

THE MECHANICS OF BIOLOGICAL POROUS MEDIA : VARIATIONAL APPROACH AND NUMERICAL APPROXIMATION

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INTRODUCTION

For a host of medical outlets it is paramount to know the rheological behavior of biological materials. Preliminary studies have enabled to describe such a behavior through viscoelasticity, ascribing to each tissue a model in which several parameters, without any particular physical meaning, come into play.

However, a structural model is necessary to understand the governing phenomena. Taking into account the diphasic characteristic of these tissues, which can be modelised as a solid elastic matrix embedded in an interstitial viscous fluid, at the first stage, the theory of poroelasticity seems to be adequate. Developed by Biot /1/, it allows to describe one particular type of behavior drawing from mechanical parameters which are all standard knowledge and are easily measurable.

If both theories of mixtures and homogenization allow a description of the behavior of a polyphasic material from a general point of view, the theory of consolidation (i.e. poroelasticity) help derive easily coupled partial differential equations, in the case of porous media, in particular.

We present here, their variational formulation for one class of problems, leading to an approximation by the finite-element method.

GOVERNING EQUATIONS

From the basic concepts of the mechanics of continuous media, let us take as constitutive equations, in the case where both phases are incompressible:

$$\left\{ \begin{array}{l} \sigma = 2G \epsilon + \frac{2Gv}{1-2v} \text{Tra } \epsilon - p \mathbb{I} \\ \frac{\partial (\text{Tra } \epsilon)}{\partial t} = \frac{\kappa}{\mu} \Delta p \end{array} \right.$$

where σ is the stress tensor G, v are constants with $p = 0$
 ϵ is the strain tensor μ is the dynamic viscosity of the fluid
 $\mathbb{I}$ is the unit tensor κ is the permeability of the medium as defined by Darcy's law.

Taking into account the equilibrium equation, we obtain an equation similar to Navier-Stokes'.

If we consider a cylindrical sample subject to creep, relaxation or elongation tests, an analytical method of resolution was made up, based on the superposition of transient and permanent phenomena. Our method allows to determine the different quantities which come into play. Lately, in order to study the behavior of more complex structures, we have developed a numerical method of resolution /2/.

NUMERICAL RESOLUTION

So as to develop the numerical approximation of the solutions, for one class of problems, and to prove the existence and the unicity of these solutions too, we have introduced the notion of low solutions by the variational method.

This method consists in considering the pressure of the fluid as known at a first stage. This leads us to solve a problem by displacement where the pressure gradient represents a given loading of volume. Using the Lax-Milgram theorem, the existence and the unicity of the solution can be shown. Knowing then displacement, we solve the problem by pressure, falling into two new problems. One of them tallies nicely with the abstract theory as developed for Stokes' problem. So we can establish a theorem of existence and unicity for the pressure related problem.

To sum it all up the variational formulation helps put forward an algorithm, allowing to determine displacement and pressure to the medium. As a matter of fact, to set forth this algorithm we need the finite-element method which offers a great versatility once the implementation is effected.

RESULTS

We based our algorithm on the MODULEF finite-element library (developped by I.N.R.I.A.). We take the example of a cylindrical sample subject to a creep test. A load is applied in an impermeable manner at the top, the base is impermeable and blocked axially, moreover, the lateral walls are free of any stress. Figure 1 and 2 emphasize the structure before and after deformations and the isopressures at one given time.

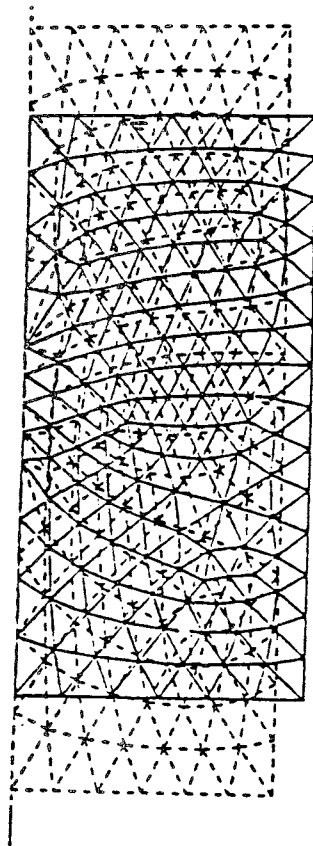


Figure 1: structure before and after deformation at one given time.

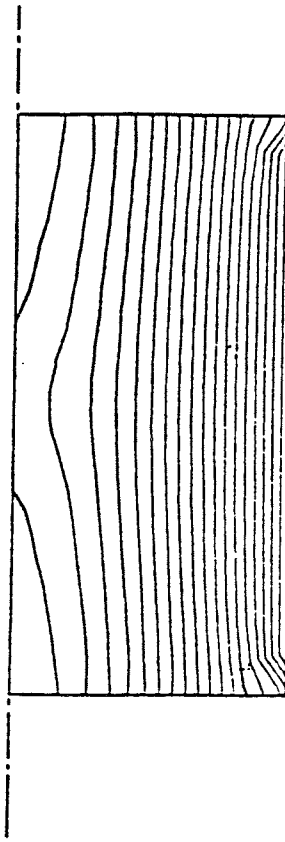


Figure 2: the isopressures at the same time.

CONCLUSION

As a conclusion, we can assert that the results are interesting for isotropic media with a constant permeability. However, to describe the behavior of other materials, we have to develop a general model with anisotropy and variable permeability, and then to compare with experiments.

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LIMB-JOINTS MECHANICAL ANALYSIS: THE USE OF FORCE PLATES AND C.C.D. CAMERAS.

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INTRODUCTION

Gait is a periodic phenomenon including a stance phase and a swing one. This study requires the simultaneous recording and analysis of kinematic and dynamic data during the same cycle. For a clinical use, special conditions must be met: the system must be entirely managed by computer, the data have to be analysed immediately after recording and the results provided as soon as possible in an easily comprehensible and readable form. Moreover, the set up of the patient for the tests must be fast and the examination itself should not exceed a few minutes.

MATERIAL

Forces between the foot and the walking track are measured by a force plate. The walking track is 10 meters long and its sensitive part (.8 m long, .3 m wide) is equipped with four three - component force transducers (strain gages).

Six contact - less displacement analysers have been built with charge transfert components (CCD video, single line camera) /1/. A small lamp is fixed on the external face of each articulation of the leg, as close as possible to its virtual axis. The beam light is focused on the device by a photographic lens. A hemi cylindrical lens, between the camera and the component makes it possible to increase the vertical field and therefore to neglect vertical movements. Two cameras are necessary to locate each point, one in the frontal plane, and the other in the sagittal plane; the fields of the camera are respectively .3 by .5 m (front.) and .3 by 1.25 m (sagit.). An electronic treatment card transforms the number of order of the lightened cell into a word of eleven bits. A preliminary calibration enables to know the coordinates of each articulation with reference to the force plate. Then, the coordinates of the articulations of the lower limb are recorded during the stance and the swing phases.

The channels from the force plate and from the cameras are sampled at the rate of 100 per second and a conversion of parallax error is automatically done after the recording of the signals (figure 1).

RESULTS

STANCE PHASE - The sequence of measurements starts when the foot hits the force plate. Forces are normalized by the weight of the subject and time is a percentage of the stance phase time. In addition to the variations of forces and of the displacement of the center of pressure under the foot, the angles between the vertical and the various segments of the limb (thigh and shank) are plotted. The variation of the knee flexion as well as horizontal speeds of the hip, the knee and the ankle, in the walking direction are plotted. Trajectories of the articulations and the respective positions of the limb segments are plotted drawn. It is assumed that the articulation of the hip remains at a constant height during gait (figure 2a).

SWING PHASE - The kinematics of the swing phase is recorded during the stance phase of the other leg. The arrangement of curves and graphs are identical to that of the stance phase.

This method makes it possible to know the respective positions of each segment of the leg during swing phase with respect to the stance phase of the opposite leg. Thus, the abnormalities in the swing phase of above knee amputees can be easily highlighted and characterized (figure 2b).

CONCLUSION

The system suggested here enables us to record in a simple and rapid manner the kinematic and dynamic parameters of gait. The small amount of time required for the examination, as well as the absence of preparation of the subject and the immediate results make the method perfectly suitable for clinical practice.

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Figure 1 - Schematic diagram of the acquisition system and of the treatment process.

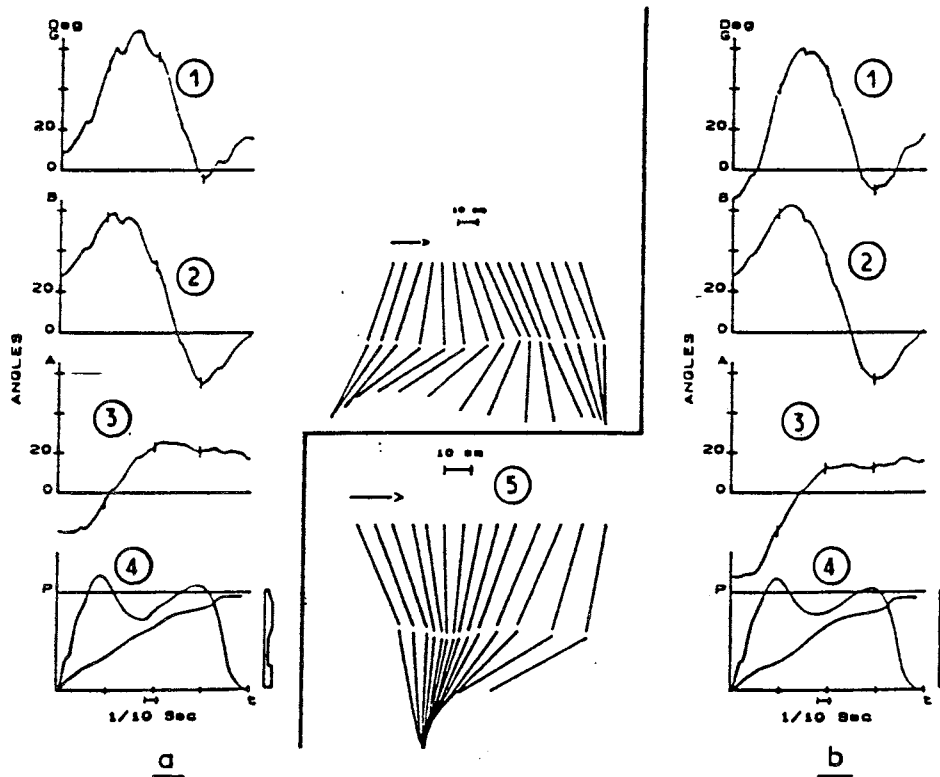
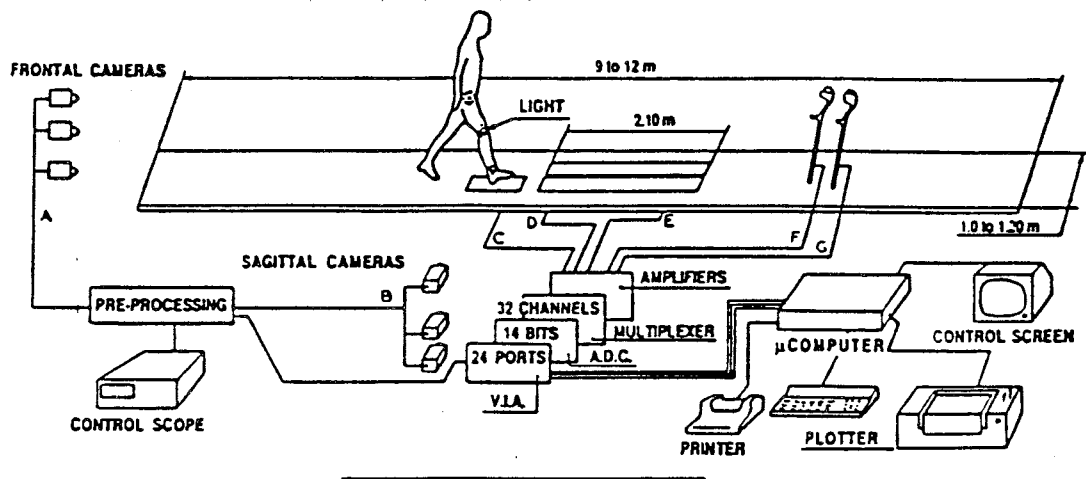


Figure 2 - Typical curves and plots for a normal subject: (a) swing phase, (b) stance phase. 1 - knee flexion; 2 - shank angle to vertical; 3 - thigh angle to vertical; 4 - vertical reaction, normalized by the subject total weight; 5 - bar plot of the leg segments.

STRENUOUS EXERCISE EFFECTS ON THE MECHANICAL PROPERTIES AND GEOMETRY OF WEIGHTBEARING VS NONWEIGHTBEARING MATURE BONES

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Introduction

Exercise-related changes in geometry reflect bone's ability to respond to changes in its loading environment. Studies (Matsuda *et al.*, 1986; Woo *et al.*, 1981) report increases in cortical cross section in weightbearing bones following exercise training. These findings may have been growth-related, however, as their studies used immature animals. Additionally, no studies have compared exercise-induced responses between weightbearing and nonweightbearing bones. The present study addressed both of these issues by investigating exercise-related changes in mechanical properties and geometry of upper and lower extremity bones in skeletally-mature roosters.

Materials and Methods

Eleven skeletally-mature (> 22 wks) Bantam roosters were used: six served as age-matched controls and received no exercise, and five ran on a motorized treadmill 1 h/day, 5 d/wk for 9 wk at 80 - 85% $\text{VO}_{2\text{max}}$. All animals were housed in caged runs (1.5 m x 6.5 m x 2.0 m). Animals were killed after nine weeks with CO_2 hypoxia. The tarsometatarsus (TMT) and radius were dissected, cleaned of soft tissue, hermetically sealed, and frozen (-20°C) for a maximum of 2 wk prior to testing.

Left TMT and radius were loaded to failure in three-point bending using a mechanical-testing system (Instron 1122) and microcomputer (IBM AT). Loads were applied at the middiaphysis (anterior-to-posterior) with the bone submerged in a circulating buffer solution (50 mM phosphate buffer, pH 7.4, 37°C). All tests were performed at a crosshead speed of 25.4 cm/min. Using a low-speed slitting saw (EMCO Unimat-3, 130 rpm), a 1-mm middiaphyseal cross section was cut from each right TMT and radius. Slides were taken of each section (Ektachrome 100 ASA, Olympus OM2S, 50 mm flat lens, Asanuma auto-bellows) and rear-projected (x 12) to quantify (Numonics 1220 and Apple IIe) cross-sectional geometry. A modified version of SLICE (Nagurka and Hayes, 1980) calculated second moments of area, centroid position, periosteal and endosteal diameters and circumferences, cortical and medullary areas, and regional cortical thicknesses. Combining geometrical and load-time data, a computer (IBM AT) program calculated proportional-limit, maximum, and failure loads; tensile linear stress, nonlinear displacement, energy at the proportional (linear), maximum, and failure loads; stiffness, and elastic modulus.

Results

Average body weights of runners (965.5 g) and controls (979.7 g) did not differ significantly. TMT total cross-sectional area differed only 6% between runners and controls, while the total cross-sectional area of the runner radius was significantly greater than controls (32%). Runner radius second moment of area about a medial-lateral axis averaged 66% greater than controls, while the same parameter for the TMT differed only 10% between the two groups. Load at the proportional limit for runner TMT exceeded that of controls by 23%, while the radius differed by only 14%. Although not statistically significant, runner TMT showed a 44% greater elastic modulus than controls, while the radius differed by less than 5%.

Table 1. Geometrical and Structural Properties of TMT and Radius

	R-TMT	C-TMT	R-RAD	C-RAD
Cortical Cross-sectional Area (mm ²)	16.1 ± 4.6	16.3 ± 3.9	5.5 ± 1.8	4.3 ± 1.2
Maximum Load (N/BW)	0.35 ± 0.08*	0.26 ± 0.05	0.05 ± 0.00	0.05 ± 0.00
Energy at Maximum Load (N·s/BW)	0.15 ± 0.05*	0.09 ± 0.04	0.04 ± 0.01	0.04 ± 0.01
Failure Load (N/BW)	0.30 ± 0.04*	0.18 ± 0.06	0.05 ± 0.00	0.05 ± 0.00
Energy to Failure (N·s/BW)	0.24 ± 0.03*	0.32 ± 0.04	0.05 ± 0.02	0.06 ± 0.01

Values are means ± S.D. Structural properties are expressed per unit body weight.

R-TMT (n = 4): runner TMT; C-TMT (n = 6): control TMT; R-RAD (n = 3): runner radius; C-RAD (n = 5): control radius.

* denotes significance between runner TMT and control TMT, $p \leq 0.05$

Discussion

Responses to strenuous running differed between weightbearing and nonweightbearing bones in avian. TMT remodeled such that the structural properties increased without significant changes in bone geometry, while radius geometry changed significantly without significant changes in structural properties. Differences between the two bones may reflect varying remodeling rates, with weightbearing bones being more mature and more mineralized than nonweightbearing bones. If loadbearing roles of the radius did not differ between exercise and control animals, then remodeling in the nonweightbearing bones may reflect a generalized systemic response to the strenuous exercise.

