Figure 1: Joints of the model and their movements.

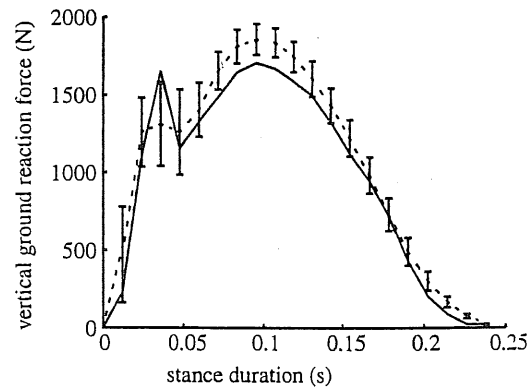


Figure 2: Vertical ground reaction force profiles for the average of ten experimental trials of one subject (dotted line, with 2 standard deviation bars) and a simulated trial (solid line).

velocities of the limb segments were recorded using video motion analysis of a subject performing ten trials of heel-toe running. Vertical ground reaction forces were recorded simultaneously. Positions and velocities of the body segments at heel strike were used as the initial conditions for the simulation.

Twelve functionally independent muscle groups were included in the model, the paths of which were described by limb segment fixed origins and insertions, as described by Delp (1990). The force-length-velocity characteristics of the muscles were represented by Hill-based, lumped parameter model. Muscle stimulations patterns were determined by minimizing the differences between the simulated and measured averaged joint angle and ground reaction force profiles.

RESULTS AND DISCUSSION

The joint angle and ground reaction force profiles generated by the simulations were reasonably close to those of the real movement (Figure 2).

The model developed here can be used to examine three dimensional joint forces during movement, including in/external rotation and ad/abduction moments, and medio-lateral forces. Also, loads caused by movements out of the sagittal plane can be examined. Therefore, this model can be used to investigate how numerous factors are related to joint loading, such as variations in muscle strength and coordination, gait kinematics, skeletal geometry, anthropometrics and footwear properties.

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OPTIMIZATION OF FLIGHT TRAJECTORY OF SKI JUMP ON THE BASIS OF WIND TUNNEL DATA

K. Yoshida, M. Murakami, I. Watanabe*, T. Ohtsuka and S. Kazama

Inst. of Engineering Mechanics, Univ. of Tsukuba, Ibaraki, Japan

*Research Center for Advanced Sci. and Tech., Univ. of Tokyo, Tokyo, Japan

INTRODUCTION

This paper describes the optimization study on the flight trajectory of ski jump on the basis of wind tunnel test data with the aid of the optimal control theory. We are primarily interested in the analysis of a flight phase among four phases of ski jump, approach, take-off, flight and landing phases. In the optimization such parameters as the body-ski angle and the ski-ski (V-style) angle are controlled for the maximum jump length. The aerodynamic data required for the calculation are obtained from a full-size model experiment in the 3-meter low speed wind tunnel at Research Center for Advanced Science and Technology, Univ. of Tokyo.

METHODS

The mathematical model of a ski jump in a flight phase is based on the equations both in translational and rotational motions of a jumper-ski combination. The equations of motion supplemented with the aerodynamic data, that is the drag area, the lift area and the moment volume, are solved numerically for the optimal solution. The angle of attack and the body-ski angle and so on are selected as the parameters representing the position of a jumper-ski combination. The optimal solution is numerically obtained by using the method of conjugate gradient algorithm derived on the basis of optimal control theory. In the optimization procedure, the jump length as a performance index is maximized under the constraint that the body-ski angle should be positive.

RESULTS AND DISCUSSION

A maximum jump length solution is given for each set of initial conditions which lead to a safety landing. The optimal solution is as follows: The solution starts with a small angle of attack, a small nose-up attitude relative to the flight trajectory, so that the drag should be small. Then a jumper body gradually rotates due to negative pitching (nose-down) moment while maximizing the lift, finally leading to a very small body-ski angle with a ski kept in a nearly horizontal position at the landing. It is the point at the final stage before landing to keep a ski horizontal for maximum lift. It should be noted that only a little increase in the ski angle (nose-up) from a horizontal position readily falls into stall resulting in drastic increase in the drag and reduction in the lift because a horizontal ski position implies a very large angle of attack (about 45°) at the final stage. It is also of interest that the optimal solution is far from the maximum L/D solution for the most part of flight[1]. It is found that the optimal solution fairly well agrees with typical jump trajectories of ranking jumpers performed in recent large-hill competitions.

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A MODEL FOR DETERMINING OPTIMAL SKI SKATING TECHNIQUES

Anton J. van den Bogert[†] and Todd L. Allinger^{†‡}

[†]Human Performance Laboratory and [‡]National Sport Centre Calgary
The University of Calgary, Alberta, Canada

INTRODUCTION

Skating techniques have led to dramatic increase in performance in cross-country skiing during the past two decades. One important aspect of skating technique is the ski placement angle². Skiers tend to choose larger angles between skis and direction of travel when going uphill. This suggests that for each condition there is an optimal angle, and that a poor choice of angle may lead to suboptimal performance. Hence, it is the purpose of this study to examine theoretically the relationship between ski placement angle and performance under different conditions.

METHODS

The theoretical model is an adaptation of a simulation model for speed skating¹ (Fig. 1). The skier is represented by a point mass supported on a massless leg. Equations of motion are based on the assumptions that the height of the centre of mass z and the ski orientation θ remain constant during the stroke. This leads to the following set of three differential equations for a stroke with the right ski:

$$\begin{aligned} m\ddot{x} &= -\frac{mg}{z}(L_h \sin \theta + \gamma L \cos \theta) - kv\dot{x} + \frac{P_p \dot{x}}{v^2} \\ m\ddot{y} &= \frac{mg}{z}(L_h \cos \theta - \gamma L \sin \theta) - kv\dot{y} + \frac{P_p \dot{y}}{v^2} \\ \dot{L}_h &= v \sin |\alpha - \theta| \end{aligned}$$

where γ is the coefficient of snow friction, k is the coefficient of aerodynamic drag, and P_p is the power generated by the poling action. Forces due to an incline are easily added to the equations. A similar equation can be derived for the left stroke. This system of equations was numerically simulated, alternating right and left strokes, until steady state was reached. Velocity was then determined from distance and time over one cycle (two strokes). A stroke ended when one of the following constraints was violated: length ($L \leq L_{\max}$), strength (force-velocity: $F \leq f_{\max}(\dot{L})$)³ and lower extremity aerobic power ($\frac{1}{T} \int F dL \leq P_{\max}$).

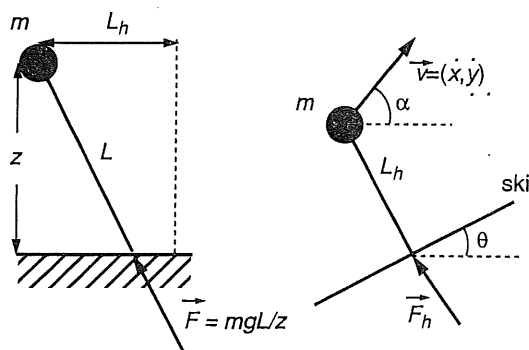


Fig. 1: The model in the frontal plane of the skier and in the horizontal plane.

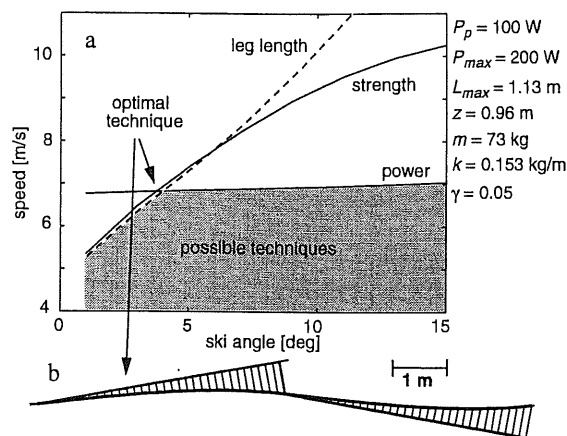


Fig. 2. (a): Relationship between technique and speed. Ski angle is defined as the angle $\alpha - \theta$ between velocity and ski at start of the stroke. (b): Optimal CM and ski trajectories.

RESULTS AND DISCUSSION

Fig. 2a shows how each of the three constraints limits performance as a function of technique, for one set of conditions. At low ski angles, the leg length constraint is reached before maximal aerobic power, leading to a significantly submaximal speed. At higher ski angles, all available aerobic power can be utilized, resulting in a small increase in speed as ski angle increases, but the stroke duration decreased dramatically. This would increase the energy required for the recovery phase, which was not considered in the model. The optimal ski angle therefore is the lowest angle where all aerobic power is used, while also satisfying the other constraints (Fig. 2). Skiers with inconsistent technique should choose a somewhat larger than optimal ski angle to ensure that all available aerobic power is used, at the expense of a higher energy cost for the recovery phase. For the conditions used in the example (Fig. 2), predicted optimal technique (ski angle, stroke duration) and speed agreed well with observations on elite skiers².

Optimal ski placement angle, as defined above, depends on the conditions (mainly snow friction and incline), as well as on the characteristics of the skier ($L_{\max}$, $f_{\max}$, and $P_{\max}$). As expected, optimal ski angle increased with increasing snow friction or increasing incline. The model can be used to help understand which changes in technique are required when conditions and/or skier characteristics change.

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MECHANICAL AND FUNCTIONAL PROPERTIES OF CAT KNEE ARTICULAR CARTILAGE 16 WEEKS POST ACL TRANSECTION

W. Herzog, J.Z. Wu, T.R. Leonard, E. Suter, S. Diet, C. Müller, P. Mayzus

The University of Calgary, Calgary, Alberta, Canada

INTRODUCTION

In the initial stages of osteoarthritis, articular cartilage becomes thicker, softer, and has a higher permeability (1). The effect of these changes in articular cartilage properties on the joint contact mechanics is largely unknown. The purpose of this study was to predict the changes in the contact mechanics of joints in the initial adaptive stages of osteoarthritis.

METHODS

Adaptations of the articular cartilage were produced by transecting the anterior cruciate ligament (ACL) in the left knee of five cats. Predictions about the changes in the contact mechanics of the experimental knee were performed using a biphasic model of articular cartilage (2). The contact mechanics of the experimental and contralateral knees were determined by measuring the *in situ* pressure distributions in the patellofemoral joints 16 weeks following ACL transection. For this purpose low grade and medium grade pressure sensitive film (Fuji) was inserted into the patellofemoral joints at 8 angles and producing 8 levels of quadriceps force. The animals were sacrificed and articular cartilage thickness and material properties were determined (3)

RESULTS

Cartilage from the experimental knee was softer and thicker than the cartilage from the corresponding contralateral knee (Fig. 1). From these results, we predicted that the joint contact areas should be larger and the peak contact pressures should be smaller in the ACL-transected compared to the contralateral limbs. This prediction was verified in all five animals (Fig. 2).