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EXPERIMENTAL MEASUREMENTS OF PROXIMAL TIBIAL DISPLACEMENTS UNDER LOAD: COMPARISON WITH FE MODELS. A. Dujovne and H.W. Wevers, Dept. of Mech. Eng., Clinical Mechanics Group, Queen's University, Kingston, Ont. Canada, K7L 3N6.

INTRODUCTION

The behaviour of a tibia under loading during locomotion is an important factor in designing an artificial tibial knee component. To develop a more anatomical design, it is necessary to know the actual deformation of the proximal portion of the human tibia. In vivo measurements are difficult to achieve. Thus, in vitro measurements or finite element (FE) predictions are more prevalent. In this case, a clamping and a load transfer mechanism, as well as a measurement methodology, were designed to allow comparison to our existing F.E. model (1).

MATERIALS AND METHODS

Five male and female tibiae, screened to exclude systemic or local diseases, were obtained from routine autopsies. Average age was 71 (from 59 to 85). The tibiae were frozen at -20°C . Prior to testing, the bones were thawed at room temperature for 12 to 16 hours. All soft tissues were carefully removed with a scalpel except for the articulating cartilage. Proximal tibiae length was 100mm. The geometrical longitudinal axis of each specimen was determined as the midpoint between the lateral and medial eminences (2) and the center of the distal shaft/cross section. Using a special locating C-clamp, a hole, 6mm in diameter, was drilled concentric to the long axis. The specimen was then placed in a mould for embedding the distal end in fiber glass reinforced resin, 10mm thick. The specimen was then clamped in the supporting frame that defined a rigid boundary condition at the distal end of the tibia. Steel measuring landmarks were positioned at the 12 mm level below the condylar surface (Figure 1). The load transfer mechanism consisted of a cylinder-bar-ball configuration which allowed free mediolateral motion of the condyles. Load was transmitted by means of composite washers to achieve a physiological load introduction. The tibia was then placed in a testing machine (Model 1122 Instron) and compressive load steps of 200 N were applied at a rate of 0.5mm/min minutes up to a maximum of 2000 N. Loads were applied centrally on each trial. Values of axial displacement were measured at four points (two anteriorly and two posteriorly) using dial gauges (resolution 0.005 mm). The axial displacement at 2000 N for each specimen was calculated as the average of the four values. The mediolateral expansion was measured at two points using dial gauges. For each specimen, actual expansion was calculated as the difference between readings of the gauges at 2000 N. Pictures of each specimen were taken and the contours of the condyles were digitized using AUTOCAD software (AUTODESK CO). Both medial and lateral condylar areas were calculated. The total load bearing area of the tibia was used to normalize results respect to the individual geometry.

Results

The load-displacement relationship for central axial loadings showed a linear elastic response. The real values of axial displacement and mediolateral expansion, as well as the normalized results, are shown in Table 1. Specimen 3 shows lower values as a result of a disease or malformation not detected by visual inspection. Therefore, specimen 3 was not taken into account in calculating the average.

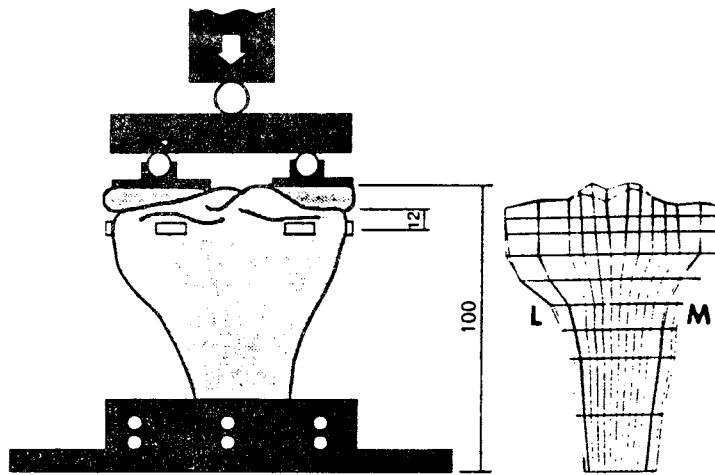


Figure 1: At left, proximal tibia showing load transfer mechanism, measuring landmarks and supporting frame. At right, finite element model.

DISCUSSION

The purpose of this work was to find an experimental method for in vitro displacement testing of human tibiae. The theoretical and experimental values have a close correlation in the axial displacements, whereas values of mediolateral expansion indicate an overestimation by the FEM. The discrepancy in mediolateral expansion may be explained by not accounting for the anisotropy of the cancellous bone and other simplifications that were justified based on the application of the model. It was also demonstrated that in vitro testing is helpful in the design process of prostheses.

ACKNOWLEDGEMENT

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SPECIMEN NUMBER	1	2	3	4	5	AVG.@	FEM
AREA [mm ²]	1090.32	1580.64	1548.38	1348.38	1232.25	1312.9	1794.86
AXIAL DISPLACEMENT [mm] *	0.140	0.120	0.050	0.140	0.165	0.140	0.095
MEDIOLATERAL EXPANSION [mm] *	0.105	0.025	0.040	0.065	0.100	0.075	0.159
AXIAL DISPLACEMENT NORMALIZED [mm] Δ	0.085	0.105		0.110	0.125	0.105	0.095
MEDIOLATERAL EXPANSION NORMALIZED [mm] Δ	0.065	0.020		0.050	0.075	0.050	0.159

Table 1: Real and normalized data (* at P = 2000N, Δ at P/A = 1.114 N/mm², @ excluding specimen 3)

CONTRIBUTION OF COUNTER MOVEMENT TO JUMPING PERFORMANCE: A
COMPARISON OF GYMNASTS VERSUS SWIMMERS

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The elastic energy stored in muscle and tendon during negative work can be utilized in the positive work phase so that may increase mechanical work and efficiency in ballistic movements¹⁾²⁾, and the extent of utilization vary with different types of movements³⁾. In the present study the extent of contribution of counter movement to jumping was compared between different types of well-trained male athletes: three gymnasts and three swimmers. A vertical jump without bending the knees appreciably, i.e., a toe-jump, was performed in the following two methods: (1) repetitive toe-jumps with counter movement(CJ), and (2) single toe-jump without counter movement(NCJ). Trials were made to jump up maximally. Ground reaction force was recorded by a force plate, and movements were filmed by 16mm cine camera. In addition, EMG recordings of gastrocnemius and soleus muscles were filtered, then rectified and integrated(iEMG). An averaging computation was also made. Mechanical work was calculated from potential energy change of the body.

The jumping height of the gymnasts was always greater (+60% in the mean) in CJ than in NCJ. In swimmers, however, the jump heights in CJ were about the same or sometimes even less than in NCJ. No significant differences in iEMG were observed between two groups both in CJ and NCJ. As an index of efficiency, so called 'electro-mechanical efficiency' ($EME = \text{work} / iEMG$) was calculated⁴⁾. The gymnasts exhibited appreciably greater EME in CJ, whereas no difference was observed between two groups in NCJ (Fig. 1). If assuming (1) a linear relationship between iEMG and energy cost⁵⁾ and (2) the efficiency of NCJ as 25%, the estimated efficiency in CJ was 41% in the gymnasts and 29% in the swimmers. These facts suggest that different athletic training exercises, as involved in gymnastics and swimming, may produce specific effects on the capability of reusing elastic energy in an antigravitational movement like jumping.

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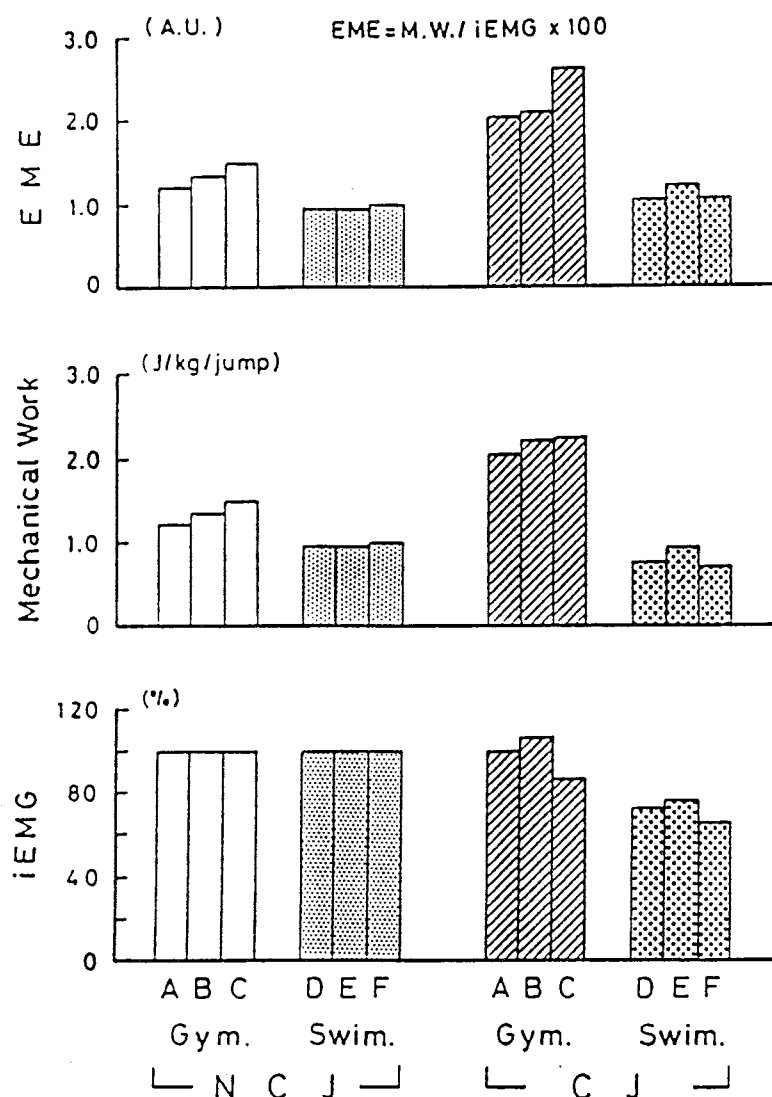


Fig. 1. Mechanical work (M.W.), iEMG, and EME (= M.W./iEMG) during non-counter jump (NCJ) and counter jump (CJ) were compared between the gymnasts (GYM) and swimmers (SWIM). Note that the gymnasts showed greater work and EME in CJ.

JOINT TORQUE PRODUCTION AND EMG ACTIVITY DURING SPRINT
RUNNING OF SPRINTERS AND DISTANCE RUNNERS

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To understand the contribution of muscle activities in generating joint force during sprint running, the EMG recordings were made with joint torque analysis. Ten track athletes consisting of 5 sprinters and 5 distance runners performed maximum sprint running. Through the measurements of ground reaction force by a force plate and the limb movements by a 16mm cine camera (100 fps), the torque, power, work around the hip, knee, and ankle joints were calculated for the entire period of both stance and recovery phases according to Winter's method¹⁾. In addition, the EMG recordings for seven leg muscles were made on two selected subjects.

The running speeds of sprinters and distance runners were 9.1 m/s and 7.8 m/s, respectively. The sprinters exhibited greater joint torque and power than did the distance runners. In the work per one cycle the sprinters have also shown to be greater, particularly in the following work phases: the positive work by the hip flexors and negative work by knee flexors during the latter half of recovery phase, and the negative work by ankle extensors during the initial part of stance phase. Marked EMG activity by m. rectus femoris at the first part of recovery phase (Fig. 1, phase a-b) might indicate a contribution of accelerating hip flexion and decelerating knee flexion when considered torque development. Activity of m. biceps femoris towards the end of recovery phase (phase d-e) suggests that this muscle contributes to accelerate hip extension and to decelerate knee extension, so as to prepare for pull-down of the foot. During stance phase, hip extension torque was associated with a burst activity of m. gluteus maximus followed by m. biceps femoris, and ankle plantar flexion first by m. gastrocnemius, then by co-contraction of the soleus muscle (phase f). Zero torque in the knee joint, which appeared at the mid-point of recovery phase (phase c), was thought to be due to an energy inflow from upper segments, since no EMG activity was observed.

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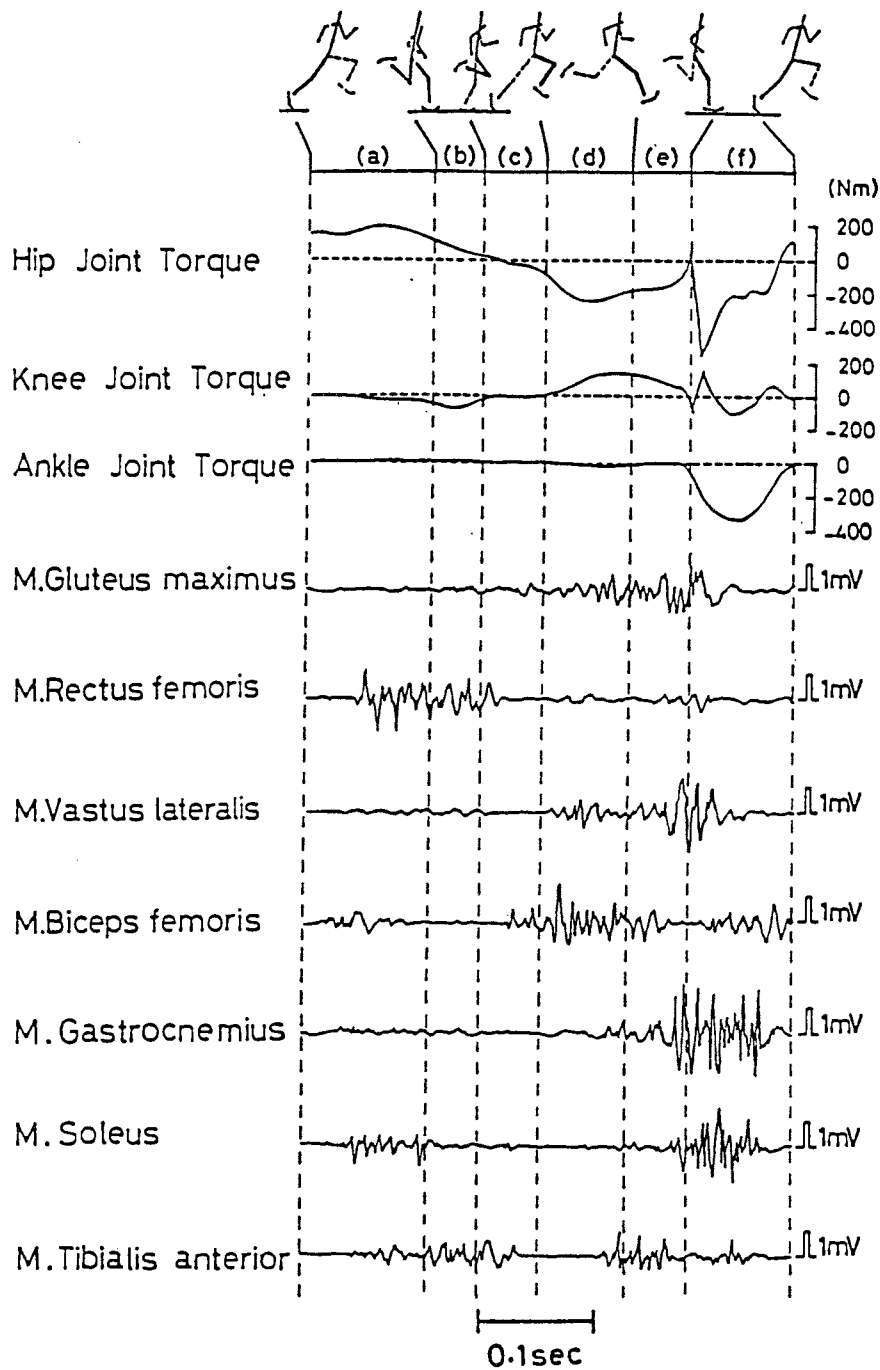


Fig. 1. Joint torque and EMGs in the lower limb during maximum sprint running (Subject: sprinter)

EFFECTS OF SPEED ON THE BIOMECHANICS OF FREESTYLE IN SWIMMING

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Introduction :

The application of electromyography (EMG) in sport research is recent (Gowitzke, 1984). In swimming, previous studies (Clarys 1983, Nuber et al 1986) showed essentially the role of muscles in the crawl stroke. In other ways, kinematic data were employed in the evaluation of swimming skill (Craig et al 1985, Schleihaut et al 1986). However, little information is available on the relationships between kinematic and EMG. The purpose of this study was to investigate the possible relationships between muscular activities, kinematic parameters and swimming speed by taking account of different data as sex, stroke and level of performance.

Methos :

Fifty freestyle swimmers (30 males, 20 females) with an average age of 18.1 years were studied. The subjects were divided into three groups (local, national, international) according to differences in their 100 m records.