DISCUSSION

At 16 weeks following ACL transection, the articular cartilage becomes thicker and softer. As a consequence, joint contact areas become larger and peak pressures become smaller in the ACL-transected compared to the contralateral knee. From these results, it appears that early osteoarthritis is associated with an unloading of normally loaded contact areas and an overloading of normally unloaded contact areas.

ACKNOWLEDGEMENTS

This work was supported by: MRC Canada, The Arthritis Society of Canada.

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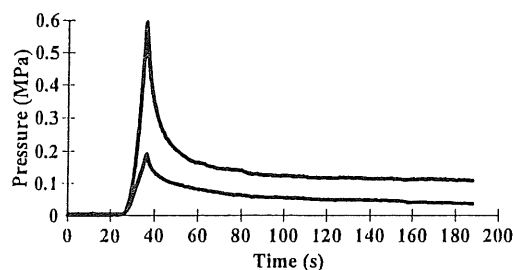


Figure 1. Representative force (pressure) relaxation curves for cartilage from the lateral femoral condyle 16 weeks following ACL transection (lower curve) and the corresponding contralateral knee (upper curve). Compression was 30 μ m in both cases. Cartilage thicknesses were 600 and 300 μ m in the sample from the transected and contralateral knee, respectively.

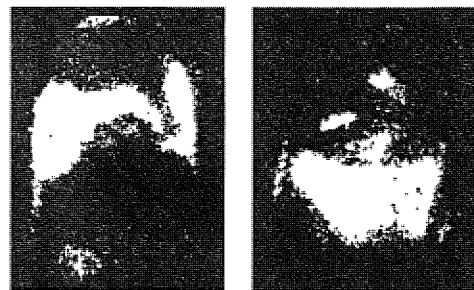


Figure 2. Contact areas from the patellofemoral joint 16 weeks post ACL transection (right) and the corresponding knee (left). The contact areas from the experimental limbs were always larger (mean = $44\% \pm 48\%$) than those from the contralateral limbs for given knee angles and loading conditions ($n=18$).

EFFECTS OF AGE ON STRESS-STRAIN AND STRESS-RELAXATION BEHAVIOURS OF THE RAT INCISOR PERIODONTAL LIGAMENT

K. Komatsu, Y. Watanabe, M. Kanazashi* and M. Chiba

Departments of Pharmacology and *Periodontology, School of Dental Medicine, Tsurumi University, Yokohama, Japan

INTRODUCTION

Previous studies have shown that age-related changes in the shear stress-strain relations of the periodontal ligament (PDL) are minor in the middle region of the rat incisor (Yamane *et al.*, 1990). However, there seem to be few studies on the effect of age on the stress-relaxation behaviour of the PDL. We examined the stress-relaxation behaviour as well as the shear stress-strain relations of the PDL in three different regions of the mandibular incisor of rats from 2 to 24 months of age.

METHODS

Four groups of ten male Wistar rats were killed by an overdose of ether at 2, 6, 12 and 24 months of age. The left and right mandibles of each rat were dissected free and transverse sections (0.65 mm in thickness) were cut from the incisal, middle and basal regions of each mandibular incisor. Using sections from one incisor of each pair, a shear stress-strain curve for the PDL was obtained with a tensiometer (Mandel *et al.*, 1986; Chiba and Komatsu, 1993). From the stress-strain curve, the maximum shear stress, maximum shear strain, tangent modulus and failure strain energy density were estimated. Each section from the other incisor was loaded up to 50 % of the maximum stress determined from the contralateral section, and with constant strain for 10 min, a stress-relaxation curve for the PDL was obtained. Normalized relaxation shear modulus $Gr(t)$ with respect to the initial value was then calculated, and analysed using a technique of exponential curve fitting (Sasaki and Enyo, 1995).

RESULTS

The maximum stress, maximum strain and failure strain energy density in the incisal region significantly increased from 2 to 24 months of age ($p < 0.01$ or 0.001), but those in the middle and basal regions did not change markedly. The tangent moduli in the three regions did not change significantly from 2 to 24 months. Modulus $Gr(t)$ for each region at each age was found to be expressed by the following equation:

$$Gr(t) = A_1 \exp(-t/\tau_1) + A_2 \exp(-t/\tau_2) + A_3 \exp(-t/\tau_3)$$

where A_i is the fraction of each component, and τ_i is the relaxation time (s) of each component. The relaxation times of the fast and medium components in the three regions did not show marked age-related changes, but those of the slow component in the incisal and basal regions increased significantly from 2 to 24 months ($p < 0.001$).

DISCUSSION

Our results indicate that the mechanical strength, extensibility and toughness of the PDL increase from 2 to 24 months of age in the incisal region of the rat mandibular incisor. Furthermore, the increased relaxation times in the incisal and basal regions indicate an increase in the viscous property of the PDL. The major role of collagen fibres in mechanical responses of the PDL has already been suggested (Chiba and Komatsu, 1993; Komatsu and Viidik, 1996). In addition, the viscous property may be related to the interfibrillar substances with their hydrophilic proteoglycans in rat tail tendons (Reihnsner and Menzel, 1994) or to water content in bovine bone (Sasaki and Enyo, 1995). Therefore, it is likely that changes occur not only in the collagen fibres but also in the interfibrillar substances or water content in the PDL during aging.

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CHANGES IN PERIARTICULAR BONE MINERAL DENSITY FOLLOWING ANTERIOR CRUCIATE LIGAMENT TRANSECTION

GR Wohl, RC Chan, R Kloiber, ME Adams, JR Matyas, RF Zernicke.

McCaig Centre for Joint Injury and Arthritis Research, University of Calgary, Calgary, Alberta, Canada.

INTRODUCTION

Rupture of the anterior cruciate ligament (ACL) alters joint kinematics¹ and joint loading,² and contributes to the development of osteoarthritis³ (OA). Since end stage OA is characterized by loss of articular cartilage, OA research has emphasized the soft tissues of the knee joint. Nevertheless, changes in both joint kinematics and loading may cause connective tissue changes that affect the entire limb. In particular, the altered loading environment may produce changes in periarticular cancellous bone. To investigate the potential temporal OA-related changes that occur in cancellous bone, at the hip and knee joints, we studied animals at several time points after anterior cruciate ligament transection (ACLX). We hypothesized that within the first 32 weeks after ACLX, ubiquitous decrements would occur in bone mineral density in the affected limb. Our results indicate that significant and deleterious cancellous bone changes occur in early experimental OA, in addition to the commonly reported hypertrophic and degenerative soft tissue changes.

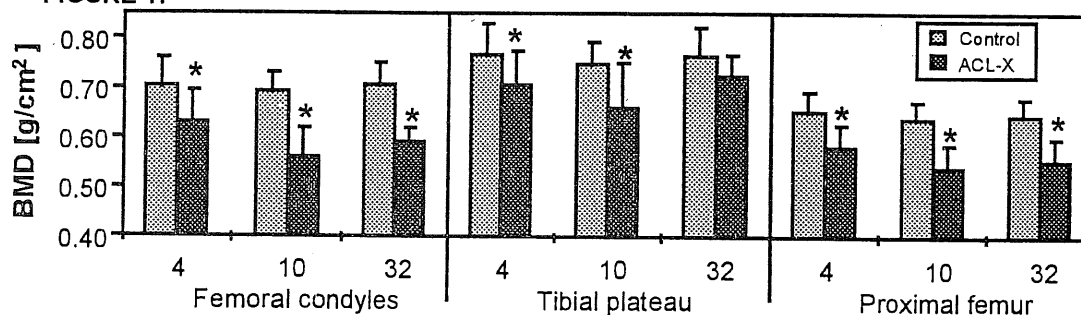
METHODS

Skeletally mature, mixed-breed dogs (n=18) were assigned randomly to three experimental groups: 4, 10, and 32 wk (time periods following ACLX). With the approval of the Animal Care Committee, each dog had unilateral ACLX; the non-operated contralateral limb served as a control. Following euthanasia, the proximal femur, femoral condyles, and tibial plateau were scanned using dual energy x-ray absorptiometry (Hologic 2000). Bone mineral density (BMD) was calculated for three regions: proximal femur, femoral condyles, and tibial plateau. BMD of the ACLX and control joints was compared within each dog, and statistical significance ($p < 0.05$) at 4, 10, and 32 wk was determined using paired Student's t-tests.

RESULTS

Significant differences between ACLX and control limb BMD were found in all periarticular regions we examined. The ACLX proximal femur and femoral condyles had significantly lower BMD at all time periods (4, 10, and 32 wk) (Fig. 1). ACLX BMD in the femoral condyles went from 90% of control at 4 wk, to 81% at 10 wk, and to 84% at 32 wk. Similarly, in the proximal femur, ACLX BMD went from 89% of control at 4 wk, to 85% at 10 wk, and 86% at 32 wk. In the tibial plateau BMD was significantly less than control at 4 (92% of control) and 10 wk (88%), but not at 32 wk (95%).

FIGURE 1:



DISCUSSION

As anticipated, we found significant changes in periarticular BMD that progressed through early stages (up to 32 wk) in this model of experimental osteoarthritis. In comparison to the non-operated contralateral limb, significant reductions in bone mineral density were found in the cancellous bone of the femoral condyles, proximal femur, and tibial plateau of ACLX limbs. Our data suggest that changes in the quality of cancellous bone occur in early OA and that the significance of these subchondral bone changes must be assessed in more detail to understand better the progression of post-traumatic OA.

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ACKNOWLEDGMENTS: Funded, in part, by The Arthritis Society of Canada and the Alberta Heritage Foundation for Medical Research.

ANALYSIS OF STRAIN DISTRIBUTION IN LIGAMENTS USING PHOTOELASTIC COATING METHOD: THE INFLUENCE OF PHOTOELASTIC MATERIAL ON THE MECHANICAL PROPERTIES OF LIGAMENT.

Takashi Kawada, Takanobu Abe, Akio Inoue, Kouji Yamamoto[†] and Shunji Hirokawa^{††}

Department of Orthopaedic Surgery, School of Medicine, Kurume University,

[†] Department of Mechanical Engineering, Faculty of Engineering, Kurume Institute of Technology

^{††} Department of Graphic Science, Faculty of Engineering, Kyushu University,

Introduction

Strains on the surface of ligaments have been determined mainly by following the displacement of markers in tensile testing with a video with subsequent measurement (the VDA system). Measurable strains are limited to a marked range. In order to measure strains in the entire ligament area, we prepared a film adhesive to ligaments, examined strains using a photoelastic coating method and attempted to analyze stress on the basis of the strains. In this study we report the effects of coating on the mechanical properties of ligaments, and measurements of strains in uniaxial tensile testing.

Materials and Methods

This study was comprised of two experiments. First we prepared pieces of fascia lata removed from eight human cadaveric knees (donor age 45 to 70 years) and cut in the form of test pieces, and 1 mm thick eight polyurethane film (NIPPORAN 5120, JAPAN POLYURETHANE Co.Ltd.) pieces of the same shape. We performed tensile tests individually, and performed further tensile tests with a ligament to which the polyurethane film was adhered to and then compared the mechanical properties of these test pieces. In the second experiment, ten skeletally mature male Japanese White Rabbits (body weight = 3.0 ± 0.2 Kg, mean $\pm$ S.D.) were used. To prepare films, we applied polyurethane coating to thickness of 0.1 mm on the entire MCL surface area in rabbit's femur-MCL-tibia complex (FMTC). We performed uniaxial tensile testing (20mm/min) using these films and analyzed the measurements of strains from the insertion of the entire ligament area by the photoelastic coating method (Fig.1). The result was obtained by fringe distribution analyzed from isochromatics (Fig.2a). Animal treatment in this study was fully approved by our institutional Animal Care and Use Committee.

Results

Polyurethane, which has the properties of high strain sensitivity, adhered well to ligaments and could sufficiently follow the strain behavior of ligaments. The Young's modulus of polyurethane film is about 1/30 of that of ligaments, and a film 0.1 mm in thickness hardly affected them (strengthening phenomena). Continuous strain-measurement was possible over the entire ligament area in uniaxial tensile testing (20 mm/min), and the insertion of the ligaments showed a strain larger than that of the midsubstance (Fig.2b).

Discussion

In studies using the conventional method, measurement of strain and stress was possible only in the direction of uniaxial and, in actual measurement of strain, it was difficult to measure the strain 3-dimensionally in all directions. The photoelastic coating method allows continuous strain-grasping over the entire ligament area, which is not achieved by other measuring methods that use a strain gauge, and therefore allows quantitative determination of strains. Therefore, it is suggested that the method may become an effective measurement to investigate where ligament strains have concentrated. By establishing this method, we can expect supplement in actual limit of the existing methods and contribute to the progress of study in this specific field.

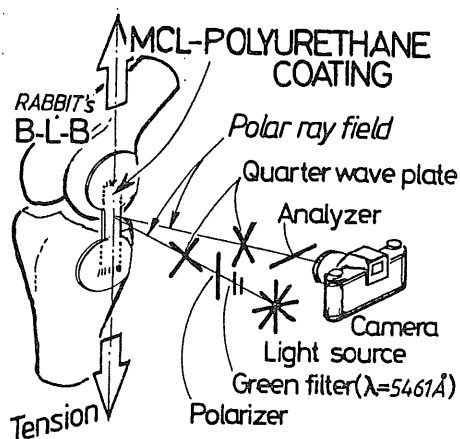


Fig.1 Schematic draw of experimental apparatus.

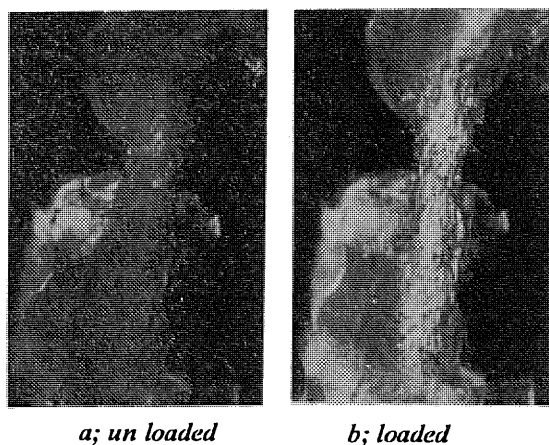


Fig.2 Isochromatics on the MCL surface in FMTC.

ANALYSIS OF STRAIN DISTRIBUTION IN LIGAMENTS USING PHOTOELASTIC COATING METHOD: STUDY ON SIMULATED KNEE MODEL

Kouji Yamamoto, Shunji Hirokawa[†] and Takashi Kawada^{††}

Department of Mechanical Engineering, Faculty of Engineering, Kurume Institute of Technology

[†] Department of Graphic Science, Faculty of Engineering, Kyushu University,

^{††} Department of Orthopaedic Surgery, School of Medicine, Kurume University,

Introduction

Although many studies have been done on the cruciate ligaments' tensile characteristics, the length patterns and differential function of the ligaments, especially those of the anterior cruciate ligament (ACL) remain controversial issues. Yet it is impossible to directly measure, *in situ*, the non-uniform distribution of biaxial strain over the entire surface of the intra-articular ligaments. To avoid these difficulties, we propose an alternative to using actual knee-joints and ligaments: the use of accurate models of the joint and ligaments fabricated in a way which allows deformation and strain distribution to be directly observed and measured. We fabricated and employed such models in a study whose objective was to measure the strain distribution over the entire surface of the model ACL in association with its three-dimensional deformation due to knee flexion.

Materials and Methods

Transparent knee-condyle models were fabricated from epoxy resin: two types of araldite (GY 250 and GY 1252, Chiba Gigy Co.Ltd., Japan) and hardener, and the ligament model from silicone rubber (KE-RTV, Shinetsu Silicone Co.Ltd., Japan). The resulting material of ligament model possessed stress-strain characteristics very similar to those of actual human ligaments. Both bone and ligament models were casted from silicone rubber mould originated by a cadaveric knee (male, 70 years, right). For photoelastic measurement, polyurethane film (NIPPORAN 5120, Nippon Polyurethane Co.Ltd., Japan) was found most suitable, which is capable of higher elongation with lower stiffness, and have optically high fringe-sensitivity. To heighten transparency for photoelastic measurement, the assembled knee model was immersed into the epoxy monomer liquid with the same refractive index as the transparent bone model. The knee model was manipulated by the link mechanism so as to duplicate natural knee motion. Then isochromatics and isoclinics appeared on the ligament were photographed and measured. The experimental apparatus is shown in Fig.1.

Results and Discussion

Typical examples of the results are shown in Fig.2 and 3. The diagram of the principal strain obtained from isoclinics (Fig.2) was close to the fiber direction in the actual ligament. Also the changes in shape of the ligament model demonstrated similar to the diagrams of changes of the actual ligament drawn from the observations. Strain distribution (Fig.3) clearly varied, even along the principal strain lines, and large strain gradients were observed in the areas near the femoral insertion.

In this study, measurement was done using a unique model of a knee joint to overcome the difficulties associated with attempting to measure highly variable deformations of the actual ligament. The photoelastic coating method provided continuous information of the conditions of strains over the whole ligament surface, and provided quantitative measurements of those strains. Even though the material and structure of the model ligament were different from those of a natural ligament, the model technique allowed us to make comparisons with and better understand the results of previous clinical experiments.

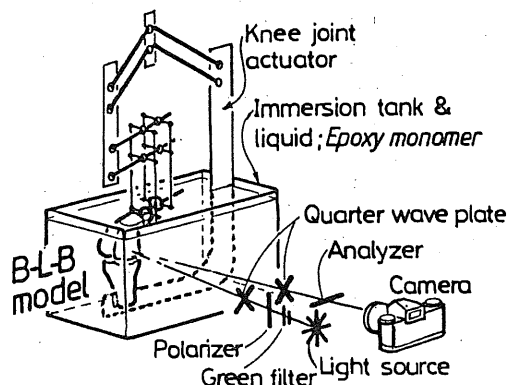


Fig.1 Schematic draw of experimental apparatus.

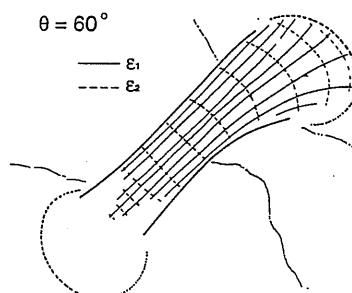


Fig.2 Principle strain diagram

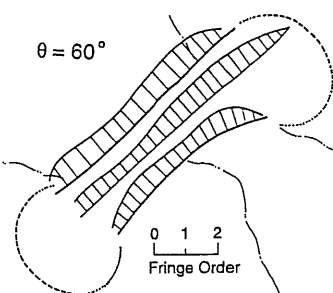


Fig.3 Strain distribution

THREE DIMENSIONAL DEFORMATION AND STRAIN DISTRIBUTION OF THE CRUCIATE LIGAMENT: MODEL ANALYSIS OF FIBER-REINFORCED HYPERELASTIC MATERIAL

Shunji Hirokawa and Reiji Tsuruno[†]

Department of Graphic Science, Faculty of Engineering, Kyushu University,

[†]Department of Visual Communication Design, Kyushu Institute of Design,

Introduction

Highly variable deformations of the anterior cruciate ligament (ACL) cannot be adequately quantified by one-dimensional and/or localized measurements. Mathematical model analysis can be useful tools for the studies of complex deformation and strain distribution of the ACL. This paper describes a new structurally-motivated phenomenological approach.

Constitutive Equation of the Ligament

Such assumption was made that the ACL could be idealized as being composed of a homogeneous matrix in which two families of densely distributed extensible fibers are embedded. The fibers of one family are structurally oriented in parallel with the longitudinal direction, and those of the other are radially arranged in the oblique angle. It was assumed that both the fibers and the matrix follow the gross deformation. Then the ACL was considered be anisotropic continua, and its pseudo strain-energy function was formulated as,

$$W(C) = v_m W_m(C) + v_a W_{fa}(\lambda_a) + v_b W_{fb}(\lambda_b), \quad v_m + v_a + v_b = 1 \quad (1)$$

where C is the Cauchy-Green deformation tensor. W_m , W_{fa} and W_{fb} and v_m , v_a and v_b represent the stored-energy and the volume ratio of the matrix and two family fibers respectively. λ_a and λ_b represent the extension ratio of two family fibers. The matrix was assumed be fabricated from a Mooney-Rivlin material. The Kirchhoff stress S_{ij} is determined as,

$$S_{ij} = 2v_m \{ C_{1ij} + C_2(\delta_{ij} C_{\ell\ell} - C_{ij}) \} + v_a \frac{\partial W_{fa}(\lambda_a)}{\partial \lambda_a} \frac{A_{ij}^{(0)}}{\lambda_a} + \frac{v_b}{8} \sum_{k=1}^8 \frac{\partial W_{fb}(\lambda_{bk})}{\partial \lambda_{bk}} \frac{b_{i,k}^{(0)} b_{j,k}^{(0)}}{\lambda_{bk}} - p \delta_{ij} \quad (2)$$

where C_1 , C_2 : Mooney-Rivlin constants of the matrix, $A_{ij}^{(0)}$, $b_{i,k}^{(0)}$: the initial fiber directions of two families, p : hydrostatic pressure, δ_{ij} : the Kronecker delta.