Each subject performed 2X25 meter swims at four different speeds. The given velocities were 75% of the swimmer's 100m best performance, 85%, 95% and more than 100%. To minimize fatigue, a rest of 3 minutes between each swim was observed.

A video camera (25fps) was fixed in a waterproof case being moving alongside the pool in pace with the swimmer to obtain lateral views continuously during the test. Using a telemetric system (Rouard et al 1987); surface EMG were obtained from the deltoidus (pars anterior and medialis), biceps and triceps brachii, brachioradialis and flexor carpi ulnaris. Each amplified muscular activity was coded and summed all together before to be transmitting to a receiving device located along the pool. This summation was recorded during all the test and stored on a tape-recorder with synchronized signals of the video.

From the video, hip displacement, total time were calculated for each stroke which was divided into 5 phases from the arm - trunk angle : entry of the hand to 45°, 45° to 90°, 90° to 135°, 135° to the hand exit, recovery phase. The time of each phase was expressed in percentage of the total time of the stroke.

After rectified the raw EMG's, the area per unit of time was calculated for each stroke and each phase. To normalize the results, each mean temporal area was expressed in per cent of the maximum mean temporal area found during all the test for a given subject.

A principal component analysis was used to treat the relationships between EMG, kinematic data, speed and level of performance.

Results :

To increase the speed, swimmers increased the stroke rate in decreasing the times of the phases 1 (from the hand entry to a 45° arm - trunk angle), 4 (from 135° to the hand exit) and 5 (recovery). These time modifications were not linearly related to an increase of muscular recruitments.

The 6 studied muscles had their maximum activities for the lowest and highest speeds. Muscular recruitments were decreasing for the intermediate speeds. At these velocities (85%, 95% of the maximum speed on 100m), swimmers used more their legs. So their body was higher on the water and encountered less front frontal drag. In this case, upper arm muscles had less to work to propulse the body through the water.

Flexor muscles (biceps, brachioradialis, flexor carpi ulnaris) were more required especially during the phases 2 (arm - trunk angle from 45° to 90°) and 4 (from 135° to hand exit).

At the extreme speeds (75% and more than 100%), the selectivity of the muscles decreased and their recruitments increased considerably during phase 1 (entry of the hand to a 45° arm - trunk angle) and 5 (recovery). At these velocities, the other studied muscles were more active than during the intermediate speeds (85%, 95%).

These results had to be moderate with the swimmer's performance level.

For all the speeds, best performers were characterized by a greater hip displacement produced by a lower and more selective muscular activities.

Females had similar results with lower hip displacement and swim speed.

To summarize, the relationships between increasing speed and stroke rate seems to be linear but not dependent of upper arm muscular activities. Muscle recruitments were maximum at lowest and highest speed. During intermediate velocities, muscles were less required and had more selective activities.

This specificity of muscular contractions with the speed and level of performance could be considering in training process.

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COMPUTER PREDICTED OPTIMAL RELEASE CHARACTERISTICS
FOR VARYING ABILITY FEMALE JAVELIN THROWERS

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Optimal 2-dimensional release conditions (figure 1) for female athletes of varying release speed capability (v_n = maximal attainable release speed at a release angle of 35°) were predicted using a computer flight simulation program incorporating measured aerodynamic data from a T I Apollo ladies Aerodyne DR javelin (see Best, 1988; Best and Bartlett, in press, for methods). Previous simulations have studied the men's event only. Of the 2-d release parameters described in figure 1 only a_o , b_o , w_o and v_o need be considered as the most relevant athlete specific variables (e.g. Hubbard and Alaways, 1987).

Results and Discussion

Maximum range capability (R_{max} in metres) was predicted to increase as an essentially quadratic function of v_n (m/s) viz;

$$R_{max} = 0.95 + 0.283v_n + 0.0860v_n^2 \quad (1)$$

Differentiating equation 1 gives an insight into athlete improvement potential such that $dR_{max}/dv_n = 0.172v_n + 0.283$. This relationship is similar to that found for men's new rules javelins (Best, Bartlett and Sawyer, 1989) and reinforces the importance of release speed to the athlete with improvement potentials of over 5m per 1m/s increase in v_n possible for top throwers.

Figure 2 emphasizes the quadratic dependance of range on v_o , observing a best fit second order polynomial regression viz;

$$R_{max} = 0.929 + 0.0466v_o + 0.0956v_o^2 \quad (2)$$

It is not surprising, in observation of figure 2, why high linear correlations have been found in the previous literature despite the well documented quadratic dependance of projectile range on release speed.

Where v_n needs to be maximized, the parameters a_o , b_o and w_o were all found to be optima variables. Figures 3, 4 and 5 show how optimal a_o , b_o and w_o respectively vary as a function of v_n . Each figure shows important implications for coaching differing ability athletes. The positive slope relating v_n and optimal a_o is similar to that found by Best, Bartlett and Sawyer (1989) for the mens event. Slightly positive optimal b_o values for the very elite female thrower becomes negative for $v_n < 28$ m/s. Indeed, the discontinuity in figures 4 and 5 may be caused by a similar dual peak phenomena described by Hubbard and Alaways (1987). Optimal b_o and w_o appear to be negatively related, possibly a compensatory effect in attaining the best possible flight characteristics.

In conclusion, the results show predicted technique variations for differing ability female athletes. However, it should be noted that simulation programs incorporate methodological assumptions that limit the findings (Best, Bartlett and Sawyer, 1989). For example, flat landings for the ladies javelin could not be simulated.

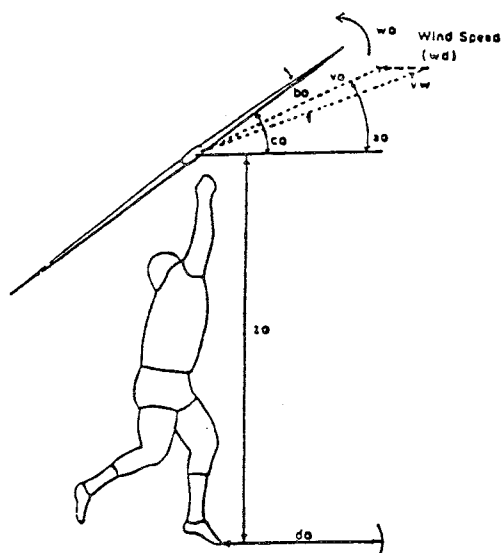
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Figure 1 Javelin vertical plane release parameters



- so = release angle
- bo = release angle of attack
- co = release attitude angle
- do = front foot to foul line distance
- vo = release speed
- vw = javelin speed relative to air
- wd = wind speed
- wo = release pitch rate
- zo = release height

Figure 2 Maximum range potential vs. Optimal release speed

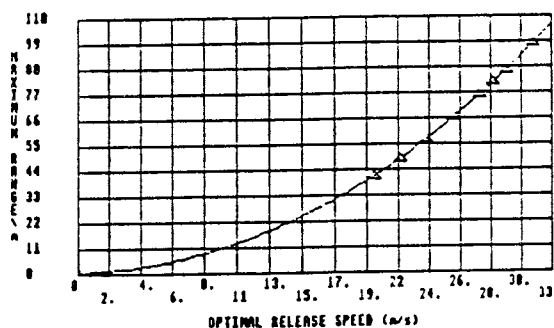


Figure 3 Optimal release angle vs. Nominal velocity

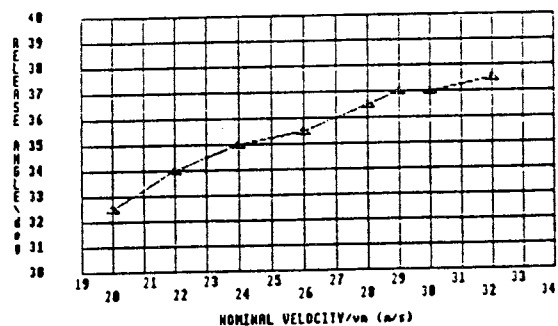


Figure 4 Optimal release angle of attack vs. Nominal velocity

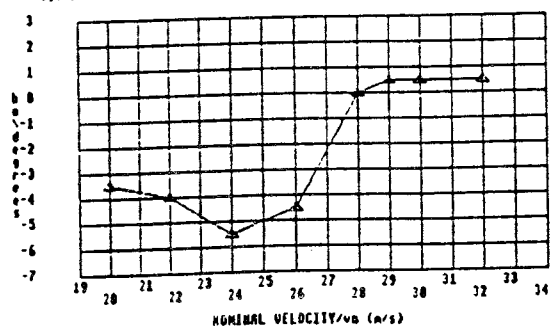
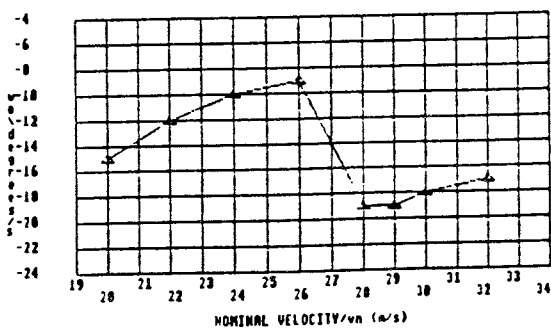


Figure 5 Optimal release pitch rate vs. Nominal velocity



COMPUTER PREDICTED OPTIMAL RELEASE CHARACTERISTICS
FOR VARYING ABILITY MALE JAVELIN THROWERS

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Optimal release conditions in the javelin event can be found if the flight of the implement can be accurately simulated. 3-d variables such as flutter, spin and yaw are assumed to be of secondary importance in terms of range (see Bartlett and Best, 1988) simplifying the problem to 2 dimensions (figure 1). Of the remaining parameters only ao , bo , wo and vo need consideration as athlete variables (assuming still air and using methods described in Best, 1988). This paper predicts optimal 2-dimensional release conditions for male athletes of varying release speed capability (vn = maximal release speed generated at a release angle of 35°) using computer flight simulations of a TI Apollo men's new rules javelin. Previous simulations for this javelin utilized speculative data only (e.g. Hubbard and Alaways, 1987). An important assumption of this and previous simulations is that vo , vn (in m/s) and ao (in $^\circ$) are related viz;

$$vo = vn - 0.127*(ao - 35) \quad (1)$$

Results and Discussion

Table 1 shows optimum release solutions for varying vn . Maximum range capability (R_{max} in metres) increases as a quadratic function of vn viz;

$$R_{max} = 1.08 + 0.437vn + 0.0823vn^2 \quad (2)$$

Equation 2 differs from Hubbard and Alaways' findings whose related discussion was misleading since R and R_{max} range are primarily determined by vo^2 and not vn^2 . Plotting R_{max} against optimal release speed (vo) correctly emphasizes the quadratic dependence of range on vo ;

$$R_{max} = 1.06 + 0.0808vo + 0.0954vo^2 \quad (3)$$

Hubbard and Alaways' "nearly constant" optimal ao , using speculative aerodynamic data, meant that vn and optimal vo observed similar relationships with R_{max} . The authors' data show optimal ao to vary considerably with vn . Relating to equation (1) this means that vn and optimal vo observe different relationships with R_{max} . Differentiating equation 2 predicts improvement potential since $dR_{max}/dvn = 0.165vn + 0.437$. This implies that for top class throwers ($vn > 28$ m/s) a 0.2 m/s increase in vn will result in > 1 m increase in range potential.

The variables ao , bo and wo exhibit inverted-U relationships with range (constant vn) rendering each an optima variable. Within present norms for competitive male javelin throwing, a greater vn has a corresponding higher optimal ao and bo . This has important consequences for coaching differing ability throwers since optimal ao for a world class male thrower and a developing young athlete (or top class decathlete) may differ by up to 5° .

For values of $v_n < 22 \text{ m/s}$ the sensitivity of optimal a_o to v_n increases such that for $v_n = 15 \text{ m/s}$ optimal $a_o = 24^\circ$. This characteristic, also found by Hubbard and Alaways, appears to be a function of the assumption made by equation 1. As v_n decreases the importance of the term $-0.127(a_o - 35)$ increases making lower, faster throws preferable (Hubbard and Alaways). This emphasises the importance of equation 1, and perhaps, a limitation by implying the same speed/angle sensitivity for all athletes ($-0.127 \text{ m/s/}^\circ$). The authors' simulation predicted negative optimal b_o values for all athletes and 10° differences in optimal b_o for differing ability throwers. Optimal w_o and b_o are negatively related and are the release conditions that together form the basis by which an athlete can attempt to optimise positive lift impulses during flight. At lower values of v_n , positive values of optimal w_o act to compensate the negative values of optimal b_o .

In conclusion, the results show possible technique variations for differing ability male athletes although 2-dimensionality, methodological assumptions and the lack of simulation validation, owing to the sheer complexity of problem, may limit the findings.

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Figure 1 Javelin vertical plane release parameters

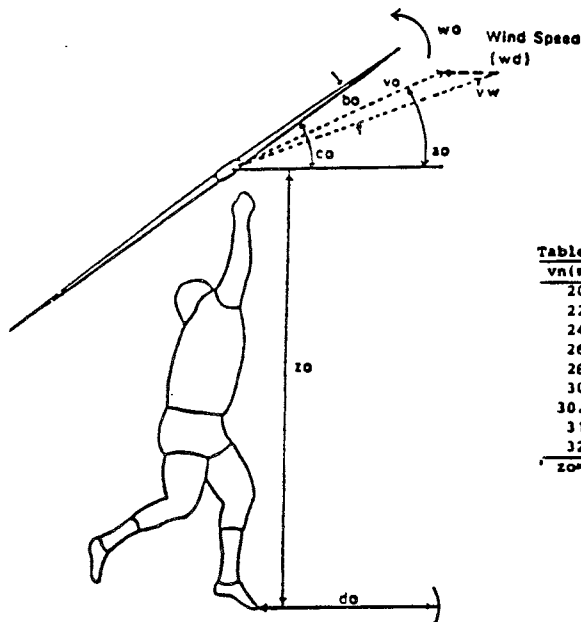


Table 1 Optimal solutions for varying release speed capability

$v_n(\text{m/s})^*$	$a_o(^{\circ})$	$v_o(\text{m/s})$	$b_o(^{\circ})$	$w_o(^{\circ}/\text{s})$	$R_{\text{max}}(\text{m})$
20	30.5	20.57	-11.5	18	43.06
22	32.5	22.32	-10.0	10	50.54
24	33.5	24.19	-8.0	7	58.83
26	34.5	26.06	-6.0	3	67.88
28	35.5	27.94	-5.0	-2	77.67
30	36.5	29.81	-3.0	-8	88.20
30.48	36.5	30.29	-2.5	-8	90.85
31	37.0	30.75	-2.0	-10	93.77
32	37.5	31.68	-1.0	-12	99.56

* $z_o = 2\text{m}$, $w_d = 0 \text{ m/s}$

- a_o = release angle
- b_o = release angle of attack
- c_o = release attitude angle
- d_o = front foot to foul line distance
- v_o = release speed
- v_w = javelin speed relative to air
- w_d = wind speed
- w_o = release pitch rate
- z_o = release height

MODULATION OF THE GASTROCNEMIUS AND SOLEUS H-REFLEXES DURING HOPPING

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INTRODUCTION

In studying the cat's fast (lateral gastrocnemius, LG) and slow (soleus, SOL) synergistic ankle extensors during paw shaking, Smith et al. (1980) demonstrated selective recruitment of the fast LG muscle without concomitant recruitment of the slow SOL muscle, possibly due to the time constraints imposed by the rapid movements during which the recruitment of slow muscle would be incompatible with the demands of the movement. We have recently reported some evidence of phase dependent and preferential activation of the relatively "fast" gastrocnemius muscle (as compared to "slow" soleus) with increasing demands of force and speed during different types of hopping in man. To further shed some light on this matter, the present study examined the underlying spinal level neural modulation of the two functionally specialized muscles (gastrocnemius and soleus) within the ankle extensors during different types of hopping movements which required a wide range of force and speed of contraction.

METHODS

Due to the nature of the experiments (see below), only an adult male subject volunteered for this study. The subject performed three different types of hopping movements on three separate days with the resultant mean hopping height and mechanical power being 26.7, 6.2 and 1.7 cm and 2011, 654 and 285 W for SMALL, LIGHT and MAX hopping, respectively. The detailed electrical stimulation and recording procedures have been described elsewhere (Moritani et al., 1985). Miniature size bipolar surface electrodes were attached over the motor point area of the medial gastrocnemius (MG) and soleus (SOL). To determine the H-reflex amplitude at various phases of hopping, "phase-dependent" averaging technique was employed by a computerized data processing system. Owing to the difficulty of isolating the H-reflex potentials from the background EMG activities, particularly during the ground contact phase, a minimum of 20 to 30 signal averaging was necessary to assure an accurate H-reflex amplitude determination. Furthermore, to obtain fine time resolution of the H-reflex changes during the entire hopping cycle, measurements were performed at every 10 to 30 ms time bins to cover the entire hopping cycle. To achieve these measurements, the subject performed more than 400 to 500 hopping for each type of hopping, with a maximal number of hopping never exceeding 10 times (5 for MAX) per trial. A minimum of 2 min rest was provided between trials to avoid muscle fatigue.