The exponential stress-strain relation of the fiber was approximated by a tri-linear line; Each family has 3 groups of fiber lengths γ_1 , γ_2 , γ_3 , which become taut at 3 strain levels λ_1 , λ_2 , λ_3 , respectively. Then, $\partial W_f(\lambda)/\partial \lambda \cdot 1/\lambda$ in Eq (2) is give by,

$$\begin{aligned} \frac{\partial W(\lambda)}{\partial \lambda} \frac{1}{\lambda} &= k_s(\lambda-1), & 0 \leq \lambda < \lambda_1 \\ &= [k_s + \gamma_1(k_e - k_s)](\lambda-1) - (k_e - k_s)\gamma_1(\lambda_1-1), & \lambda_1 \leq \lambda < \lambda_2 \\ &= [k_s + (\gamma_1 + \gamma_2)k_e - k_s](\lambda-1) - (k_e - k_s)[\gamma_1(\lambda_1-1) + \gamma_2(\lambda_2-1)], & \lambda_2 \leq \lambda < \lambda_3 \\ &= k_e(\lambda-1) - (k_e - k_s)[\gamma_1(\lambda_1-1) + \gamma_2(\lambda_2-1) + \gamma_3(\lambda_3-1)], & \lambda_3 \leq \lambda \end{aligned} \quad (3)$$

By substituting Eq(3) into Eq(2), the constitutive equation of the ACL model is introduced.

Nonlinear Stiffness Relation

We derive a nonlinear stiffness relation between node forces and displacements from the principle of virtual work. On the assumption that node force and node displacement shift, by infinitesimal amounts, Δu and Δp from their equilibrium state, a linearized equation is obtained thus,

$$\Delta \bar{P}_{Ni} = v_0 B_{Ni} \{ K_{Ni} \} \Delta \bar{U}_{Ni}, \quad \{ K_{Ni} \} = \left\{ B_{Ns} S_{sr} + (\delta_{si} + B_{Ns} \bar{U}_{Ni}) \left(\frac{\partial S_{sr}}{\partial \bar{U}_{Ni}} \right) \right\} \quad (4)$$

where B_{Ns} is a shape function. We assemble the elements into a unit so as to represent the ACL shape using a mapping function; this completes the formulation of the problem.

Simulation

Using the finite element method, the finite deformation of the ACL as a function of knee flexion was theoretically analyzed. The ACL was divided into 2380 elements of trigonal pyramid in shape, containing 3700 nodes. Boundary conditions were applied by prescribing the displacements at the boundary nodes corresponding to the insertions. The Newton-Raphson method was used to solve the nonlinear equations associated with finite element representations of the ACL. Special attention was paid for the distribution of longitudinal strain. The results demonstrated that large strain gradients were observed in the regions near the insertions. The influences of some material and histological constants on the strain were also examined. The following values were obtained as a function of knee flexion: three-dimensional change in shape and strain distribution, length change of each fiber, length changes in positions along a fiber, and resultant pull force on the tibial insertion. Further our model made it possible to calculate intra-ligamentary elongations.

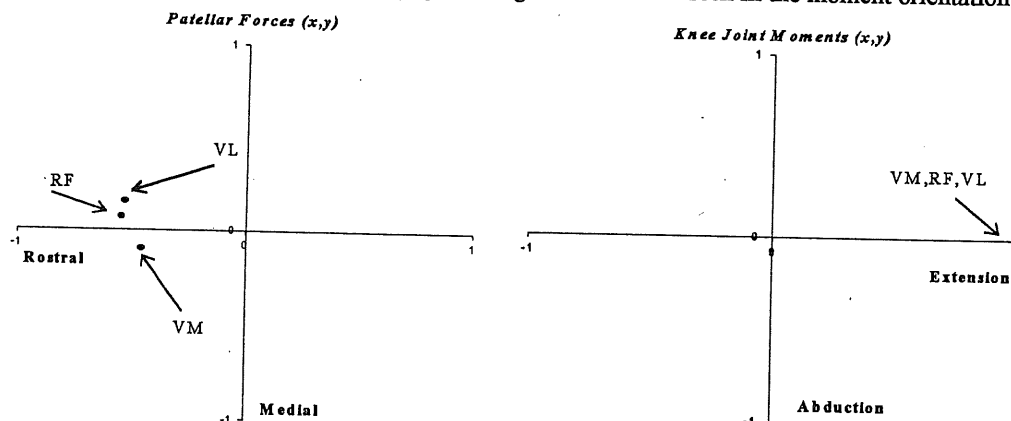
THREE DIMENSIONAL MEASUREMENT OF THE QUADRICEPS FORCES AT THE PATELLA AND THE MOMENTS GENERATED ABOUT THE KNEE JOINT IN CATS

T.A. Abelew and T.R. Nichols
Dep't of Physiology, Emory University,
Atlanta, Georgia, USA

INTRODUCTION: It has been suggested recently that the individual quadriceps muscles may play a role in stabilizing the knee joint during varus-valgus perturbations (Buchanan et al., 1996). Activating muscles selectively could increase tension in portions of the knee joint capsule that are attached to a specific muscle (e.g., vastus medialis) thereby resisting excess varus or valgus displacement. Another way this might be accomplished is by generating varus-valgus moments of the tibia through the patellar ligament. While the individual vasti have divergent lines of action, their attachment at the patella and subsequent attachment to the tibia via patellar ligament suggests a common function. The purpose of this investigation was to quantify the magnitude and direction of the quadriceps forces applied to the patella as well as the moments generated at the tibia.

METHODS: Cats were deeply anesthetized with pentobarbital sodium (Nembutal) and the surfaces of the vastus medialis (VM), vastus lateralis (VL) and rectus femoris (RF) muscles were exposed. Each muscle was implanted with a pair of stimulating electrodes so that contractions could be elicited in each muscle individually. A six degree of freedom force/moment sensor was used to record isometric forces and moments in three dimensions (3D) while attached to the tibia initially and then directly to the patella. Direction cosines for force and moments were calculated so that the orientation of each vector could be displayed.

RESULTS: Direction cosines for average forces at the patella (fig. 1) and average moments at the tibia (fig. 2) are shown. It can be seen that the forces display a divergence that is not seen in the moment orientation at the tibia.



DISCUSSION: The orientation of the applied force on the patella in the medial-lateral direction varies among the individual quadriceps muscles. Figure 1 shows a distinct progression from medial pull on the patella (VM) to a lateral pull (RF and VL) with the greatest lateral pull exerted by VL. Directional differences between these muscles are not seen when the 3D rotational tendencies of the tibia (fig. 2) are examined indicating that the patellar mechanism "couples" the muscles into a predominantly extensor mechanism. The beauty of the knee joint complex is better appreciated when both patello-femoral and tibio-femoral "joints" are considered. The individual force trajectories of the quadriceps exerted on the patella provide a mechanism to insure adequate patellar tracking at the patello-femoral joint. The combined effect of coupling the quadriceps through the patellar ligament maximizes the extensor moment capabilities of the tibio-femoral joint that are required for many natural movements (e.g. jumping). While varus-valgus stabilization does not seem to be accomplished by generating knee joint moments via quadriceps, data have been reported that indicate that the individual hamstring muscles may play a significant role in generating significant knee joint moments in all three dimensions (Abelew et al., 1996). Varus-valgus stabilization by individual quadriceps muscles does, however, remain a possibility by virtue of their attachments to the capsular structures of the knee joint. Supported by NS 20855.

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INDIVIDUAL MUSCLE CONTRIBUTIONS TO THE *IN VIVO* ACHILLES TENDON FORCE

A.N. Arndt¹, P.V. Komi², G.-P. Brüggemann¹ and J. Lukkariniemi²

¹ Institute for Athletics and Gymnastics, German Sport University Cologne, Germany

² Department of Biology of Physical Activity, University of Jyväskylä, Finland

INTRODUCTION

Asymmetrical stress within the achilles tendon is a recognised mechanism in the etiology of degenerative achilles tendon injuries. Individual forces developed by the separate triceps surae components may contribute to such asymmetrical loading. This study was designed to assess the contribution of individual muscle activity to the *in vivo* achilles tendon force. An *in vivo* approach was chosen to eliminate assumptions required in mathematical modelling.

METHODS

While lying in a supine position the subjects (n=4) performed 100% isometric plantarflexion (PF) contractions at systematically varied knee angles and held this force for a one second period. The optic fibre technique of Komi *et al.* (1996, linearity: $r = 0.992$) was used for measuring the achilles tendon force and a strain gauge under the metatarsal heads provided data for calculating the ankle torque. Surface EMG was collected from the soleus (SOL), gastrocnemius medialis (GM) and gastrocnemius lateralis (GL). The integrated rectified EMG signal for each muscle was normalised to the maximum value produced by each subject across all trials. All analog data were collected at 1000Hz.

RESULTS

Fig.1 illustrates the achilles tendon force (ATF) values calculated through a moment arm approach (Hyp.) at 30° and 90° and the *in vivo* ATF (Real). These data are shown at four exertion levels represented as % max.

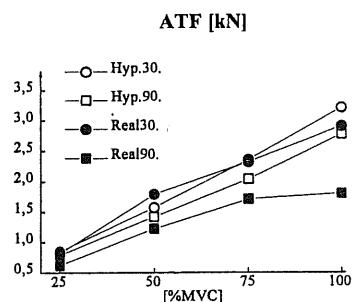


Figure 1.

isometric contraction of the corresponding angle. At 90° the *in vivo* ATF deviated considerably from the hypothetically calculated force: from 76.5% of Hyp at 25% max. to 62% at 100% max. At 30° a far higher proportion of the ankle torque was contributed by the achilles tendon (92.5% and 90% respectively). Fig. 2 represents the normalised EMG for the same conditions. Influences of non-

plantarflexing muscle groups upon the strain gauge values were found to be insignificant and similar across all conditions.

DISCUSSION

Various recent studies have calculated the proportion of PF torque about the ankle produced by the triceps surae as being between 62% and 65%. More differentiated information was obtained in this study by systematically altering both the force-length characteristics of the gastrocnemius and the exertion level. Fig. 1 illustrated a variable significance of the triceps surae on PF torque. At greater muscle length and low exertion levels this proportion was considerably greater than previously reported. Asymmetric muscle contributions to ATF were demonstrated by the EMG results. SOL was consistently more active at 90° than 30° whereas GM and to a lesser extent GL tended to have greater activation when stretched. Only at 100% max. did the gastrocnemii show increased activation even at 90° knee angle. Cresswell *et al.* (1995) described various possibilities to explain decreased gastrocnemius EMG signal when shortened. The data presented demonstrated the complex function of the achilles tendon which can not be explained with present mathematical models and furthermore illustrated physiological possibilities of asymmetrical tendon loading.

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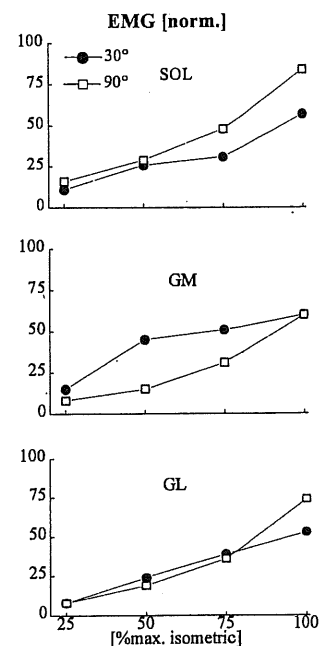


Figure 2.

KNEE FLEXOR MOMENT ARM PATTERNS FOR THE MEDIAL AND LATERAL GASTROCNEMIUS MUSCLES DURING NORMAL MOVEMENT IN THE CAT

R.J. Gregor, T.A. Abelew and A. Albrecht

Department of Health and Performance Sciences, Georgia Institute of Technology

Department of Physiology, Emory University,

Atlanta, Georgia, USA

INTRODUCTION It is generally considered in the movement control literature that the medial (MG) and lateral (LG) portions of the gastrocnemius muscle in the cat are distinctly different in both form and function. Evidence to support this position is increasing as highlighted by recent reports using implanted force transducers in the study of muscle function (Fowler et al., 1993). The histochemical, biochemical, and morphological evidence indicates the LG be considered more powerful than the MG and a muscle that may be used as the final reserve in movements requiring high power outputs (Abelew and Gregor, 1995). While a substantial amount of evidence has accumulated regarding the function of these muscles at the ankle in the sagittal plane, very limited evidence exists on their differential effects at the knee. Important evidence needed to describe their differential affect on knee function, both in the sagittal and non-sagittal planes, rests with an evaluation of moment arms at the knee. Consequently, the purpose of this investigation was to evaluate the differential patterns of moment arms for these two muscles at the knee during normal movements in the cat. The significance of these data lies, for example, with the affect these moment arms may have on potential power transfer and on movements outside the sagittal plane.

METHODS Four cats were implanted with buckle transducers on the tendons of the MG and LG. Following implantation, both kinematic and kinetic data were collected during overground locomotion, incline locomotion, and jumping as described by Fowler et al. (1993) and Abelew and Gregor (1995). Essentially, moment arms for each muscle at the knee were calculated employing an algorithm that produced coefficients relating knee angle and muscle-tendon unit excursion obtained from high-speed film. For each trial, muscle-tendon unit lengths were plotted as a function of knee joint angle, fit with a third order polynomial and differentiated to yield equations describing the instantaneous moment arms (An et al., 1983).

RESULTS AND DISCUSSION Mean moment arm patterns for both MG and LG are presented in Figures 1 and 2, respectively, for eight overground walking trials. These data reveal several features, each with significant implications for muscle function during normal walking. First, it is apparent that the moment arms are relatively constant during a major portion of the stance phase of overground locomotion (25% - 80%). Second, the LG seems to have a longer moment arm than the MG by approximately 2 mm. These data are supported by post mortem dissection. Thirdly, it is apparent that the least amount of variability across trials within a cat occurs during the period of maximum force development (approx. 25% of stance) while the greatest variability across trials occurs early in stance (0-20%) and again during late stance (80-100%) when the cat is preparing for swing. These features suggest that while several options may be available to perform this locomotor task, variability in moment arm patterns is lowest during a key portion of the stance phase. These magnitudes will vary with kinematic output, animal size, etc., but the general patterns agree with those previously reported (Abelew and Gregor, 1995). Finally, it is apparent that the two heads of the gastrocnemius are different muscles, and as a result have profound differential effects on knee joint function.

Fig.1

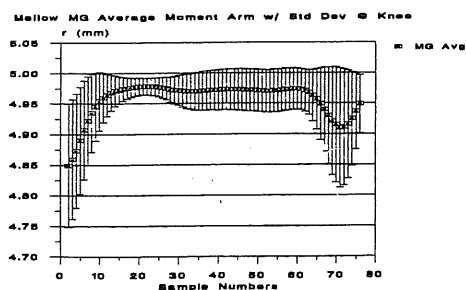
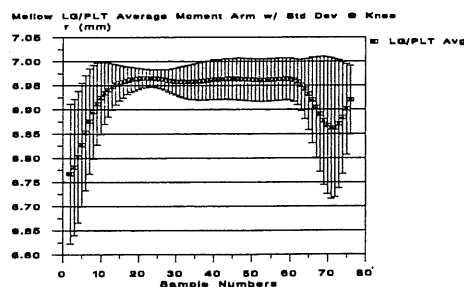


Fig.2



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AN ALTERNATIVE RELATION BETWEEN MUSCLE AND FIBER FORCE IN PLANIMETRIC MODELS

H.F.J.M. Koopman*, B.J.J.J. van der Linden*, P.A.J.B.M. Huijting*^s and H.J. Grootenboer*

* Institute for Biomedical Technology, University of Twente, Enschede, The Netherlands

^s Institute for Fundamental and Clinical Movement Science, VU, Amsterdam

INTRODUCTION

The planimetric model is a relatively simplified representation of pennate muscle for studying the effects of architecture on muscle output [1]. This 2-D model, shown schematically in fig. 1a, consists of straight aponeuroses and fibers. By assuming fiber force-length characteristics and demanding that muscle volume (in this case muscle area) is constant, muscle force can be estimated as a function of muscle length. However, the relation between fiber force and muscle force, derived by Otten [2] and used frequently in this type of model, does not represent mechanical equilibrium when no further assumptions are made. An alternative equation is derived and the effect on the estimated muscle force-length characteristic is compared to experimental data.

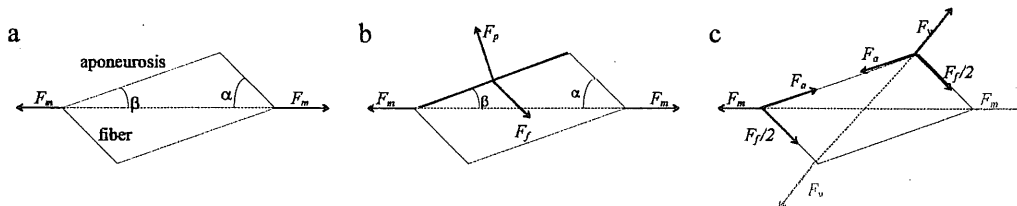


fig. 1a: Planimetric model, with aponeurosis angle β and fiber angle α ; b: Equilibrium of aponeurosis as a free body (eq. 1); c: Equilibrium at corners (eq. 2).

METHODS

The muscle-fiber force relation from Otten [2] is based on equilibrium of the aponeurosis (fig. 1b). A force F_p is assumed perpendicular to the aponeurosis and represents the internal pressure of the muscle. F_p keeps the muscle volume constant. Force equilibrium yields eq. 1 relating muscle force (F_m) and fiber force (F_f). Unless F_p is placed eccentric, there is no equilibrium for the moments of force. Also, the aponeurosis force F_a cannot be derived and equilibrium at the muscle corner points is not assured. Therefore, an alternative description is derived, based on equilibrium at the corner points (fig. 1c). A volume force F_v is assumed in two corners in order to keep the muscle volume constant. This force may in some indirect way be related to internal pressure of the model. The relation between muscle force and fiber force is then described by equation 2. The input of the model consists of an experimental fiber force-length characteristic for the rat GM [3].

$$F_m = F_f \frac{\cos(\alpha + \beta)}{\cos(\beta)} \quad (\text{eq. 1})$$

$$F_m = \frac{1}{2} F_f \frac{\sin(\alpha + \beta)}{\sin(\beta)} \quad (\text{eq. 2})$$

RESULTS and DISCUSSION

The estimated muscle length-normalized force curves for both equations are compared to experimental data for rat GM [4] (fig. 2). The describing difference between both models is the prediction of the optimum muscle length. The original model (eq. 1) overestimates this optimum by about 3 mm, whereas with eq. 2 this overestimate is reduced to only 1 mm. With both models, slack length is predicted well. There is also not much difference between the prediction of architectural parameters, such as the angles α and β , the fiber length etc (not shown here). The model using equation 2 more resembles a finite method, with guaranteed equilibrium in each point of the muscle, both for forces and moments of force. A disadvantage is that the volume force F_v cannot be directly related to the muscle pressure.

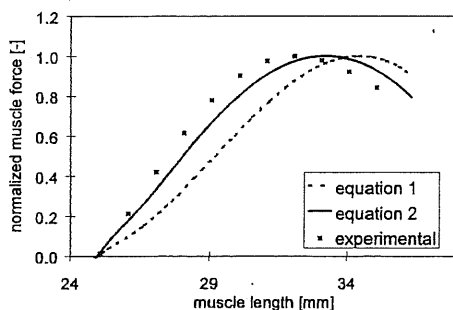


fig. 2: Muscle length-normalized force relations.

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EXPERIMENTAL VALIDATION OF A KNEE JOINT MODEL WITH ELECTRICAL STIMULATION

Maria K. Lebedowska and Richard. B. Stein

Southern Illinois University, School of Medicine, Springfield, Illinois, USA.

Division of Neuroscience, University of Alberta, Edmonton, Canada

INTRODUCTION

The complexity of human body models varies from one to 42 degrees of freedom. Many models neglect the influence of passive and active dynamics and antagonistic muscles activation. The active viscoelastic properties depend on the level of antagonistic muscles activation, but the relations are known only for a limited number of joints.

The aim of this study was to develop a model of the knee joint in the flexion/extension direction including the change of dynamics due to activation of agonist and antagonistic muscles based on experimentally determined parameters.

METHODOLOGY

The human knee flexion/extension model was developed and created with MATLAB software. The model includes the following properties of the knee joint: inertia and mass of the lower leg segment, stiffness and damping changes due to the flexor and extensor muscles activation, stiffness and damping changes due to the angular position, the maximal isometric knee flexion and extension torques at the midrange, the changes of the maximal torque as a function of knee angle. The adequate parameters and functions were measured in two healthy subjects and one patient with spastic diplegia according to the technique developed by Stein *et al.* (1996). Stimulation pulses at 25 Hz for 3 s were delivered to the vastus lateralis muscle with surface electrodes. The isometric, extension torques in the midrange were measured with a strain-gauge transducer; the knee motion as a result of the same stimulation level was also measured with Penny-Giles goniometers. The experimental knee motion was compared with the results of simulation to validate the model.