RESULTS & DISCUSSION

Figure 1 represents a set of computer output showing the force, tibialis anterior (TA) and medial gastrocnemius (MG) rectified EMGs together with the corresponding MG H-reflex amplitude changes during various phases of MAX hopping. Results indicated that the H-reflex amplitude (H_{exp}) from the

MG showed progressive and significant increases over the resting control (Hcont), i.e. the peak H-reflex amplitude increased approximately 3.8, 4.2 and 5.0 times during SMALL, LIGHT and MAX hopping conditions, respectively. The corresponding changes for the SOL were 4.2, 3.1 and 1.6 times, respectively. Cross correlation analyses revealed that the H-reflex amplitude changes preceded the force curves approximately 45, 60 and 70 ms during SMALL, LIGHT and MAX hopping. Interestingly, Capady and Stein (1987) have demonstrated that H-reflex amplitude (a spinal reflex of largely monosynaptic component) of the soleus increases progressively during the stance phase and reaches its peak amplitude late in the stance phase during walking. During running, however, the H-reflex is found to be significantly smaller than during walking, suggesting a modified spinal reflex gain for the different functional requirements of the motor behavior. Our data are in agreement with these results and suggest that spinal reflexes are not stereotyped motor patterns, but can be specifically modulated for different functional requirements of the muscles.

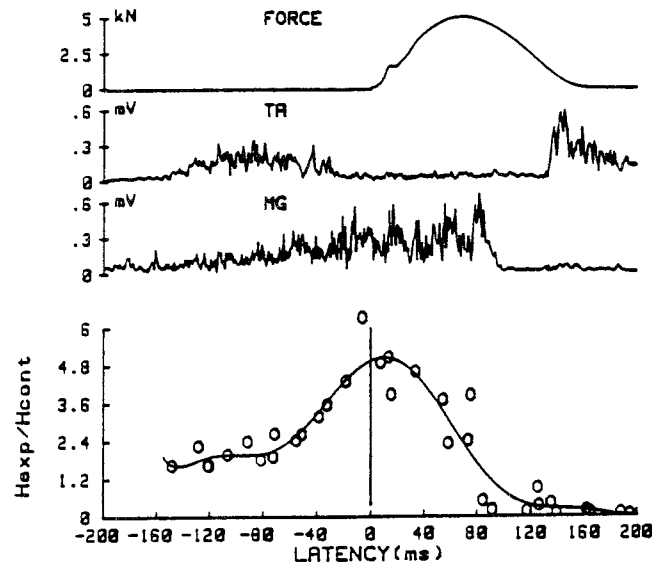


Fig. 1. A typical set of computer output showing the time course changes of force, TA and MG rectified EMGs together with the MG H-reflex amplitude being normalized by the standing rest control value (Hcont).

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MODELLING OF PROCESSES OF IMPACT IN SPORTS

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1. Introduction

In this study a computer program was developed which represents one module within the general model of sports movements (Nicol, 1987). This general model contains seven phases concerning the athlete and three phases concerning the device. It is constructed to simulate all kinds of sports movements. The computer program within this study represents the second phase concerning the device, called "the impact of athlete and device". In this work an impact is defined as a short-time interaction between two bodies. Such impacts can be seen within all kinds of back stroke games like tennis for instance. It is the main purpose of this study to analyse the influence of friction on the process of impact, because the literature searched yielded no qualitative calculations on that topic.

2. Construction of the module

The elementary equations of impact are expanded so that the qualitative influence of the attribute of elasticity and friction are considered in the equations too. There are three forms of friction which can occur within a given problem of impact: (i) merely sliding friction, (ii) merely adhesive friction or (iii) the combination of both with sliding friction coming first. When sliding friction occurs the tangential velocity of the partners of impact are different from each other. As soon as the tangential velocity of the partners are the same adhesive friction occurs. The computer program first decides whether the tangential exchange of momentum of a certain impact process is high enough to make the tangential velocity of the partners of impact equal. Merely sliding friction occurs if the tangential component of the exchange of momentum is lower than this critical rate, otherwise adhesive friction occurs as well. Merely adhesive friction occurs if the exchange of tangential momentum is so low that it is not able to surmount the adhesive resistance. After the computer program has decided which form of friction occurs the computer calculates the following three output parameters for one of the partners of impact: (i) velocity after impact, (ii) angular velocity after impact and (iii) angle of reflection. The computer does not calculate the same outputs for the other partner because these characteristics mostly are not relevant in sports.

3. Simulation of processes of impact in sports

After constructing and verifying the program various processes of impact were simulated in order to analyse the influence of some input parameters on the process. The process of simulation is conducted in that way that one input parameter is varied while the others remain constant. Among others the following

results were gained:

(a) First the impact between soccerball and ground was simulated. The mass of the ball is 0.425 kg, the velocity is 9 m/s, the angular velocity is 0, the angle of incidence is 45° and the coefficient of restitution is 0.7. The coefficient of friction is varied from 0 to 1.0. As long as the coefficient of friction is small enough so that the process of impact consists of sliding friction only, the velocity of the ball decreases linearly from 8.63 m/s (for $\mu = 0$) to 7.15 (for $\mu = 0.16$). In the same interval concerning the angular velocity decreases linearly from 0 to 5.2/s while the angle of reflection increases from 54.5° to 46.2° .

(b) As soon as adhesive friction occurs i.e. when the critical rate of $\mu = .17$ is exceeded, the outputs are independent on μ ; the velocity of the ball is 6.65 m/s, the angular velocity is 6.97/s and the angle of reflection is 41.9° .

(c) During the transition period between sliding friction and adhesive friction ($0.16 < \mu < 0.17$) the output parameters change considerably. The velocity of the ball decreases from 7.15 m/s to 6.65 m/s, the angular velocity decreases from 46.2° to 41.9° . The great difference of the outputs between sliding friction and adhesive friction has remarkable consequences for sports movements. Fast serving tennis players for instance prefer a smooth ground because during the process of impact the velocity of the ball does not decrease as much as on rough ground.

(d) the next simulation analysed in how far the occurrence of adhesive friction is dependent on the angle of incidence. For this purpose an impact between a soccerball and a ground is simulated again. The angle of incidence is varied between 0 and 90° , the angular velocity is (i) 0, (ii) 4/s, (iii) -4/s; the other outputs listed in (a) are not changed. The most interesting result of this simulation is the parabolical dependence of the occurrence of adhesive friction on the angle of incidence for all angular velocities. The critical rate of becomes zero, if the absolute value of the tangential velocity of the ball generated by the angular velocity and the absolute value of the tangential component of the velocity of the ball are the same and the direction of both are contrary directed.

(e) The velocity of the kicking leg on the output parameters within a given problem of impact in soccer was investigated. The velocity and the angular velocity of the ball are linearly dependent on the velocity of the kicking leg while the angle of reflection remains constant.

(f) Finally the influence of the excentric distance on the output parameters were calculated. The velocity of the ball is parabolically dependent on the excentric distance whereas the angular velocity and the angle of reflection are almost linearly dependent on the excentric distance until a critical is reached; up to this critical rate the process of impact consists of sliding friction and adhesive friction. If this limit is exceeded sliding friction occurs only and both outputs decrease considerably.

SIMULATION OF THE FLIGHT PHASE OF THROWING DEVICES

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Introduction

The present study covers the development of a mathematical deterministic model which simulates the flight phases of throwing devices. This model can be used as a modul of the "General Model for Sport Movements" (Nicol 1987), but it can also be an autonomous model. The model is subdivided into four groups, which depend on the characteristics of the device and the environment: (1) flight of a device independent of its characteristics and without regard to air resistance; (2) flight of shot or ball; (3) discus and (4) javelin, all with regard to air resistance. These four groups represents nearly all ball games and all athletics throwing disciplines. The first group serves for a standard of comparison, where the influence of air resistance can be investigated.

Construction of the model

The model is a mathematical deterministic one, that is why the trajectory of the flying devices is described with the aid of mathematical equations, which can be solved unequivocally. The system consists of five differential equations of first and second order which are derived from the work of TUTJOWITSCH (1976) and which are partly modified. So, the five unknown parameters height, distance, velocity, flight angle (angle between the direction of velocity and the horizontal) and angle of attack (angle between the longitudinal axis of the device and the horizontal) can be computed unequivocally at each moment of the flight. Because of their complexity the equations are solved by numerical integration. The simulation is implemented with regard to four aspects: (a) Investigation of the flight characteristics of athletics throwing devices. Change of velocity or of angle of attack of the javelin is calculated in relation to time, for example. (b) Investigation of the initial values for an optimum flight. It is the purpose to find out those conditions such as height, velocity and angle of release which guarantee an optimum flight phase. (c) Investigation of the exterior factors. The influence of exterior factors like wind, air density and gravity is calculated. (d) Investigation of the form of the devices. The form of the devices is not exactly determined, e.g. the length of the javelin varies from 2,60 m to 2,70 m. Therefore it is possible that a specific construction of a device is favourable for the given initial values.

Results

The investigation of the flight characteristics has shown that there are similarities between shotput and hammerthrow on the one hand and discus and javelin throw on the other hand. This can be illustrated by looking at the velocity time plot. For shot and hammer the velocity decreases after the release down to a minimum which is near to the angular point of the trajec-

tory. Then the velocity increases up to the value of the velocity of release. This behaviour is mostly caused by gravity. For the other devices the velocity decreases very strongly, it reaches the minimum after the angular point of the trajectory and the final velocity is lower than the velocity of release. That can be attributed to the fact that the angle of attack of discus and javelin is greater than the flight angle, so that a lift component is produced which compensates gravity by part. The investigation of the initial values has shown that the variation of the height of release has only a marginal influence on the throwing distance and so it can be neglected in hammer, discus and javelin throw. If the devices are thrown with a high velocity, the optimum angle of release for shot and hammer is found between 40° and 45° and for discus and javelin between 30° and 35° . There is a strong decrease of the optimum angle for decreasing initial velocity for shotput (30° - 35°) and an only marginal one for hammerthrow (40° - 42°). In contrast to these results the optimum angle of release increases for discus and javelin throw for decreasing initial velocity (35° - 40°). The optimum initial angle of attack for discus and javelin throw amounts -14° which is nearly independent of the initial velocity. The deviation from these optimum values is greater for high velocities than for lower ones. That means that a low deviation leads to a great loss of flight distance in highly skilled athletes.

The influence of realistic variations of gravity and air density is very low. The most surprising result is that high air density is favourable for an inefficient javelin thrower, whereas it prevents a throw of a good athlete. The influence of the wind is more important. If there is head-wind, the distance decreases for all devices except of discus. Strong following wind is favourable for shotput and hammer throw, whereas a low one supports the javelin throw. This is in contrast to the general opinion that a low head-wind favours the javelin flight. The influence of the wind for all devices is very strong for high initial-velocity, decreasing for declining velocity. The possibilities to vary the form of shot, hammer and discus are limited so that the effect of the variation can be neglected in general. Only the variation of length and local of the center of gravity of the javelin yields to interesting results. If an athlete is able to reach a high initial velocity he has to take a long javelin with the center of gravity located more at the back part. But for pupils it is better to take a shorter javelin with the center of gravity in the forepart.

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COMPUTER SIMULATION OF THE TWO-DIMENSIONAL FLIGHT PHASE OF AN ATHLETE

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Introduction

A computerized mathematical model was developed to simulate each planar flight phase of an athlete. The movements coming into question have to fulfill two conditions: (a) Only such actions of the athlete are accepted which have no link to the surrounding (ground, horizontal bar, ...), (b) All segmental actions have to take place in the sagittal plane. The following movements fulfill these two assumptions: Athletics: HAY-technique; pole-jump; long jump (triple jump), but there are clear bounds for the simulations of the pole- and long-jump, because of the angular momentum in respect of the longitudinal axis. This angular momentum is the consequence of the one-legged take-off and the arm motions, it cannot be neglected. Gymnastics: Somersaults and handsprings (forward and backward), dismounts of the horizontal bar and the asymmetrical bars. Swimming and diving: The start entry, somersaults (forward, backward, full gainer, dolphin).

Constructions of the model

The model exists of the two parts "human body" and "motion". The human model by ZATSIORSKY (1983) was modified to develop a 6-, 8- and 9-segmental model so that there is a suitable human model available for each simulation. The ZATSIORSKY regression equations are based on anthropometric data of athletes, they have the advantage that the mass and inertia characteristics (masses, moments of inertia and radii of the segments) can be determined by body height and body weight only. In contrast to the model by HANAVAN (1964) the ZATSIORSKY model has the disadvantage that the computer simulation only outputs a kinegramm but not a konturogramm, because there are no equations which calculate the distances between the longitudinal axis and the periphery of the segments. The number of the segments of the model used depends on the technique of the movement. So, the 8-segmental model shall simulate those movements where the extremities are parallel (e. g. somersaults). The equations of motion were derived from PREISS (1987). The user has to input mass and inertia characteristics, and length of each segment, angular momentum of the whole body, locale and velocity of the Center of Gravity and orientation of the lower part of the torso (angle of release) for initial values in addition to the angles of each joint at each time interval. The model was verified by the cinematography of the backward somersault.

Simulation

In this study the simulation was performed for squatted and jackknifed somersaults in diving. All jumps with one to four rotations and a good dip-in were studied. The dip-in was defined to be good if the angle between the lower part of the

torso and the water varies from 84° to 92° (foot-jump) or from -92° to -84° (header). The flight phase has been divided up into four parts (time intervals) with four defined different attitudes. The athlete has influence on the movement by variation of the initial values (angular momentum, L_0 , vertical velocity v_{yso} , angle of release φ_0) and by variation of the joint angles during the flight phase. The horizontal velocity only conditions the flight distance, but not the number of rotations and the dip-in. That is why it is always fixed at $v_{xso} = 200$ cm/s. First, the parameters of the take-off have been varied (the angular momentum varied from 20 to 68 kg·m²/s, the orientation of the lower part of the torso varied from 60° to 90° and the vertical velocity was 100, 200, 300, 400 cm/s). The joint angles at each moment have been defined according to the movement. The simulation yielded for each of the four vertical velocities those combinations of angular momentum and angle of release so that lead to a good dip-in. It turned out that the angular momentum of a successful jump is quite exactly fixed while the angle of release can vary in a greater interval. For instance, a 2 1/2-times forward jackknifed somersault is managed for $v_{yso} = 400$ cm/s if $L_0 = 61$ kg·m²/s and $\varphi_0 = 60^\circ$, but also if $L_0 = 63$ kg·m²/s and $\varphi_0 = 86^\circ$. With the given conditions a single squatted somersault is only possible for $v_{yso} = 100$ cm/s and $v_{yso} = 200$ cm/s. On the other hand, a 3 1/2-times squatted somersault is impossible with these vertical velocities. Also a 2 1/2-times jackknifed somersault is only possible for $v_{yso} = 300$ cm/s and $v_{yso} = 400$ cm/s. For the jackknifed somersaults with the same number of rotations the increasing of the angular momentum requires a greater angle of release. This tendency increases for increasing vertical velocity. There are more good combinations of angular momentum and angle of release for a jackknifed somersault than for a squatted one. Common to all, the angular momentum has the greatest influence on the motion followed by the vertical velocity and the take-off angle. Second, the joint motions have been varied for constant initial values ($L_0 = 45$ kg·m²/s, $\varphi_0 = 75^\circ$, $v_{yso} = 200$ cm/s). That means the time in which the athlete gets from one attitude to the following one is varied (e.g. from the take-off position to the jackknifed one). The simulation has shown that there are more possibilities regarding the change of the joint angles to execute a jackknifed somersault than a squatted one. The comparison of these two kinds of execution proves that the moment of inertia has a surprisingly influence.

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CHARACTERISTICS OF PRESSURE DISTRIBUTION WHILE SITTING ON VARIOUS KINDS OF SEATS

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Sitting has been of considerable interest under a biomechanical point of view during the last decades. Main points of interest being blood flow in the pressed tissue and the shape of the spine. As these internal characteristics are correlated with some external characteristics of the distribution of pressure on seat and back, measurement of pressure distribution seems a suitable procedure of gaining reliable results without applying inconvenient procedures to the body.

Measurement of pressure distribution under the feet has been developed to a powerful means during the last decade but the inflexible measuring platforms used by these authors are not suitable for application on soft seats. So a new sensor compound was developed by Diebschlag and Müller-Limroth (1981) and applied for assessment of pressure on the seat. As we needed even more flexibility for measurement even on mattresses and on pillows we developed capacitance type lowcost sensors sized 10 x 8 x 2 mm which were glued on fabric and wired with unshielded cable, so that an extremely flexible mat is attained (Saad and Nicol, 1988). Resolution is + 1 mbar, reproducibility + 2 %, linearity and hysteresis after software correction is around + 5 %. Output of data is provided by colored numbers, gray scale (figure 1) and colored 3D-display.

For evaluation two different methods are used up to now. In the visual method coarse results are gained by just looking at the display. In the second method quantitative parameters such as pressure maximum and size of area pressed are calculated, these parameters can be treated statistically. A third method was developed by us which provides parameters which are more representative, moreover a mean pressure distribution is gained. This can not be done by simply calculating the mean of the signals of each sensor with respect to all measurements because varying location of contact, body size and body weight cause high error rates.

In a first step, a reference pressure distribution is established. This can be done either by selecting a measured one or by constructing a distribution which for instance is considered to be ideal. In a second step every distribution to be evaluated is transformed mathematically so that the difference between this one and the reference becomes minimum. The transformations selected are shifting and rotating on the base plane and enlargement or reduction in respect to all three axes. This transformation is either applied to the whole pressure distribution or to parts of it. So a set of transformation parameters is gained which describes the

difference of the two pressure distributions to be compared i.e. the set describes the measurement in total in respect to the reference. In contrast, the parameters of the second evaluation method mentioned above describe just certain aspects. The parameters of the third method can be treated statistically; when a mean pressure distribution is gained.

Three office seats, a kitchen chair, a cafeteria seat, two relaxing seats and two car seats were investigated. Pressure distribution on back and seat were measured as well, using a 1024 sensor system. Position of sitting, seat height and back angle were varied, whereas just one subject was used for easing comparison. Main results gained are: (a) Only on the wooden kitchen chair the pressure distribution shows the shape which is usually expected, with pressure up to 500 mbar under the ischium and much lower pressure in the neighbouring areas. (b) The cafeteria seat is quite soft at the beginning, but after about 10 minutes the air is pressed out of the foam of the cushion to that the characteristic turns over to that of a wooden seat, reminding the user to leave. (c) The pressure distribution on office seats is heavily influenced by errors in adjusting the seat height. Similarly,

pressure on the hollow of the knee in car seats is extremely high, when the seat is pushed back to much for sportive driving position. (d) Some office chairs provide excellent lordosis support, so that upright sitting position can be maintained without using the upper part of the back and without extended muscular action. But most users do not sit back enough in order to take advantage of this support. (e) Several seats show extremely high pressure under the coccyx which is the consequence of concave shape of the seat under load. (f) A relaxing chair showed a nearly ideal pressure distribution though the cushioning layer was just 20 mm thick. Consequently, this construction offers several degrees of freedom for minimizing problems of temperature, humidity and hygiene too.

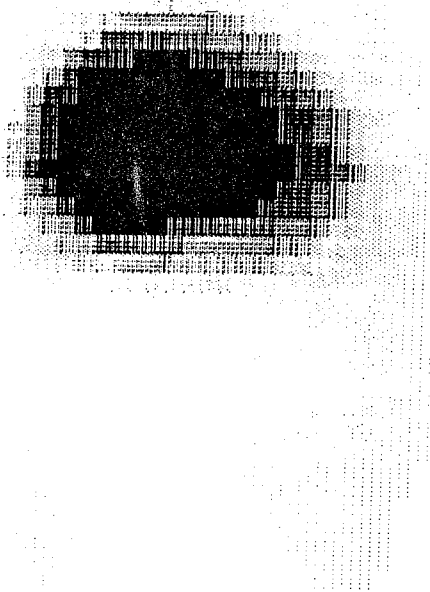


Figure 1: Pressure distribution on a car seat, displayed by means of gray scale

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TWO-DIMENSIONAL ASSESSMENT OF HAND MOVEMENTS

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In artistic gymnastics a contestant should combine an excellent manipulation of a requisite with dancing techniques. Few can fulfil requirement. Motion tasks differentiating candidates for this sports discipline were investigated. One of those tasks is the following of a model trajectory $y_o/x_o/$ in the OXY plane by the hand. The model trajectory makes possible the assessment of precision of movement under different conditions. The execution of a given task $y_o/x_o/$ with an assumed precision $\pm \Delta y_o$ in a predetermined time T and in spite of external disturbances is possible of neuromuscular coordination. The following of a model trajectory is a continuous stochastic process. If it is a stationary ergodic process, then autocorrelation function can be found for it. Then the number of model trajectory overruns in an assumed time T can be determined by the following formula

$$N = \frac{T}{\pi} \cdot \sqrt{\mathcal{L}^2 + \beta^2} \cdot \exp \left[- \frac{(\Delta y_o)^2}{2 \cdot A} \right] \quad /1/$$

where:

A - stochastic signal variance,

β - dominating harmonic of the stochastic signal,

$\mathcal{L}$ - a parameter characterizing the irregularity of the dominating harmonic.

Parameters: A, β , $\mathcal{L}$, $\pm \Delta y_o$, N, T can be used in the assessment of execution of the motion task formulated as following the trajectory $y/x/$ in time T.

Technical realization of this method requires an ultrasonic range-finder. The recording of hand's movements in the OXY plane requires the measurements of three quantities: l_1, l_2, l_3

which denote the distance of the hand's centre of mass from three predetermined points in the 3-D space. An ultrasonic transmitter is placed on the dorsal side of the hand.

The three ultrasonic receivers register the reproduction of the trajectory $y/x/$ of the hand's centre of mass in the OXYZ reference coordinate frame. Interfacing the range-finder to the computer enables the recording of a trajectory followed by the hand. This makes possible the automatization of measurements and processing data at the test stand named an Electronic Indicator of the Error of Movement. The essential feature of the device is a set of electromagnets exerting forces which disturb the course of the movement performed by the subject. The operational panel includes the area with a drawn path $y_0/x_0/$ programming the motion of a slider and four electromagnets affixed to the plate, which is able to rotate under the panel's lid. The subjects have been required to lead the slider so that the tip of the indicator traveled precisely along the path. The task has been performed with the upper extremity while in a sitting position. The 3D-UMA system /three dimensional ultrasonic motion analyser/ has been utilized in recording 2-D motion trajectories $y/x/$.

In order to show an application of the formula /1/ , the results, obtained during the investigation of a few subjects, have been compared. The compared subjects have been required to lead the slider within the path with approximately the same time. The paper contains the numerical assessment of their hand movements.

HAMSTRING LENGTH IN CROUCH GAIT

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INTRODUCTION

One of the most common abnormalities seen in cerebral palsy patients is crouch gait, in which the child walks with flexed hip and knee, and often with accompanying hip internal rotation and adduction. The etiology of the kinematic events of crouch has been the source of speculation, and includes excessive iliopsoas spasticity or excessive hamstring spasticity, and there have been several operative procedures designed to restore sufficient balance so that the patient may walk more upright. However, with modern 3-dimensional pre- and post-operative gait analysis it has become clear that not all patients with crouch gait have the same abnormalities, and not all will have favorable outcomes with the same surgery.

This paper describes a method of determining hamstring excursion during crouch gait, using 3-dimensional kinematic data and an anatomic model of muscle origins and insertions. When the variation in muscle length is correlated with dynamic EMG, distinct patterns of hamstring function emerge. The normal hamstring is a decelerator muscle, exhibiting eccentric contraction only; however, some patients with cerebral palsy exhibit prolonged concentric contraction during the entire stance phase of gait, so that the hamstrings are doing mechanical work to extend the hip. In these patients, a different surgical approach is warranted to preserve hip extension.

REVIEW AND THEORY

For decades investigators have recognized the critical interaction between muscles of the lower limb segments in crouch gait (Bleck, 1986). Of particular interest are two-joint muscles such as psoas, hamstrings, and gastrocnemius, since neurologically impaired children have more difficulty controlling them. Such muscles act to transfer energy between limb segments during walking (Yack and Winter, 1988), and in cerebral palsy may have contradictory actions. For example, the hamstrings may worsen knee flexion during swing and early weight acceptance, yet be responsible for assisting in extension of the hip against a spastic iliopsoas. Surgical lengthening of such a muscle improves knee extension, but worsens pelvic tilt.

Inconsistent treatment results have prompted investigators to analyse pre- and post-operative results much more closely using gait analysis (Sutherland and Cooper 1978). It is possible to quantitate the effects of surgery on the kinematic function of multiple joints. Such investigations have revealed a group of patients who have unexpected consequences of hamstring surgery.

METHODOLOGY

The length variation of four hamstring muscles (semimembranosus, semitendinosus, long head of biceps, gracilis) was calculated during gait, both pre-operatively and following selective hamstring lengthening surgery. Kinematic data were collected by 3-dimensional analysis of children with cerebral palsy, using a Motion Analysis Corp. ExpertVision system. The muscle lengths were calculated using a 3-dimensional skeletal model of origin and insertion, treating the muscles as straight lines. Dynamic surface EMGs were obtained pre-operatively over the major muscle groups, and timing was correlated with kinematic data so that EMG activity could be determined as a function of hamstring length.

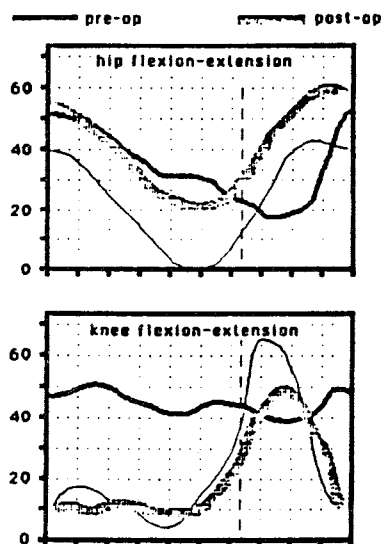


Figure 1. Pre- and post-operative hip and knee flexion following hamstring lengthening for crouch gait. The knee has improved, but hip flexion and pelvic tilt have worsened slightly from loss of the hip extension power of hamstrings.

RESULTS

Although there are minor variations among the muscles, for simplicity the semitendinosus muscle will be described.

During normal gait, the semitendinosus is longest (114% resting length) at foot strike, gradually passively shortening to resting length just before toe-off. EMG electrical activity occurs only during isometric or lengthening phases of muscle length, so that the muscle is always storing or transferring energy. No mechanical work is performed, since the muscle does not actively contract during shortening.

In subjects with neuromuscular disease, variations on this pattern were seen in dynamic muscle length, muscle excursion range, and EMG timing. Distinct patterns observed were:

- Type 1: Increased length, normal excursion, normal EMG. This was commonly seen in patients who had flexion deformities of the hip with crouch gait. Surgical release (elongation or weakening) will generally not worsen hip flexion. Hip flexor release will actually decrease the elongation of the muscle since hip deformity improves.
- Type 2: Increased length, normal excursion, EMG activity throughout stance phase of gait. These subjects also exhibited crouch gait, but some had mild contracture of the knee as well. More importantly, the hamstring activity indicates the muscle is doing work; in this case, the hamstring is extending the hip. Surgical elongation of the muscles will improve knee motion but worsen the hip flexion deformity unless significant hip flexor release is also performed. Transfer of hamstring to the femur (converting a two-joint muscle to a one-joint hip extensor) may be indicated in the group (Eggers 1952).
- Type 3: Increased length, decreased excursion, EMG activity throughout stance phase of gait. These subjects have hip and knee deformity and often internal rotation of the thigh as well. If release is done, hip flexion can worsen. If derotational osteotomy is performed, medial hamstring lengthening must be done to reduce postoperative internal rotation.
- Type 4: Normal or increased length, decreased excursion, continuous EMG activity. This pattern indicates marked co-contraction of hamstrings and knee extensors, and is usually not compatible with reciprocal walking.

Static testing also revealed that the hamstrings tightened more rapidly by hip flexion than by knee extension, an indicator of the relative difference in moment arms of the muscle at the two joints. The approximate ratio of these moment arms is 2.25 to 1.

SUMMARY

Hamstring muscles studied by a muscle length model and dynamic EMG during crouch gait show variations in mechanical function which may not be clearly defined by physical examination alone. In all patients with true crouch gait, the hamstring muscles are longer than normal (i.e., "stretched" or "tight"), but some patients exhibit patterns of concentric contraction that make surgical procedures risky. Simple surgical "lengthening" of the hamstrings is a misnomer, since the muscle may actually shorten if crouch is improved. If the mechanical function of hamstrings in subjects with abnormal neural control is recognized, alternative procedures (such as hamstring transfer) may improve functional results.

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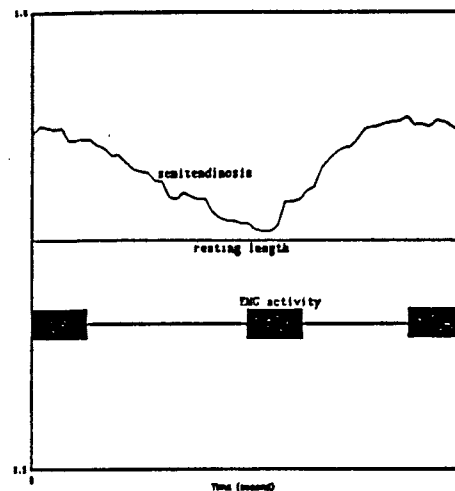


Figure 2. Hamstring length during normal gait, with EMG activity. The muscle always exhibits eccentric contraction, functioning to decelerate limb and storing energy.

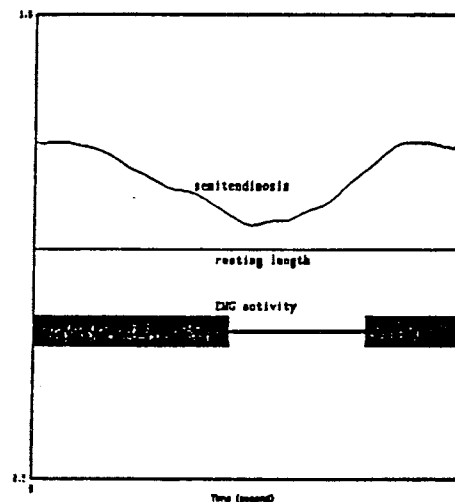


Figure 3. Hamstring length in patient with cerebral palsy and crouch gait. The hamstring is elongated with relatively normal excursion, but shows concentric activity throughout stance, so that it is doing work (extending the hip). This represents a Type 2 pattern described above.

Ball Impact Mechanics: Biomechanical Implications

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Spherical shelled balls comprise parts of the equipment of many games and sports events, including basketball, squash, tennis, soccer, handball and many others. According to the rules of play, the balls are often involved in collisions with fixed surfaces (floors or walls), hand held implements, and even unprotected body parts (including the cranium during heading in soccer). Biomechanical models of these impacts which seek to compute their effects on the body often rely on ad hoc assumptions regarding the magnitude, duration, and shape of contact forces (Brannigan & Adali(1981)). Clearly, however, errors in these assumptions will lead to errors in calculations and conclusions.

There have been several experimental studies which measured ball contact time, but most of these have been for the sport of tennis(Brody(1979), Baker & Putnam(1979)). No studies of a more general nature have been made which assume only that the ball is spherical shelled but not that it is a particular kind of ball, e.g squash or tennis. Nor, according to Frolich(1986), have there been any theoretical or experimental studies of what happens to the air pressure in a ball when it is struck by the foot or other body part. Thus, although needed, no general model exists for the impact mechanics of spherical shelled balls. This paper presents results of some recent experiments performed in support of the development of such a general model. We have simultaneously measured three variables during the impact phase; contact force, internal pressure and mass center kinematics and believe that our work is the first in-depth investigation of ball impact considering all three simultaneously.

Two elastic spherical shelled balls of varying mass and radius (see Table 1) were dropped from rest from a range of heights (0-4.65 m) onto a Kistler force plate(thus pre-impact velocity varied between 0 and roughly 6.5 m/s). The force was sampled by a PDP-11/23 computer at 1750 Hz. Ball internal pressure was also instrumented with a solid state pressure sensor (SenSym LX0503A) mounted just inside the top wall, whose signal was sampled via a light coaxial cable at the same rate. Finally the motion of the ball prior to, during, and after impact was filmed with a high speed (1000 Hz) video camera (Kodak Spin Physics Ektapro 1000). The pressure and force signals were synchronized with the video to within 1 ms using a photocell actuated switch which simultaneously perturbed the force trace and energized an LED in the camera field of view.

Coordinates of eight points on the image periphery were digitized before and after impact and used to compute mass center position and velocity. Thus it also was possible to compute the

commonly used 'coefficient of restitution,' the ratio of the post- and pre-impact mass center velocities. Coefficient of restitution was independently computed as the ratio of the force impulse during the rebound and compression phases of the collision. Typically these two agreed to within 6%, probably mostly due to the synchronization uncertainty of 1 ms.

Shown in Fig. 1 is a typical time history of the contact force (bottom) and the internal pressure (top) signals. Although the pressure is apparently much noisier than the force trace, both exhibit high frequency oscillations during the compression phase not found during rebound. The cause of these oscillations is believed to be internal pressure wave reflections. The apparently random pressure 'noise' is about an order of magnitude larger than the noise when the ball is at rest before release and is believed to result from actual pressure fluctuations in the turbulent wake transmitted through the wall. Although contact times for tennis balls of 4-5 ms have been reported (Brody(1984), Baker & Putnam(1979)), as the ball mass and compliance increase the contact time increases (to roughly 35 ms in Fig. 1). Contact time also decreases, for a given ball, with increasing inflation pressure and increasing pre-impact velocity.