RESULTS

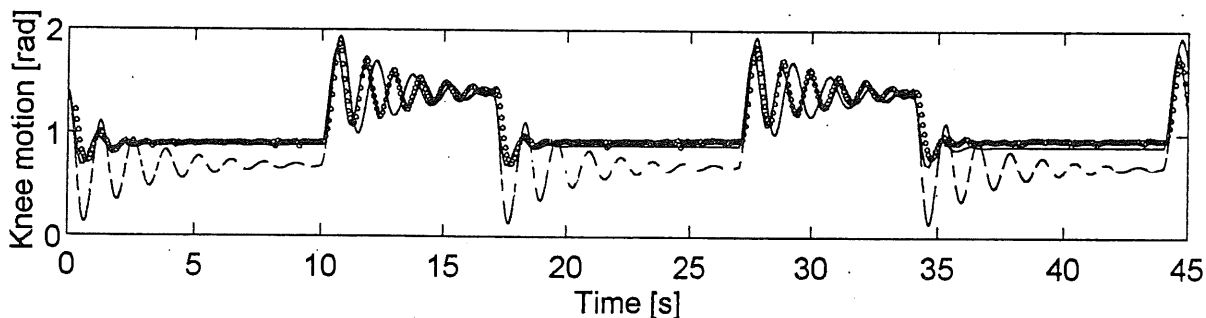


Fig.1 Knee motion measured (dotted line) and obtained from simulation of the model: restricted to the mass and inertial properties (dashed line) and the full model (solid line).

The proportional increase of stiffness and viscous damping as a function of muscle activation was determined for the knee joint muscles. The viscous damping of the knee joint was described by quadratic polynomial equation. There is a definite discrepancy between the experimental data (dotted line) and the model limited to the inertial and mass properties (dashed line), Fig.1. The inclusion of experimentally established nonlinearities into the model increases its accuracy dramatically (solid line).

DISCUSSION

The proportional increase of stiffness and viscous damping agrees with the results obtained for other joints (e.g., Mansour and Audu, 1986). The nonlinear relationship between the damping and angular position agrees with results obtained for the elbow joint by Hayes and Hatze (1977). The passive properties of the joints may be ignored in the midrange, but not at the end of the range of motion as is also true for the knee in walking. The damping properties introduced by the active muscles are an important feature of joint movement and they determine the time of establishing a new state.

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ESTIMATION OF MUSCLE MOMENT ARMS IN FLEXION-PRONATION MOVEMENTS

DirkJan (H.E.J.) Veeger¹, Frans C.T. van der Helm²

¹Institute of Fundamental and Clinical Human Movement Sciences, VUA, the Netherlands

²Laboratory for Measurement and Control, DUT, the Netherlands

To develop a three-dimensional model of the human shoulder and elbow, the flexion-extension and pronation-supination axes of rotation of five human arms were measured, as well as the muscle attachment sites and the contours of relevant bony structures (Veeger et al, 1997). The resulting data have been added to an already existing model of the human shoulder. Special features of the model are the isomorph descriptions of the flexion-extension axis and the pronation-supination axis. In addition, the model is equipped with geometrical representations of bony contours and thus allows for muscle wrapping during simulated arm movements. Once attachment sites are entered into the model, the muscle moment arms relative to arm position can be calculated. Also, regression equations between arm orientation and moment arms can be determined.

The following attachment sites of elbow- and forearm muscles were modelled:

M. Biceps Brachii (BB)	M. Brachioradialis (BR)	M. Pronator Teres
M. Triceps Brachii	M. Anconeus	M. Pronator Quadratus
M. Brachialis	M. Supinator	

The model was programmed to step through combinations of 13 elbow flexion angles (0° - 130°, with 0° as full extension) and 4 pronation angles (0° - 135°, with 0° defined as the anatomical [fully supinated] position). For each of the 52 combinations of elbow flexion and forearm pronation, the moment arms of muscles or muscle-elements was calculated. To evaluate future modelling results, during the dissection procedure, a selection of $dL/d\phi$ measurements were performed on one specimen, from which the moment arms could be derived. Subsequently, the relationship between flexion angle, pronation angle and moment arms were estimated with a multiple regression method. Moment arms for poly-articular muscles showed a significant interaction between rotation modes. As might be expected, moment arms for BB and BR were dependent on both rotation axes. Figure 1 illustrates the moment arms for pronation and supination in BB.

Traditionally, moment arms have been estimated by determining the relationship between muscle length change and changes in joint angles ($M=dL/d\phi$, Fick, 1911, Murray et al, 1995). The disadvantage of this method, however, is the difficulty in determining muscle length changes during combined movements (Murray et al, 1995) as well as the experimental difficulties in the estimation of the moment arm angular displacements around more than one axis. In this study, the moment arms calculated by the model and $dL/d\phi$ results from the dissection revealed identical results. It can thus be concluded that the computer model accurately estimates muscle moments and that muscle wrapping and isomorphic rotation axes are a useful, if not necessary, extension to the model. Also, moment arms could well be predicted from regression equations. In addition, the calculated moment arms were comparable to results for elbow flexion by Murray et al (1995).

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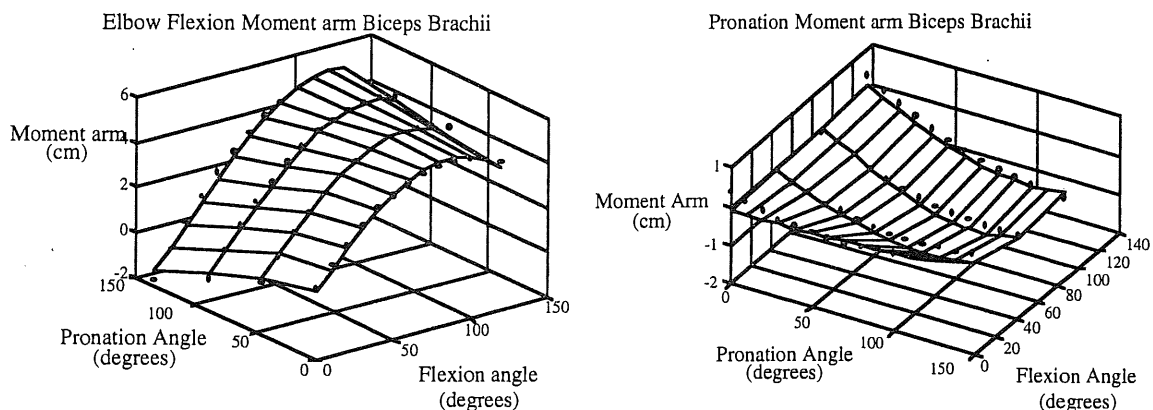


Figure 1

Calculated (lines) and estimated (dots) moment arms for flexion an pronation (BB, N=1)

ADAPTATION OF THE HUMAN CALCANEUS TO IMPACT FORCES DURING RUNNING

Kersting, U.G., Brüggemann, G.-P.

Institute for Athletics and Gymnastics, German Sport University Cologne

INTRODUCTION

High impact forces at touch down have previously been associated with overuse syndromes during heel-toe-running. Recent work by Nigg *et al.* (1995) questioned the direct relation between high impacts and overuse injuries because many earlier studies concerning the etiology of injuries showed no significant correlation between these two factors. However, experiments concerned with the dependency between shoe construction and impact magnitude showed that forces at the shoe-foot-interface may vary while forces at the shoe-ground-interface remain unchanged (Kersting & Brüggemann, 1995). According to the fact that GRF-measurements mirror the resultant forces acting on the whole body system it can be assumed that the load of internal structures in the foot is more closely related to the forces under the heel at touchdown. In order to quantify the effect of impact magnitude on the adaptational response of biological tissues, e.g. tendons, cartilage and bone, longitudinally designed studies are necessary to investigate the dependency of these factors. The aim of this study was to look at bony adaptation of the calcaneus as it was assumed to be one structure of the skeletal system which is exposed to the impact forces during heel contact in running.

METHODS

26 subjects were equipped with running shoes of different midsole hardness and performed a running program of 40 km per week minimum over a period of 20 weeks. GRF and in-shoe forces under the heel were measured by a force platform and a strain gauge force sensor respectively to describe the mechanics of foot-ground contact. Rearfoot motion was recorded by an electro-goniometer. Four different parameters representing bone quality of the calcaneus were investigated. The ultrasound-transmission velocity (SOS) was measured at two different sites of the calcaneus. MRI-Scans of the rearfoot were performed using three different protocols: Spin echo (SE), partial saturation (PS) and short T1 inversion recovery (STIR). Images were quantitatively analysed to look at differences in signal intensity in the center and basis of the calcaneus and its total cross sectional area. Statistical comparison was carried out by means of repeated measures ANOVA.

RESULTS

Measures of the external forces, impact and active maximum, showed no variation according to altered midsole hardness. In-shoe forces also varied independently of the physical characteristics of the shoes. Moreover, the observed differences in impact magnitude, whether inside the shoe or external, could not be explained by rearfoot motion patterns. The training group was therefore, subdivided into three groups of high (H), medium (M) and low (L) in-shoe impacts. The bone quality parameters revealed specific differences in the investigated groups but no variation was observed in dependency of the bone region. Figure 1 represents the measured values before and after the training period. The SOS increase for every subgroup and the total sample was significant ($p < 0.01$); the absolute difference was greatest for group M. SE signal experienced only minor changes in any of the subject groups. PS signal was significantly ($p < 0.05$) increased for runners with medium in-shoe impacts while the two other groups showed only slight gains. The STIR intensity showed a significant ($p < 0.01$) decrease for H, M and L which was most pronounced in group M.

DISCUSSION

The present data illustrates that bone adaptation during running is impact selective since no differences in active forces were observed. The constant SE values suggest that no alteration in trabecular density occurred (Majumdar, 1991). However, the changes in PS and STIR intensity indicate metabolic changes of the bone marrow. Current literature does not provide sufficient information whether the identified alterations characterise increased or decreased bone formation. Future research should focus on the physiological background of the observed changes.

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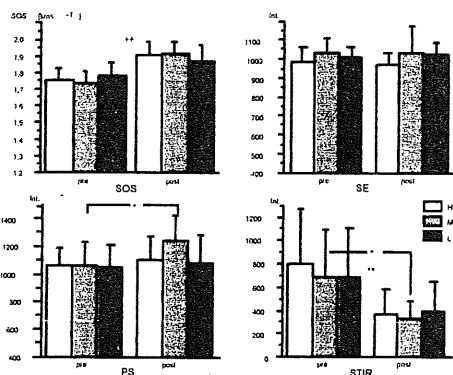


Figure 1: Values of bone parameters before and after 20 weeks of running. * = significant difference ($p < 0.05$) pre vs. post; ** = sig. diff. ($p < 0.01$) total sample; ++ = sig. diff. ($p < 0.01$) all subgroups and total sample; Int. = Intensity in MRI image.

EFFECTS OF SHOE CONSTRUCTION ON SKELETAL MOVEMENTS DURING RUNNING

A. Stacoff*, B.M. Nigg*, C. Reinschmidt*, A.J. van den Bogert*, A. Lundberg†, J. Denoth‡, E. Stüssi‡

* Human Performance Laboratory, The University of Calgary, Calgary

† Karolinska Institute, Dept. of Orthopaedics, University Hospital, Huddinge, Sweden

‡ Biomechanics Laboratory, ETH Zürich, Schlieren, Switzerland

INTRODUCTION

Excessive foot eversion and tibial rotation has been associated with various running injuries at the lower extremities [1]. Initial studies with shoe or skin markers concluded that specific shoe sole geometries and/or foot orthoses can be used to control/reduce foot eversion and tibial rotation [2][3]. However, recent study results show that skin and shoe markers overestimate the skeletal movements [4]. Therefore, initial conclusions on the effects of shoe sole geometry and foot orthoses alterations may be incorrect. The purpose of this study was to quantify the effects of medial foot orthoses and lateral heel flares on skeletal movements of calcaneus and tibia during the stance phase in running. The hypotheses to be tested were:

- maximum calcaneal eversion, maximum internal tibial rotation and their respective velocities are smallest for the round shoe sole and posterior foot orthoses and largest for the flared shoe sole and no orthoses.