A major conclusion of our experiments is that impact energy losses are much larger than the hysteresis losses measured in quasi-static compression tests. A general theoretical model, presently under development, should be able to predict this as well as the dependence of contact force shape, magnitude and duration as a function of the independent variables; ball mass, compliance, radius, inflation pressure, and impact velocity.

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Table 1		
Ball Parameters		
Ball	1	2
mass	1237 g	426 g
radius	50.6 cm	28.2 cm

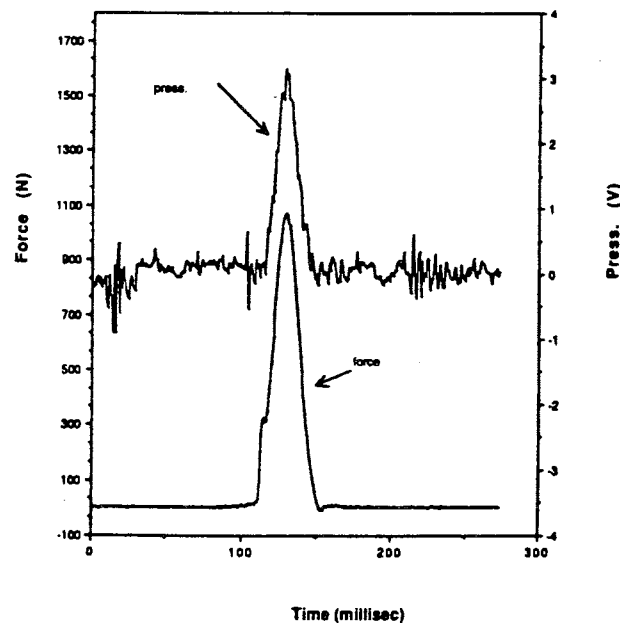


Fig. 1 Contact force and internal pressure during bounce of ball 1 from 4.65 m

FRictional FORCES AND TORQUES DURING A FOREHAND DRIVE AND A STANDARD LABORATORY TEST FOR DIFFERENT TENNIS SHOES AND PLAYING SURFACES

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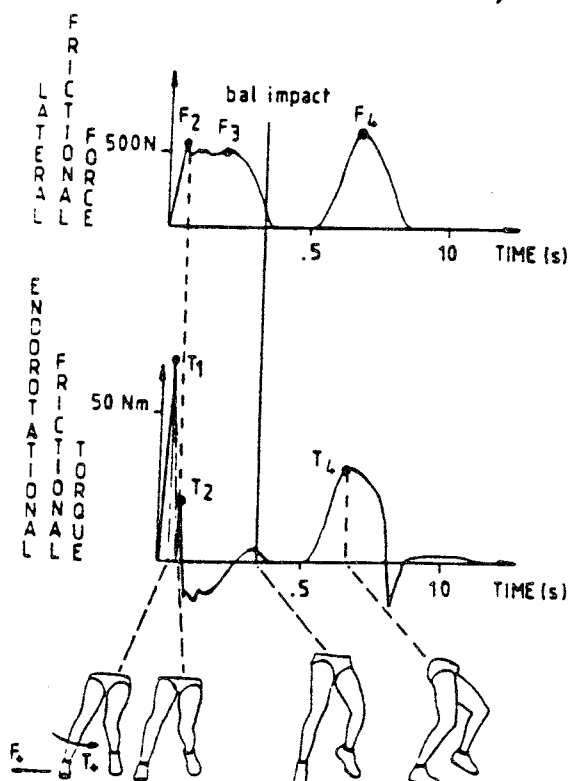
It is the aim of this study to compare the frictional behavior of different combinations of tennis shoes and playing surfaces, during the execution of a open stance forehand tennis drive in a fast up and down running action and during a laboratory test in standardized conditions.

Experimental techniques

Frictional data from twelve tennis players were acquired on a large Kistler force plate (60x90cm), built in at the base line of a real indoor tennis court. The following four different playing surfaces were tested: Tennis Fluid Texilast and Chevron 400 Fluid (a textile, respectively a rubber surface, both covered with structured rubber granules), Master Turf 20 SF (a sand imbedded turf) and Sportflex (an allround rubber surface for ball games). The players wore three different tennis shoes: the Nike All Court (a low cost all surface shoe), the Nike Wimbledon GTS (manufactured with a rubber outsole) and the Nike Royal (with a polyurethane outsole).

A high-speed film camera (running at 300 frames per second) was used simultaneously to document the force data.

A standardized test (Van Gheluwe et al., 1982), was used to evaluate the friction generated by the same shoe-surface combinations, in laboratory conditions.



Friction forces and torques during an open stance forehand drive

Results and discussion.

Analysis of the force plate data (mean and maximal values) from the forehand drive for the different shoes, reveals significant differences (at the .05 level) only for the lateral friction at first full foot contact (F2 in the figure). The higher friction values correspond to the shoes with a more pronounced sole profile.

Frictional torques, however, are not significantly affected by the differences of the selected shoes.

On the other hand, the execution of a forehand on different playing surfaces does generate varying frictional torques, and frictional forces as well. The "Fluid" surfaces, in general, display lower friction than the sand imbedded turf.

However, the Sportflex material (the only "dry" surface without loose granules or sand on top of it) presents a contradictory behavior. It significantly displays the highest frictional torques T2 and T4, but at the same time the lowest lateral frictional forces F1, F2 and F3, lower than the "Fluid" surfaces.

This can be explained if we assume the kinematics of the movement influencing or being influenced by the frictional interaction with the floor, as suggested by Nigg et al. (1988) and demonstrated by Stucke et al. (1984). This is supported by the fact that for Sportflex, contrary to the other surfaces, the frictional torque T4 during the push-off, has an opposite sign than shown in the figure, suggesting an alternative kinematic sequence.

As for the data from the laboratory experiments, no coherent pattern could be found from one shoe to another. This finding further decreases the global influence from the shoes on friction to a non-significant level.

However, differences in frictional force and torque, did show up again for the different playing surfaces. This time, the Sportflex surface displayed a consistent behavior. It scored the highest friction values, both for force and torque, especially for the unipodal position. As differences in kinematics are non-existent in the laboratory testing conditions, they cannot interfere with frictional behavior as they may do during "in vivo" experiments.

Conclusion

Ground friction, both during realistically executed and during simulated movements, varies according to the different characteristics of playing surfaces, rather than to those of actual sport shoes.

Surfaces from the "fluid" type are hereby displaying mostly the lowest friction values and, therefore, are likely to be less injury prone.

However, inherent kinematic adaptations during the life execution of a movement may prohibit the extrapolation of the results from friction tests in a laboratory environment, to real life conditions.

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A method for measuring small joint surfaces, and preliminary results for the metacarpophalangeal joints

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Introduction

A commercial three dimensional coordinate measuring device has been used to obtain data on the finger metacarpophalangeal (mcp) joints. The data gives information on joint size, shape and orientation with respect to bony landmarks. The method is applicable to other small joints. Previous studies of the mcp joints (Aleksandrowicz 1974, Unsworth 1971) have used two dimensional measurement in one or more planes to model the joint in terms of circular sections. Using three dimensional data it is possible to fit various functions to the surface, and separately analyse distinct regions of the joint surface.

Materials and method

Fresh or preserved bones, with intact articular cartilage, are used. In this series, preserved material only was used- a right hand, the index and middle fingers of a left hand and the index of a small right hand.

The bones are cast with their articular cartilage in a hydrated state. This does not exactly mimic the in vivo situation, but all joints are measured in a similar state, and as close to the in vivo condition as is reasonably possible. Fast curing dental silicone impression material is used to coat the articular surface and as much as is required of the bone's shaft. A plaster case is made around the coated bone. When set, this is opened, the bone removed, and the plaster and silicone case carefully reassembled. A resin is poured into the case and left to set in partial vacuum to form the cast.

Measurement is made on a three dimensional coordinate measurement device produced by ITP Ltd. A cluster of up to five Renishaw contact probes are used on the device. Data in terms of the machine's X,Y and Z axes is automatically held on a microcomputer. The bone to be measured is mounted such that point measurements may be taken of the bony landmarks, the perimeter of the joint surface and on the articular surface itself. Accuracy of casting, measurement and analysis was tested by casting a 1/2 inch (12.70 mm) diameter metal sphere on a rod, measuring both the original sphere and the cast and analysing the results.

The data is analysed on the university mainframe computer, approximating a polynomial to the data. In the case of the mcp joints, a sphere is fitted to the data for the surface, and the centre of the sphere defines the origin of a new set of axes. These axes are oriented using the data for the bony landmarks. A plane may be fitted to the perimeter of the joint surface to define its orientation.

Results

The metal sphere gave a radius of 6.33 mm, the correct value being 6.35. Casts of the sphere gave radii of 6.30 and 6.42 mm. A set of data for the sphere, covering a small arc, similar to that of the proximal phalangeal surface at the mcp joint, gave a radius of 6.23 mm, 0.12 mm in error. Data for a cast over a similar arc gave a radius of 6.21 mm.

In all joints studied the phalangeal base had a larger radius than the metacarpal head, results for one hand are shown in table 1. All surfaces fitted sufficiently well to spheres to discourage efforts to fit more complex shapes. The volar tubercles of the metacarpals were analysed separately and provided a less satisfactory fit to a sphere, but the results were used to give an indication of their size. They were linearly related to the metacarpal radii with a correlation coefficient of 0.53.

Discussion

The system is capable of accurate measurement, in this case the most important element of error arose in the fitting of a sphere to data for a very small arc, a problem common to most measurement techniques. The use of axes based on bony landmarks allows results to be related to information from biplanar radiography, other imaging techniques and drawings of sectioned bones, provided landmarks are accurately defined. Results for mcp joints agree with findings of other authors (Aleksandrowicz 1974, Unsworth 1979), in indicating that the joint surfaces are approximately, but not exactly, spherical. Measurement of further samples will allow conclusions on the possible relationship of the ratio P/M (table 1) to joint mobility, and should reveal further information on this important group of joints.

Table 1 Radii of mcp joints, in mm, for a right hand, with standard deviation in brackets[].

Digit	metacarpal head (M)	phalangeal base (P)	P/M
Index	8.84 [0.34]	9.49 [0.21]	1.07
Middle	8.83 [0.38]	9.10 [0.09]	1.03
Ring	7.37 [0.22]	8.80 [0.10]	1.19
Little	7.66 [0.18]	11.38 [0.16]	1.49

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REGULATION OF STROKE FREQUENCY AND MECHANICAL WORK
PER STROKE IN SPEED SKATING THE CURVES

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Due to the peculiar characteristics of the push off in speed skating (forward gliding skate and sideward push off) there are certain constraints in the combination of mechanical work per stroke A and stroke frequency f . According to the geometrical model (J. Biomech 21:445-450, 1988), f is a constraint frequency dependent on the velocity, work per stroke and radius of the curve, with P the external power ($A \cdot f$) required to follow the trajectory of the curve. High speed film and video measurements, taken during the 1988 Olympic Winter Games, yielded mechanical work per stroke A and stroke frequency f on the straightaways. With these values the stroke frequency f was predicted in the curve following the straightaway analyzed (mean values in table).

Results:	P (W.kg ⁻¹)	f pred (s ⁻¹)	f actual (s ⁻¹)	corr.coef.
mens 1500m	7.36	2.55	2.26	0.38 * (n=53)
ladies 1500m	5.11	2.27	2.26	0.60 * (n=43)

Results from the ladies races supported the model. The lower correlation for the mens races and the 11.4% lower actual stroke frequencies may be a result of a necessary change in technique (turning skate inwards at the end of the push off, manipulation of skating radius) resulting in a lower actual stroke frequency.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

CARDIOVASCULAR RESPONSES OF SEVERELY MENTALLY AND PHYSICALLY DISABLED PERSONS DURING LOWER BODY NEGATIVE PRESSURE

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** Asahikawa Children's Rehabilitation Center, Asahikawa.

*** Japan Women's College of Physical Education, Tokyo, Japan.

The Child Welfare Laws of Japan state that severely mentally and physically disabled persons are those who suffer from both severe mental retardation and severe physical disabilities (Japan Ministry of Health and Welfare, 1979). A considerable number of the disabled persons are lying down on bed during the entire day. Not only does the physical inactivity lead to a serious risk of decrease in the cardiovascular function, but it causes consequential diseases such as decubitus, edema and infectious disease. In order to sustain good function and health, it is necessary to involve them in training of cardiovascular fitness even by passively simulated physical work.

Lower body negative pressure (LBNP) provides a mean-to apply reduced pressure usually to lower body and simulates the effect of gravity (Blamick et al., 1988). The LBNP may be one of possible methods for stimulating the cardiovascular function. We evaluated cardiovascular function of severely mentally and physically disabled persons using the LBNP method and possibility of the training using the passive way was discussed.

METHODS : The subjects were ten persons from a medicare-provided residential institution for the severely multi-disabled. All the subjects lay down on bed during the entire day in the past and present. They had not opportunity of orthostatic posture such as sitting and standing in daily living. Ten healthy people participated for the control group.

The LBNP protocol used a stepwise reduction in pressure. Data were recorded during 5 min of a control period at supine rest, followed by a total of 20 min of the graded LBNP at -5, -10, -15 and -20 mmHg for 5 min at each level.

The heart rate was calculated using the ECG detected by precordial unipolar electrodes. The photoplethysmogram was used as an indirect measure of peripheral blood flow. It was derived from the left big toe. Power spectral density of the plethysmogram was computed every minute. Furthermore, power was determined by accumulating the spectral density at the frequencies of more than 0.5 Hz. The blood pressure of the right upper arm was measured every minute. In order to examine stationary values of the above parameters, the processing was performed on the signals for a period of three minutes from the third minute to the end at each level of the LBNP.

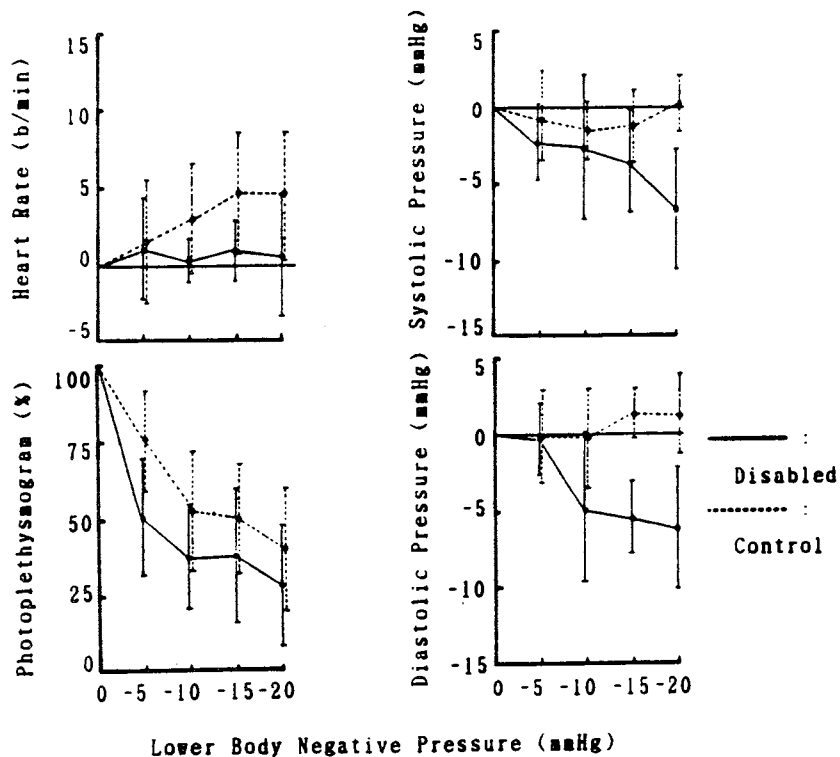
RESULTS and DISCUSSION : The LBNP from -5 to -20 mmHg produced graded increase in the heart rate with decrease in the plethysmogram for the healthy subjects as shown in the figure. Increase in the heart rate was significant at the LBNP levels of lower than -15 mmHg and was 4.8 b/min. The heart rate for the disabled persons did not change during the graded

LBNP. Depression of the plethysmogram for the disabled subjects was 12 to 24 % greater than for the healthy subjects. During the LBNP at -5 and -10 mmHg, there was not detectable change in both the systolic and diastolic pressures of the control subjects, whereas, during the LBNP at more than -15 mmHg, the diastolic pressure showed a tendency of increase. Both the systolic and diastolic pressures of the disabled persons decreased progressively during the graded LBNP. These variables fell by -7 mmHg at the level of -20 mmHg.