METHODS

Intracortical Hofman pins with triads of reflective markers were inserted under standard local anesthetic into the calcaneus and tibia of five healthy male subjects. The calcaneus rotations with respect to the tibia were determined using a joint coordinate system approach, with a three-dimensional marker reconstruction [5]. Shoe sole constructions were tested with three lateral heel flares of 25° (flared), 0° (straight), and a rounded sole. The effect of medial foot orthoses was tested with two 1cm thick cork supports, one anteriorly to support the foot arch and one posteriorly, to support the sustentaculum tali; the normal insert had no support.

RESULTS AND DISCUSSION

The effects of heel flare and medial arch support on eversion (Table 1) and tibial rotation were small (less than 1.2°) and not systematic. The inter-subject differences were larger than the intra-subject differences. Also, maximum internal tibial rotation was found to be larger than maximum eversion in four of the five subjects.

Shoes Subj.	flared		straight		round		anterior		posterior		normal	
	$\beta_{\max}$	$\dot{\beta}$	$\beta_{\max}$	$\dot{\beta}$	$\beta_{\max}$	$\dot{\beta}$	$\beta_{\max}$	$\dot{\beta}$	$\beta_{\max}$	$\dot{\beta}$	$\beta_{\max}$	$\dot{\beta}$
1	-0.1	172.5	-0.8	144.5	0.2	191.5	0.9	168.1	1.8	171.4	1.9	158.9
2	2.5	161.3	4.0	138.2	3.4	162.5	3.1	152.3	1.5	125.0	4.1	163.4
3	5.2	108.6	5.9	125.1	6.4	149.4	7.5	150.7	7.2	110.8	8.8	148.0
4	6.6	230.2	6.5	249.2	3.6	142.1	4.4	150.4	5.5	162.4	3.5	147.8
5	2.3	112.4	2.0	97.9	2.2	86.0	0.4	85.8	-0.9	96.9	2.1	77.0
Mean	3.3	157.0	3.5	151.0	3.2	146.4	3.3	141.5	3.0	133.3	4.1	139.0

Table 1: Means of maximum eversion ($\beta_{\max}$ [°]) and maximum eversion velocity ($\dot{\beta}$ [°/s]) with different heel flares (left) and different foot orthoses (right); 3 reps. per condition (mean sd = 1.1° resp. sd = 32.6°/s).

The results of this *in-vivo* study quantifying calcaneal and tibial movement during running are surprising, do not support the hypotheses to be tested, and do not agree with findings of earlier studies using skin or shoe markers, although the shoe sole and orthotic alterations of this study were similar (earlier reported heel flare and orthotic effects: 10-50% [3][6]). It may be concluded that the previously reported heel flare and orthotic effects are not apparent at the bone level, however this is contradicted by clinical evidence [7], and/or the five subjects in this study are unrepresentative, although their barefoot results were similar to previous studies [2][3]. Therefore,

- it is suggested that specific movement patterns are associated with individual foot-leg combinations such that shoes and orthoses cannot change them substantially. Hence, the major changes may concern muscle activity rather than foot and shank movement patterns, which is supported by indirect evidence [8], finally
- it is suggested that the calcaneus is not the relevant bone to be assessed, and mid and fore-foot movements are more important to the understanding of shoe sole or orthotic alteration effects.

Currently, a conclusive interpretation of shoe sole and orthotic alteration effects is not possible. Further research is necessary to increase the understanding of this presumably important aspect.

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COMMON GAIT PATTERNS BASED ON THE KNEE ABNORMALITIES IN CEREBRAL PALSY

Fong-Chin Su, Lan-Yuen Guo, You-Li Chou, Chii-Jeng Lin*, Rong-Ju Cherng**

Institute of Biomedical Engineering, *Department of Orthopaedic Surgery, **Department of Physical Therapy, National Cheng Kung University, Tainan, TAIWAN

INTRODUCTION

Sutherland et al. (1993) classified the common gait abnormalities of the knee into four types: jump knee, crouch knee, stiff knee, and recurvatum knee. In his review, each abnormality was described by physical examination, motion parameters, electromyography (EMG) data, and force plate data. Joint angles were represented in terms of projected angles into three major "planes of movement", not Euler angles. In addition, statistical comparison of motion parameters among gait patterns was not performed. The purpose of this study is to classify the common gait patterns based on the knee abnormalities in cerebral palsy and statistically analyze their differences, especially at specific gait events. The effect of joint angle representation method was also examined.

METHODOLOGY

Twenty-three children (mean age, 8.9 yrs) with cerebral palsy were recruited in this study. The ExpertVision™ motion analysis system (Motion Analysis Corp., Santa Rosa, CA) was used to collect marker trajectories. Helen Hayes marker set was used. Five trials were sampled for each subjects at 60 Hz. In addition, three Kistler force plates were used to collect ground reaction forces at 100 Hz sample rate, simultaneously. OrthoTrack software was used for kinematic and kinetic evaluation of the lower extremities in three-dimensions. Euler angles were used for joint angle representation. Three normalized (100%) gait cycles for each subject were averaged for all gait parameters.

Based on the kinematic patterns of the knee in sagittal plane (Fig. 1), we separated the 46 limbs into four groups, mild knee ($n = 17$), crouch knee ($n = 8$), recurvatum knee ($n = 14$), and jump knee ($n = 7$). One way ANOVA was used to find differences of kinematic characteristics, angular arc and peak angle, in hip and ankle, temporal/spatial parameters, and ground reaction forces. The differences of motion parameters were also evaluated at five gait events, including heel strike, opposite toe-off, opposite heel strike, toe off, and heel strike. Finally, Tukey's post-hoc test was used to determine the significance among the groups.

RESULTS

There are significant differences ($p < 0.05$) between the four groups for stride length to leg length ratio. Jump knee gait group has the largest value 1.41, and the crouch gait has the lowest value 1.01. At these five gait events, the sagittal plane angles of the hip, knee, and ankle joints are significantly different between the four groups ($P < 0.05$). In addition, there are significant differences between the four groups in the maximum and minimum angles of the hip, knee, and ankle joints ($P < 0.05$). For the hip joint angle in the sagittal plane (Fig. 2), crouch gait group has a tendency to flexion. And the maximum extension is -14.8 degrees which is significant lower ($p < 0.05$) than the other groups. For the ankle joint angle in the sagittal plane (Fig. 3), crouch gait group has a tendency to dorsiflexion and the recurvatum group is to plantarflexion. The jump knee group is similar to normal group at loading response, but later has a premature plantarflexion at initial stance.

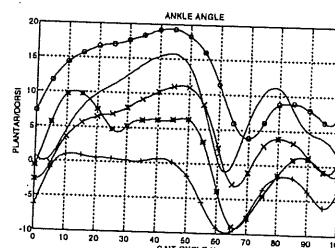
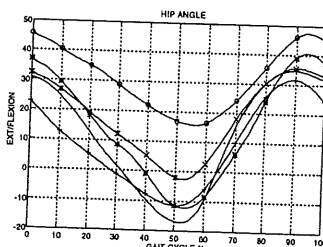
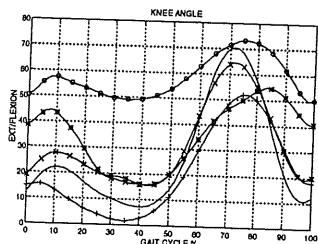


Fig. 1: Sagittal plane knee angles Fig. 2: Sagittal plane hip angles Fig. 3: Sagittal plane ankle angles
(— : Normal, o: Crouch, ×: Mild, *: Jump, + : Recurvatum)

DISCUSSION

In general, our joint motion profiles are similar with Sutherland's results except recurvatum knee joint angle and hip angles. Sutherland demonstrated a typical hypertension of knee secondary to contracted and spastic triceps surae for a pre-rhizotomy patient. In Fig. 1, the curve of recurvatum knee angle is a mean of 14 legs, which shows similar profile, but without hypertension (1 degree flexion). Furthermore, there is a 18 degree offset in hip joint angle between our 3-D Euler angle and Sutherland's 2-D projection angle. The hip of jump knee group doesn't show a pattern of exaggerated flexion, which may be caused by taking average. Also, computation algorithm of joint angle affects the final results, especially for hip joint definition. There are big differences between 2D and 3D kinematics. Therefore, each motion laboratory is recommended to have its own database. Each patient must be evaluated individually.

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ACKNOWLEDGMENTS: This work was supported by National Health Research Institute HR-410, Taiwan, ROC.

DIFFERENT SUPPORT SYSTEMS RESULT IN DIFFERENT ANKLE BIOMECHANICS DURING GAIT

Nishikawa T., Lundin T.M.*, Grabiner M.D.*

Department of Orthopaedic Surgery, Hyogo Prefectural Kakogawa Hospital, Hyogo, Japan

*Department of Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, OH, USA

INTRODUCTION: Following an ankle joint injury, a significant number of patients report residual symptoms, the most common of which are recurrent injuries. The recurrence of the injuries is not limited to vigorous activity and may occur during activities as benign as walking. There is a large body of literature in which the effects of various prophylactic ankle appliances on variables such as ankle range of motion, ankle muscle strength, reflex and voluntary reaction time, postural steadiness, and functional performance tasks have been reported. Far fewer publications report on the influence of ankle support on the biomechanics of the ankle complex during walking or running.^{1,2,3,4} The purpose of this study was to examine the effects of two ankle braces and ankle taping on the kinematic behavior of the ankle complex, in particular that of inversion-eversion, during walking. Based on previous studies that have investigated passive range of motion, it was expected that the ankle supports would reduce ankle complex inversion at ground contact and during the stance phase of walking.