As previously investigated, the blood is pooled in lower extremities during the LBNP. Furthermore, peripheral vasoconstriction activated by baroreceptor-mediated reflex reduces peripheral blood flow. Consequently, the systolic pressure does not change and the diastolic pressure elevates. Heart rate increases to compensate decreasing stroke volume. Our results for the healthy subjects confirmed that the cardiovascular response to the LBNP agreed well with the previous reports. Inadequate control in heart and vasculature during the LBNP was observed for the disabled persons. This suggested serious lower function or dysfunction in both central and peripheral circulatory system of the severely disabled persons.

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The normalization of electromyographic signal.

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REVIEW AND THEORY.

Averages coupled with standard deviations have been used extensively in trying to normalize the EMG signal of relative effort (Milner et al. 1971, Knutson and Richards, 1979, etc.). The problem with these statistics is well known: they cannot be used efficiently in evaluation when faced with non-normal distributions, wherein their significance is at best ambiguous. When we evaluate an individual, we want to determine how different or similar his results are compared to the rest of a referred-to population. This in turn necessitates the availability of a criterion, a "norm", with which the comparison of the individual's results shall be done. To establish a norm, we propose using an order statistic called "tolerance interval" (Rohatgi 1976). The interval is delimited by two values, $x_{(r)}$ and $x_{(n+1-r)}$, that is the r^{th} lowest and the r^{th} highest among n observed results. The tolerance interval for a given function F with tolerance coefficient Y is a random interval for which the probability is Y that this interval covers a specific percentage $100P$ of the population: see formula 1. With this equation, we will evaluate the number of subjects needed for given coverages and probabilities.

$$Y = \sum_{i=0}^{n-2r} \binom{n}{i} P^i (1-P)^{n-i} \quad (\text{formula 1})$$

METHODOLOGY

First we will evaluate the normality and symmetry for some results of relative effort (Normand et al. 1986) by measuring their g_1 and g_2 statistics; second we will compute, through formula 1, the number of subjects needed for defining norms with different coverages and probabilities.

RESULTS AND DISCUSSION

Figure 1 represents the values of the standardized g_1 and g_2 statistics computed for Normand et al. 1986 data. Normal and symmetric data should obtain g_1 and g_2 around zero.

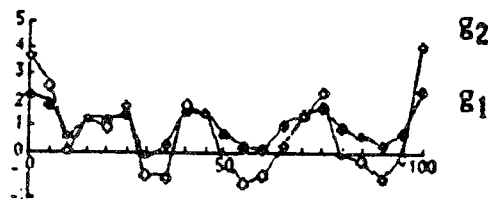


Fig.1 Results of g_1 and g_2 for gait.

Obviously, these results are not either normal or symmetric. Tables 1 and 2 represent the number of subjects needed for intervals with different coverages and probabilities.

TABLE 1

$\begin{matrix} P \\ R \end{matrix}$	.5	.9	.95	.99
1	8	46	93	473
2	12	76	153	773
3	18	103	208	1049
4	23	129	260	1312
5	28	154	311	1568

TABLE 2

$\begin{matrix} P \\ R \end{matrix}$	.5	.9	.95	.99
1	11	64	120	662
2	17	97	198	1001
3	22	127	259	1307
4	27	156	316	1596
5	33	183	371	1874

In the two tables, P is the wanted coverage and Y is the confidence that this coverage is indeed met by the population. Furthermore, intervals with $r=1$ are equivalent to the range of the obtained data; on the other hand, an interval with $r=3$ rejects the two smallest and the two highest values, a strategy that could protect norms against statistical outliers. Let us take for instance an interval with $Y=.95$ and $P=.99$ and $r=1$; this interval would be $(X_{(1)}, X_{(453)})$. If an individual obtain $X_i < X_{(1)}$ or $X_i > X_{(453)}$, we can conclude with an 95% confidence that this subject is in some way handicapped in the given EMG output. To be most accurate, with $P=.99$ $Y=.99$, we need to evaluate almost 700 subjects, a big enterprise. An alternative is possible if we consider a 95% error in the coverage and a 5% error probability. In that case, 153 subjects are necessary, which a more realistic venture.

CONCLUSION

The determination of a "norm" is definitely of a great import for evaluating individual capacity of patients with impaired muscles. With the help of tolerance intervals, we will be able to give an objective evaluation of results when distributions are not normal or symmetric.

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COMPARISON OF BACK AND LEG MUSCULAR UTILIZATION DURING A LOAD CARRYING TASK

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REVIEW AND THEORY

Load lifting techniques were extensively studied in the past (Chaffin and Andersson, 1984). However, in a normal setting, the individual must carry the object he just lifted, walk with it and, even, climb stairs. The "muscular cost" of such activities on the leg and back muscles is relatively unknown. One way is to determine the level of muscular utilization in terms of the percentage of the worker's maximal capacity (Normand et al., 1986). The purpose of this study is to compare the percentage of back and leg muscular utilization during gait and stairs climbing with and without an added load.

METHODOLOGY

Electromyograms were taken with surface electrodes fixed to the skin over the quadriceps and the erector spinae in 10 normal men. The EMG signals were pre-amplified (x 1000), rectified (full wave) and filtered (6 Hz) in order to obtain the linear envelope of the EMG signal. Kinematic parameters (angular displacements) were measured by means of an electrogoniometer. Foot contacts were recorded simultaneously with the EMG activity and used to determine the gait and stairs cycles. EMG references for the quadriceps were recorded during isotonic movements performed against a charge of 23 kg. For the erector spinae the EMG references were recorded during trunk extensions.

RESULT AND DISCUSSION

Figure 1 shows the muscular utilization of the erector spinae and the quadriceps during gait (A-B) and during stairs climbing (C-D). The abscissa is graded in percent of the total displacement whilst the ordinate is scaled in percent of muscular utilization. It is clearly illustrated that the muscular utilization patterns of each muscle remained similar during free and loaded activities. The percentage of muscular utilization of the quadriceps remains quite similar whilst the erector spinae was greatly affected. Indeed, by adding a load of 23 kg the mean muscular utilization of the erector spinae increased from 13.9 to 35.7 % during gait (Figure E) and from 20.8 to 50.9 % during stair climbing. For the quadriceps the utilization percentage increased from 8.3 to 13.7 % (Figure F) for walking and from 20.1 to 22.8 % for stair climbing. The results indicate that, unconsciously, the subjects flexed their trunk in order to decrease the stress exerted on leg muscles.

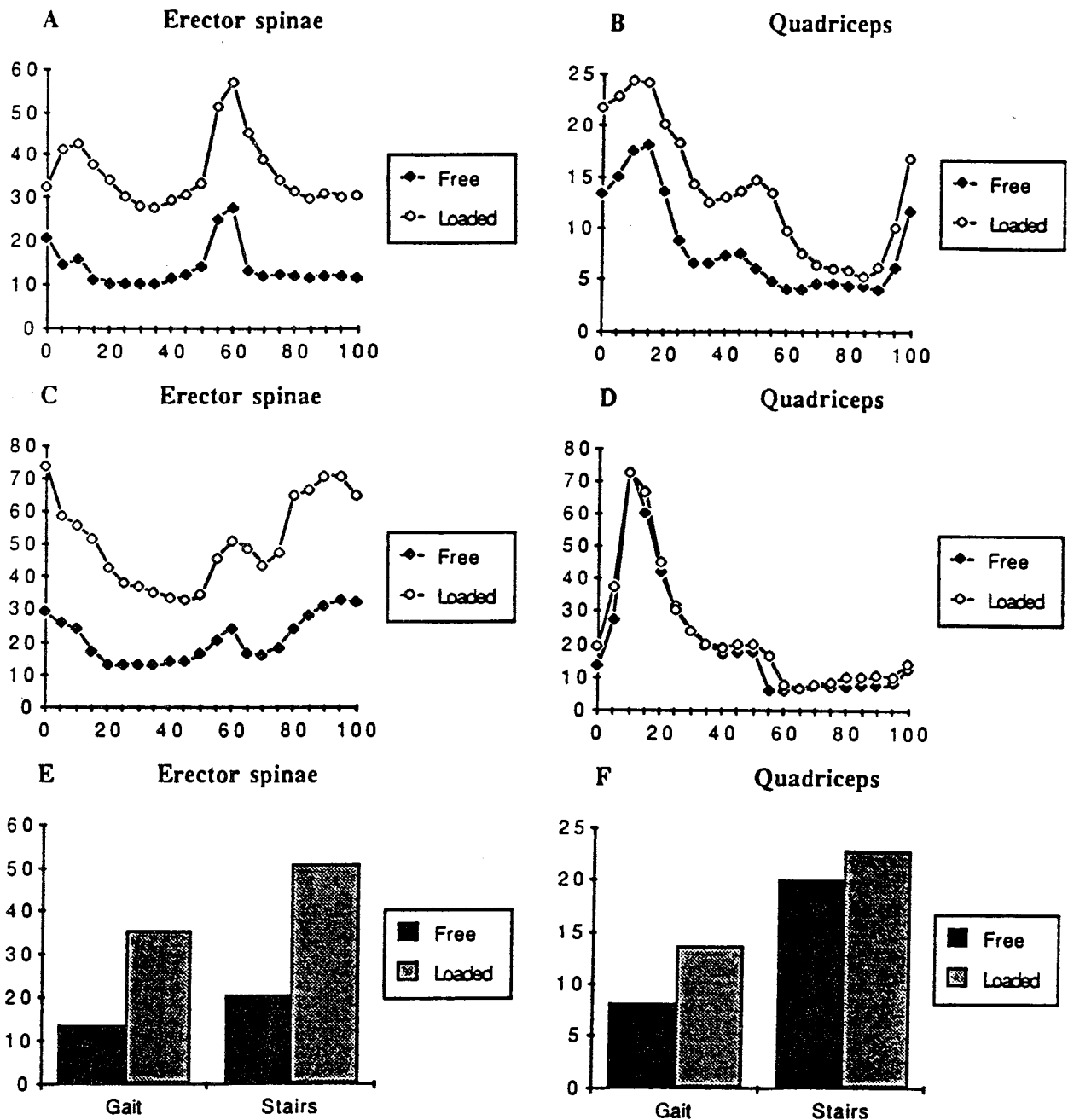


Figure 1: This Figure illustrated the muscular utilization of the erector spinae and the quadriceps during gait (A-B)) and stair climbing activities (C-D). Figures E and F illustrate the mean muscular utilization during these activities.

THE REMOVAL OF ELECTRICAL STIMULATION ARTIFACT FROM SURFACE EMG

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Introduction

The relationships between the surface electromyogram (EMG) and various mechanical parameters of human muscle contraction have received much investigation. A direct comparison of the EMG of a single muscle with its mechanical output is generally not possible due to the inability of subjects to voluntarily activate an individual muscle or the methodological problems of measuring the mechanical output of a single muscle. Electrical stimulation can be used to activate a single muscle and provided the other muscles crossing the joint remain relaxed, the mechanical output of the active muscle can be measured externally. Unfortunately, this stimulation causes artifacts on the EMG signal and distorts the activation information. The purpose of this study, therefore, was to develop a method that allows the removal of electrical stimulation artifact from the surface EMG signal so that distortion-free comparisons can be made between the activation of a single muscle and its mechanical output.

Methods

The abductor pollicis muscle of a single subject was stimulated via the median nerve at the level of the wrist. The stimulator (Grass model S88) generated a square wave pulse of 0.8 ms in duration at a frequency of 27 Hz which caused a nearly fused contraction (see Figure 1b). The subject's hand was secured to prevent wrist flexion and the thumb was connected to a force transducer (Gould model UC3) to measure the isometric moment produced by the abductor pollicis. Surface EMG was recorded using Beckman Ag-AgCl electrodes and amplification of bandwidth 10-1000 Hz and CMRR >80 db (DISA Type 15 C 01). The EMG and force signals were A/D converted (12 Bit) at a rate of 10,000 samples per second and stored on an IBM PC/AT microcomputer. The voltage of the stimulator was altered to produce varying force levels up to 30% of the maximum voluntary contraction (MVC) and a 210 second fatigue trial was also performed with one second samples collected each 30 seconds at the highest voltage.

Results and Discussion

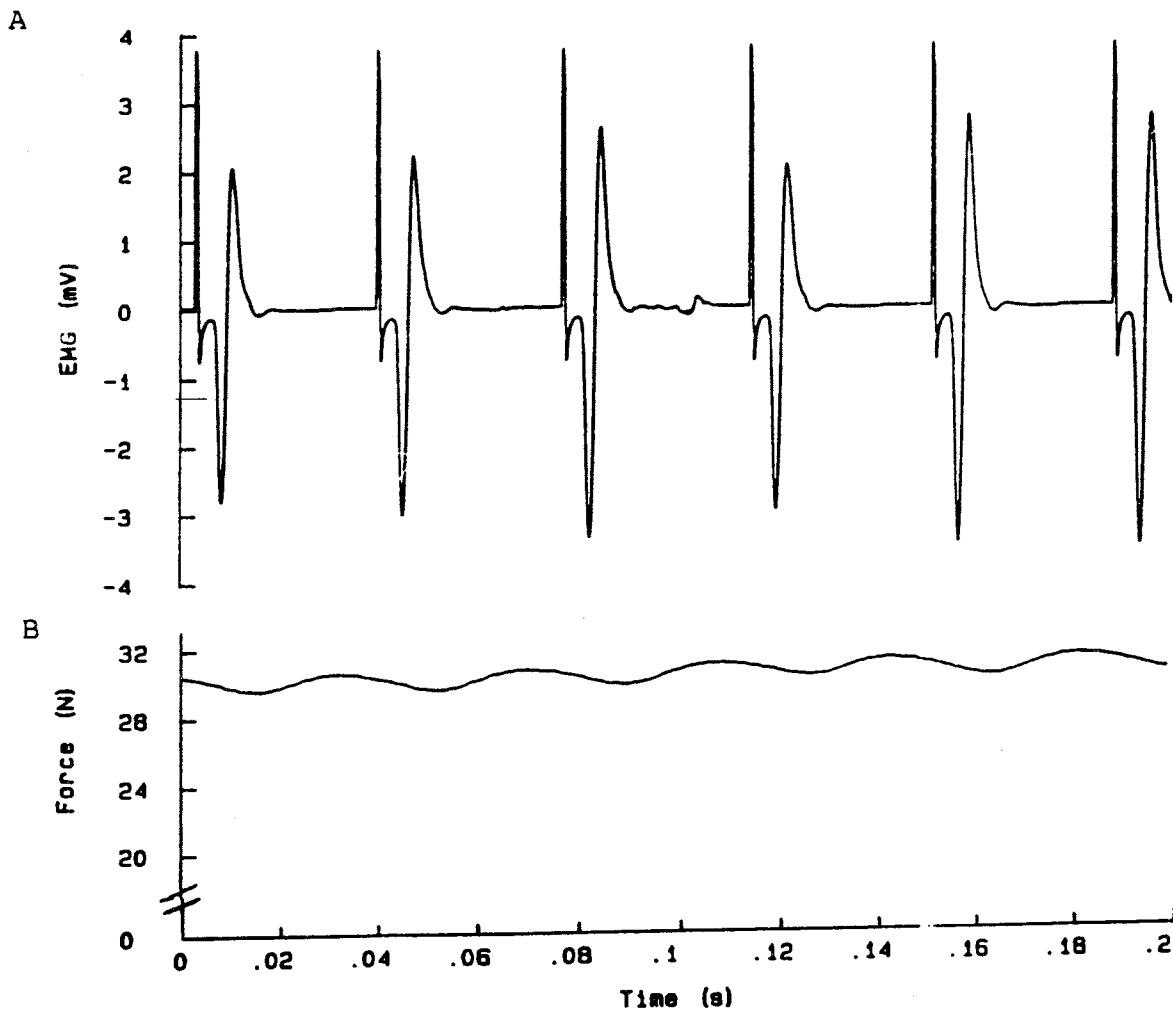
Figure 1 shows the recorded EMG (a) and the force (b) of the highest voltage (30% MVC) contraction. Each of the electrical stimulation pulses resulted in remarkably similar shaped artifacts. A computer program was written that allowed the user to specify the window that contained this first stimulus artifact. A template of the artifact was then digitally created by the program and moved through the signal sample-by-sample. At each sample, the template was cross-correlated with the signal and if the coefficient was greater than 0.95, the artifact was removed by subtracting the template. Simple linear regression was used in the template subtraction to allow for slight baseline and gain

changes that may have occurred during the collection.

An estimate of the amount of artifact not removed by each template subtraction was measured by dividing the RMS of the remaining signal in the window by the RMS of the artifact template and multiplying by 100%. It was found that $1.31\% \pm 0.93\%$ of the artifact remained following the removal process. A comparison of the average rectified EMG following artifact removal with the isometric force showed a strong positive relationship that was well represented by a straight line of best fit. This linear relationship is in good agreement with voluntary studies of small muscles of the hand. The shift to lower frequencies with fatigue often noted in voluntary contractions was also seen in this study. The median power frequency shifted from an initial value of 80.1 Hz to 50.3 Hz at the end of the fatigue trial.