METHODS: Four healthy women and 8 healthy men between 25-40 years of age, volunteered to participate in this study and provided informed consent. All of the subjects wore the same model of athletic shoe during the experiment. The subjects were required to walk approximately 7 m at a self-selected walking velocity and to target a force plate with their right foot. Six acceptable trials were collected, the criteria for which were the right foot being completely on the targeted force plate and the average walking velocity, monitored with photoelectric cells, within 10% of the previously established preferred walking velocity. Lower extremity kinematic data were collected using a video-based motion analysis system that recorded the motion of seven reflective markers placed on the base of the first metatarsal, the lateral aspect of the fifth phalanx, the posterior ankle joint line, the medial and lateral knee joint axis, and the anterior side of the tibial shaft. Ankle complex inversion and eversion were characterized as rotation about an axis extending along the midline of the foot. Eversion was associated with the lateral border of the foot moving towards the lateral aspect of the shank. The variables extracted for analysis included (a) the medial-lateral position of the center of pressure at heelstrike relative to the inversion-eversion axis, defined by the line joining the average of the two forefoot marker and heel marker and expressed as a percentage of shoe width, (b) the touchdown angle (inversion) at heelstrike (c) the maximum eversion angle, (d) the total inversion-eversion range of motion, and (e) maximum eversion velocity. These variables were compared across four ankle support conditions: taping, semirigid Aircast Air-Stirrup (Aircast, Inc.), lace-up, cloth RocketSoc (DonJoy, Inc.), and an unbraced condition. The mean value for each subject was computed and the group data were submitted a repeated measures ANOVA.

RESULTS: The mean walking velocity of the subjects was $1.56 \pm 0.07 \text{ ms}^{-1}$. As expected, the touchdown inversion angle was decreased by the bracing and taping conditions. Compared to the control, unbraced condition, in which the inversion angle at ground contact was 4.23 ± 2.57 deg, the inversion angle for the Aircast, DonJoy, and taped conditions were 2.40 ± 3.29 , 1.74 ± 2.64 and 0.75 ± 3.42 deg, respectively, the latter two of which were significantly different than the unbraced condition ($p < .01$). The DonJoy brace was found to be associated with a significantly greater lateral displacement of the COP compared to the unbraced condition. Although the total inversion-eversion range of motion was similar for both braces and the tape condition, the DonJoy brace was associated with a 17.4% larger maximum eversion velocity compared to the unbraced condition. (Table)

Table Mean value (SD) - Ankle Complex Kinematics Variables
Center of Pressure Across Ankle Supports Conditions (N=11)

	Lateral shift of COP at Heelstrike (Percentage)	Touchdown Angle (Degrees)	Max Eversion Angle (Degrees)	Total Inversion- Eversion ROM (Degrees)	Max Eversion Velocity (Degrees/Sec)
Unbrace	9.63 (6.95)	4.23 (2.57)	2.82 (1.93)	8.73 (1.55)	68.05 (18.31)
Aircast	8.56 (8.53)	2.40 (3.29)	3.46 (2.98)	8.02 (2.26)	70.89 (19.48)
Donjoy	13.06 (8.57)†	1.74 (2.64)**	4.78 (2.97)*	8.70 (2.11)	79.92 (27.40)
Tape	10.00 (8.12)	0.75 (3.42)**	4.82 (3.66)*	9.14 (2.65)	71.30 (29.60)

* $p < 0.05$, significantly different from unbraced condition

** $p < 0.01$, significantly different from unbraced condition

† $p < 0.05$, significantly different from semirigid braced condition

CONCLUSIONS: The results revealed that the braces and taping had dissimilar effects on ankle complex biomechanics during the relatively benign activity of walking. Both the DonJoy and tape conditions increased maximum eversion during the stance phase but the DonJoy alone was associated with a substantially larger maximum eversion velocity. The increased maximum eversion velocity, the significance of which requires further study may be a function of the significant lateral shift of the COP at touchdown caused by the DonJoy brace and the lack of medial restraints on the DonJoy brace. Although the data collection was restricted to the leg and foot, the results indicate that the two braces and the tape do not similarly affect ankle complex biomechanics during gait. These observed differences may have subtle but important considerations during rehabilitation from ankle injuries but may require inclusion of the upper body in the analysis.

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Acknowledgments: The authors gratefully acknowledge funding provided by Aircast, Inc.

GAIT ADAPTATIONS IN BELOW KNEE AMPUTEES AND EFFECT OF WEIGHT CHANGE IN PROSTHETIC COMPONENTS: A PILOT STUDY

S.J. Olney, H. Bateni, M.P. Griffin and I.D. McBride

School of Rehabilitation Therapy and Department of Mathematics and Statistics, Queen's University, Kingston, Ontario, Canada

INTRODUCTION

Persons who have sustained an amputation below the knee have lost most of the power-generating capabilities of the ankle joint. As a result, the knee does not flex as readily as a normal limb as it enters swing phase, and the amputee has a sensation that the prosthesis is heavy. New components are entering the market with great rapidity, and one of the factors considered is the weight of the components. The most common material for the modular shaft is steel. Alternatively, sometimes titanium is used which is very light but also expensive. The purpose of the study was to identify gait adaptations commonly used by below-knee amputees and to provide preliminary information on whether the use of lighter modular prosthetic components result in changes to the gait.

METHODS

Five unilateral experienced below-knee amputees with no medical or physical problems participated in the study. An instrument was developed to assure constancy of alignment when components were changed. Three gait trials for each side of each subject under each condition were collected using a Peak Performance 2-D motion analysis system and an AMTI force platform. The motion and force data were used in a four-segment link segment inverse dynamic model to calculate spatio-temporal values, joint angles, moments and powers. A heart-rate based Energy Expenditure Index (Rose et al., 1990) was used over a five minute walk to assess comparative energy costs. Subjects were tested using both steel and titanium components with the order randomly allocated. All gait data were processed to yield normalized stride profiles, and important values were extracted from the data base. The profiles were compared visually and important variables were tested using a Student-t test.

RESULTS AND DISCUSSION

The mean stride length was 1.31 m, stride time was 1.91 s, and walking speed was 1.10 m/s. Kinematically, most subjects showed hip flexion angles at initial contact increased by at least 10 degrees above normal values on both sides. In most subjects the knee joint flexed excessively during the initial flexion phase in early stance on the normal side. The most likely explanation for this unexpected finding is that the resulting large power burst was a compensation for the deficiency in ankle plantarflexion power on the prosthetic side. The ankle angles of the prosthetic side showed normal or above normal values of dorsiflexion in stance phase, but little rebound plantarflexion during push-off. In contrast, the ankle joints of the normal sides usually showed higher than normal dorsiflexion, perhaps related to the increased step length of the prosthetic side. Push-off occurred later than normal, consistent with the longer stance phase of the normal side.

Regarding kinetics, many subjects showed increased hip extensor moments in early stance on the prosthetic side; most showed this feature on the normal side. This was consistent with the need to exploit available sources of power generation to compensate for the lack of ankle plantarflexion power on the prosthetic side. Similarly, a larger than normal hip flexor moment was evident in late stance on both sides. This feature, noted in other pathologies (Olney and Richards, 1996) results in high values for the positive power generation by the hip flexors in push off and early swing phase. On both sides, the knee extensor moments in late stance were usually high, consistent with large amounts of absorption, typically occurring when there is some gait instability. Ankle moments were reduced on the prosthetic side. The most common compensatory power patterns were increased generation at pull-off of the normal-side hip, and frequently of the prosthetic side, and in the hip extensor generative phase in early stance on both sides. Ankle generation on the normal side was often increased, as was generation in the knee extension phase of early stance. Excessive absorption by knee extensors in late stance was evident on both sides in all trials. Although small differences in gait variables were noted with changes in weight of components, they were not significant or consistent. The Energy Expenditure Index showed significantly lower values with lighter components, which warrants further study on larger numbers of subjects.

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CEREBRAL PALSY AND RHIZOTOMY: A 10 YEAR FOLLOW-UP WITH GAIT ANALYSIS

Nivedita Subramanian*, Christopher Vaughan**, Jonathan Peter**, Leila Arens**

* New York University, USA ** University of Cape Town, South Africa

Introduction

Selective dorsal rhizotomy is a neurosurgical procedure performed for the relief of spasticity in children with cerebral palsy (Peacock and Arens, 1982), but its long-term functional efficacy is still being assessed. Our three year follow-up study demonstrated improvement in functional range of motion during gait (Vaughan *et al.*, 1991), while our subjective assessments after ten years have also provided some evidence of benefit (Peter and Arens, 1993). In the current study we sought to evaluate objectively the long-term efficacy by means of gait analysis.

Methods

We followed 11 children with spastic diplegia (mean age at initial surgery was 7.8 years) using computer based gait analysis. Retro-reflective markers were placed bilaterally on the hip, knee and ankle joints, allowing us to measure temporal-distance parameters as well as sagittal plane ranges of motion (Vaughan *et al.*, 1991). This was a prospective study with the children being evaluated pre-operatively and then at one, three and at least ten years after surgery. For comparison purposes, a group of 12 age-matched normals were also studied.

Results

The knee and hip ranges of motion were significantly restricted prior to surgery, but were within normal limits after ten years (*cf.* Figure 1 below left for the knee). The knee and hip mid-range values -- which are indicative of posture -- were significantly elevated pre-operatively, and increased sharply at one year, but by ten years they had decreased to within normal limits (*cf.* Figure 1 below right for the hip). Stride length and velocity improved post-operatively but were not in the normal range after ten years.

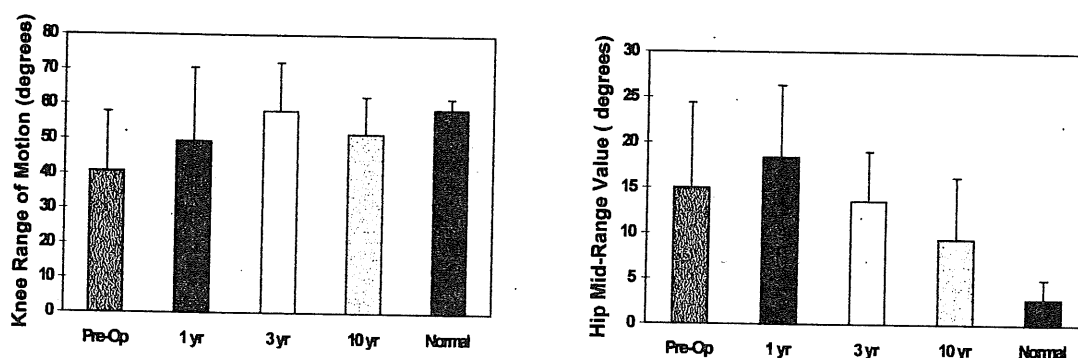


Figure 1 Ensemble averages (with standard deviations) for the knee range of motion (left) and the hip mid-range value (right) of 11 cerebral palsy children compared with 12 normals. The patients were studied before surgery and at one, three and ten years after surgery.

Discussion

It has been well-established that the immediate consequence of selective dorsal rhizotomy is a dramatic reduction in spasticity (Engsberg *et al.*, 1996). There is thus the *potential* to improve the patients' neuromuscular function. The objective data that have been presented to date suggest that in the short-term (*i.e.*, from one to three years after surgery), patients do benefit (Vaughan *et al.*, 1991). While there are complex issues such as the influence of subsequent physiotherapy and orthopaedic surgery that need to be addressed, the results from the current study provide some evidence to show that the long-term functional range of motion has improved. We believe that this improvement can be directly attributed to the rhizotomy surgery.

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