Conclusions

The template subtraction method was extremely effective in removing electrical stimulation artifact from surface EMG. On average, the method was able to remove over 98% of the artifact. The resulting EMG signal was found to be nearly linear with force and showed the same spectral shifts with fatigue that is evident in voluntary contractions.



EMG ACTIVITY DURING LONG JUMP TAKE-OFF FROM DIFFERENT HEIGHTS

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Introduction

Long jump performed in normal conditions has been interestingly studied on the basis of biomechanical parameters (Hatze, 1981; Ridka-Drdacka, 1986). However, during training, athletes often use drills in which the take-off is carried out from platforms of different heights, in order not only to provide a longer period of flight but also to improve the take-off. These considerations which are of practical interest in track and field, have not yet been discussed objectively on the basis of neurophysiological experimental data. The present study examines this problem by comparing the electromyographic activity of synergistic muscles with mechanical parameters during the long jump take-off.

Material and Methods

Twelve male long jumpers (22 ± 4.6 years old) participated in this study. Their best performances ranged from 7.72 m to 5.85 m. The experiments were done in an indoor hall and all athletes were well accustomed to the testing procedure. Long jumps were performed from 5 different heights of take-off with respect to the track level : - 6 cm, 0 cm, + 6 cm, + 12 cm and + 18 cm (3 trials in each condition). The mean running speed over 5 m was measured 5 m before the take-off board by way of two infra-red photo-electric cells. The ground contact time (CT) of the last two steps and the duration between both were recorded by way of two contact platforms. The length of the last stride was simultaneously measured. Additional experiments were conducted to record the electromyographic (EMG) activities of the lateral gastrocnemius (LG), the vastus medialis (VM) and the biceps femoris (BF) of the take-off leg. The EMG were obtained through steel needle electrodes inserted subcutaneously over the motor point of the muscles. The signals were amplified and rectified before being measured with a digital oscilloscope (Nicolet, 4094C).

Results

Our results indicate that the mean run-up speed is $9.00 \pm 0.03 \text{ m.s}^{-1}$. This parameter is well correlated ($r = 0.95$) with the length of the jump. For take-off at ground level, the CT of the penultimate and the last steps were respectively $0.12 \pm 0.01 \text{ s}$ and $0.16 \pm 0.03 \text{ s}$. The length and the duration of the last stride were $2.12 \pm 0.08 \text{ m}$ and $0.22 \pm 0.02 \text{ s}$. Rectified and averaged ($n = 4$) EMG of BF, VM and MG before and during take-off at ground level are illustrated for one subject in Fig. 1. It appears that before the ground contact, BF has a more marked and earlier EMG activity than VM and MG. VM shows the largest activity just after ground contact, whereas MG and BF are much more active at the end of the take-off. In our experimental conditions, the comparison between take-offs carried out from different heights didn't show any statistical difference neither for the time related

mechanical parameters nor for the EMG characteristics (integrated EMG and duration).

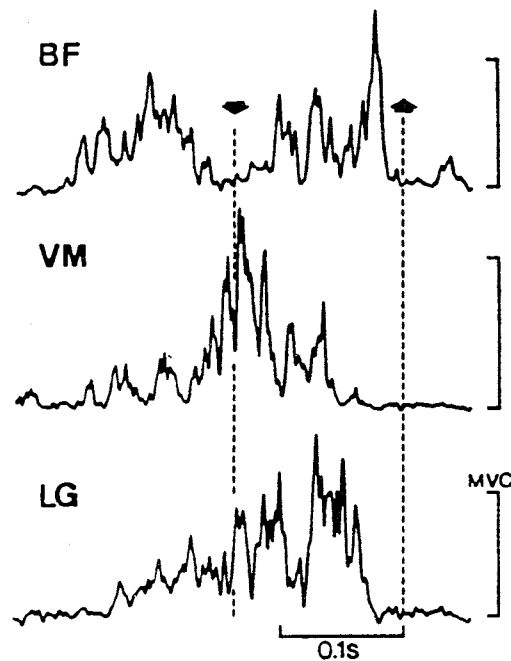


Fig 1. EMG activity of the biceps femoris (BF), vastus medialis (VM) and lateral gastrocnemius (LG) during long jump take-off in one subject (best performance : 6m91). The take-off was performed at the track level and the arrows indicate the time at which the foot touches (▼) and leaves (▲) the ground. The EMG amplitudes are compared to those recorded during a maximal isometric voluntary contraction (MVC).

Discussion

This work confirms earlier studies (Nigg et al, 1975; Kollath, 1982) which found that the run-up velocity is the important parameter when performing a long jump. Our results also show that VM is essentially active at the beginning of the ground impact, whereas MG and BF are principally active at the end of the pushing phase. The original finding that take-off from different heights (- 6cm to + 18cm) do not modify the timing and the level of activation of the muscles should be of practical interest for the track and field coaches. It is proposed that raising the take-off essentially increases the length of the jump, thus allowing more time to perform specific movements during the flight.

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EFFECTS OF DISUSE ATROPHY ON THE ELECTRICAL AND MECHANICAL PROPERTIES OF SINGLE MOTOR UNITS IN HUMANS

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Introduction

In a previous paper (Duchateau and Hainaut, 1987), it was reported that after immobilization following forearm fracture, the human adductor pollicis muscle shows significant changes of its contractile properties. The comparison of the electrical and mechanical alterations observed during voluntary contractions and in contractions evoked by the electrical stimulation of the motor nerve, indicates that immobilization not only changes the peripheral mechanisms of the contraction, but also affects its neural command. In this work, we examined the effects of 6-8 weeks disuse on single motor units (MU) in two small human hand muscles.

Methods

Single MU were recorded from the adductor pollicis and first dorsal interosseous muscles after 6-8 weeks of immobilization and compared with the contralateral muscle in five normal subjects. The mechanical properties of the MU were studied using the spike triggered averaging method (cf Stein et al, 1972). This method consists in recording individual fibers action potential through highly selective wire electrodes (40 μ m diameter) during a steady contraction against a strain gauge transducer (Hainaut et al, 1981). By repetitively triggering an averager (Nicolet, 4094C) with the intramuscular potential and recording the isometric force, it is possible to extract a single MU contribution from the whole mechanical force. The subject also performed 3 to 5 slow isometric ramp contractions following a target on the oscilloscope screen. For each ramp contraction, we determined the recruitment threshold as the level of force at which the MU action potential is recruited. We also recorded the minimal and maximal firing rates of each MU. Instantaneous firing frequency was calculated from the raw data stored on FM magnetic tape (Hewlett Packard, 3960), by a Apple II computer after being introduced into a window discriminator.

Results

In the two muscles, the immobilization resulted in similar modifications of the single MU mechanical twitch and firing rate. In disused muscles, the twitch time course was significantly slower in all recorded MU. The time to peak increased and the twitch tension decreased in a similar way in MU of different force threshold. The maximal firing rate was reduced in all recorded MU, whereas the minimal rate of firing was not significantly changed after immobilization. The maximal firing rate decreased more in MU of lower force threshold. The above reported modifications of the single MU twitch and rate of firing were not associated with a change of their orderly recruitment (Henneman et al, 1965).

Discussion

Previous finding that disuse atrophy does not only modify the peripheral mechanisms of muscle contraction, but also affects the neural command of the contraction (Duchateau and Hainaut, 1987) is rather difficult to discuss when looking at the muscle electrical and mechanical responses. The analysis of the single MU twitch parameters appears to be useful and provides interesting informations. The results indicate that 6-8 weeks of immobilization change similarly the mechanical properties of small hand muscles of different fiber type composition and of MU of different force threshold. The analysis of the decrease in maximal MU firing frequency suggests that immobilization changes differently the neural command of MU of different force threshold. Anyhow, it appears that the smaller force observed in disused muscles results from smaller MU tension development and decreased MU maximal rate of firing.

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AN ANALYSIS OF TRAJECTORY CONTROL STRATEGIES IN A GOAL ORIENTED THROWING TASK

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The ability to throw a rock at a moving target requires the parallel computation of the trajectories of both the stone and prey, which implies a highly organized brain and perceptual-motor ability (Calvin, 1983). Mastery of such a task demands movement planning with respect to the constraints of: 1) the geometry of the external world and 2) real-time (Hollerbach, 1985). Although the motor skill of throwing is no longer of vital importance to survival, the utilization of this particular highly skilled motor task still persists in many modern athletic endeavours. In 1986 an attempt was made by Anderson and Pitcairn to discover the parameters of a throw which would predict success by comparing the kinematics of the hits and misses. They also examined how these parameters were adjusted by the subject when throwing at a new target created by changing the distance from the target. Unfortunately the ability level of their subjects was quite low as reflected by their low success rates. This may have influenced their concluding statement, "luck was the most important parameter for success in dart throwing". The purpose of this study was to investigate the kinematic variables reflecting the variant and invariant aspects of the strategies of subjects of two ability levels performing a dart throwing task.

Methods

Four males, two novice and two elite dart players volunteered to be subjects in this study. The novice players were recreational dart players who had less than one year of experience and had not competed in an organized league, while the elite players had competed in an "A" division of the Montreal dart league. The subjects were required to perform the dart throw under three counter-balanced conditions of height, the double twenty, bullseye and the double three. The double twenty and double three were located 17 cm above and below the bullseye respectively while the bullseye was 173 cm from the floor. The throwing line was situated 237 cm from the target. High speed cine data was obtained using a Redlake Locam camera operating at 150 frames per second. The subjects performed eight trials for each of the three conditions.

Results

The dependant variables selected in this study were : 1) the resultant velocity of the styloid process of the wrist at release, 2) the resultant velocity of the dart at release, 3) the angle of projection of the dart at release and 4) the vertical coordinate of the dart at release. The following figures present the 4 dependant variables for each target and each subject.

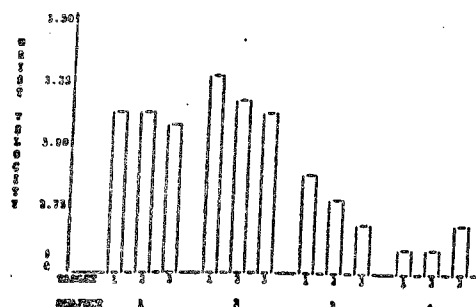


Figure 1: Resultant velocity (m/sec) of the wrist marker at release.

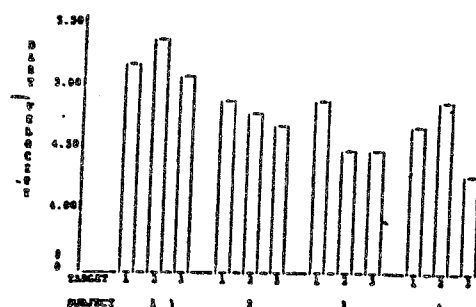


Figure 2: Resultant velocity (m/s) of the dart at release.

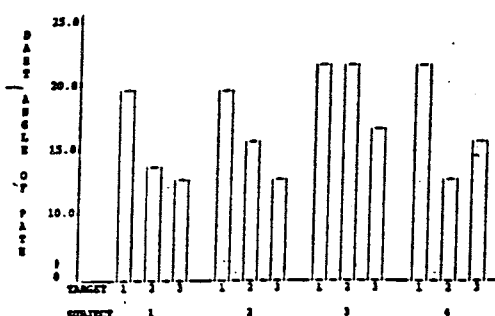


Figure 3 : Angle of path (degrees) of the dart at release.

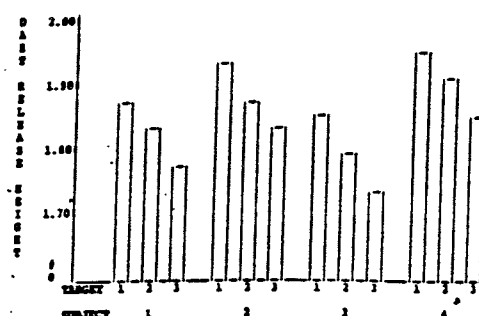


Figure 4 : Vertical coordinate at release (meters).

While there was no significant difference among the three levels of the independent variable a distinct trend was exhibited by subjects 2 & 3 demonstrating a decrease in wrist velocity at release as the target was lowered. However when horizontal and vertical velocities were analyzed separately, the elite players demonstrated a significant decrease in vertical velocity of the wrist as the target was lowered. Significant differences for dart velocity at release was demonstrated among all conditions for all subjects. While there was some flexibility concerning whether dart velocity was the sole strategy employed by the athletes it was available and employed by both elite and novice subjects. The angle of path of the dart was also controlled by both the elite and novice subjects. Once again the strategies employed by the subjects were flexible and not homogenous across conditions. The difference in release height of the dart was both significant and consistent. It seems that all subjects adjusted to the target height by decreasing the release height of the dart as the target was lowered. It should also be noted that the elite players consistently demonstrated lower standard deviations for all kinematic variables in comparison to the novice players, reflecting a more developed movement pattern. The results of this study suggest that both elite and novice dart players use a very simple strategy to adjust to target heights. This involves the manipulation of release height and the dart's angle of path at release.

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INTERTESTER AND DAY TO DAY VARIATION OF KNEE JOINT LAXITY
CHARACTERISTICS AS MEASURED BY THE GENUCON

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Recent developments in computerized equipment for the evaluation of knee joint laxities have made obtaining data a great deal easier. However the method of applying forces and positioning the limb is still done manually and requires both dexterity and precision. One of the most advanced instruments available commercially, the Genucon Knee analyzer has been demonstrated as being effective in identifying knee joint deficiencies (Oliver, & Coughlin 1986 ; Brien & Hoshizaki, 1988). While these studies provided data to detect changes in laxity with injuries there has been no attempt to establish the variation of these measures day to day or between testers. The purpose of this investigation was to establish the amount variation in knee laxity variables occurring between days as well as between testers.

Methods

Four male volunteers with no history of knee joint damage were enlisted as subjects for this study. Two trained testers with a minimum of six months of experience served as the testers. One was a male of 183 cms and 75 kgs while the other was a female of 155 cms and 45 kgs. the subjects visited the laboratory on two separate days during which they had their knees examined. Each subject was tested by both testers two times on two different days. The order of the testing was counter balanced. The testing session consisted of positioning the thigh in the restraints according to the procedures outlined in the Genucon operations manual (Faro,1984). This was followed by a soft tissue compensation procedure in order to account for the movement of the femur during force applications. The anterior-posterior drawer test was first done at a knee flexion angle of 30 degrees. A 98 newton force was applied to the tibia perpendicular to the long axis of the bone first anteriorly then posteriorly. The force was applied firmly but smoothly while being monitored by the Genucon. The translation occurring between the femur and tibia was recorded at a frequency of 30 Hz . This same procedure was repeated with the knee at an angle of 90 degrees. Internal- external rotation of the knee was measured at 90 degrees of knee flexion under a torque of 11 Nm. while varus-valgus laxity was measured under an applied torque of 14.9 Nm. with the knee at 20 degrees of flexion.

Results

The first comparison of interest to the authors consisted of the variation of measurements made on the same day but under two completely separate set ups. As can be noted in table 1 there were no significant differences between trials for either tester. In order to evaluate the effect of testing the same four subjects on two different days we compared the between day variation in table 2. Once again the values obtained on the two separate days proved to be very stable for both testers. Finally , a comparison of the two technicians revealed a significant difference for the anterior drawer at 90 degrees and for the posterior drawer at 30 degrees (Table 3).

TABLE 1
MEANS AND STANDARD DEVIATIONS FOR KNEE LAXITY VALUES
BETWEEN TRIALS FOR DAY 1

		Tester I		Tester II	
		Trial 1	Trial 2	Trial 1	Trial 2
Internal rotation		16.20	15.78	15.98	16.19
(degrees)	sd	2.06	1.05	3.13	4.23
External rotation		19.08	17.75	18.10	21.13
(degrees)	sd	7.32	3.64	5.89	4.79
Varus (mm)		5.85	5.55	6.23	5.55
	sd	1.34	2.07	1.93	1.52
Valgus (mm)		4.87	3.17	5.45	5.60
	sd	1.61	0.54	1.49	1.43
Anterior 30		6.90	7.87	7.55	8.03
(mm)	sd	1.36	1.33	0.78	2.42
Anterior 90		4.28	5.03	7.42	6.77
(mm)	sd	2.81	1.13	3.74	3.23
Posterior 30		4.57	4.13	2.57	1.70
(mm)	sd	1.16	0.61	0.38	1.20
Posterior 90		3.83	2.20	2.95	3.73
(mm)	sd	2.84	0.25	1.34	2.75

TABLE 2
MEANS AND STANDARD DEVIATIONS FOR LAXITY VALUES OF EACH TESTER
FOR DAY 1 AND DAY 2

		Tester I		Tester II	
		Day 1	Day 2	Day 1	Day 2
Internal Rotation		15.99	16.36	16.03	17.08
(degrees)	sd	1.54	2.60	3.29	4.05
External Rotation		18.41	19.89	19.57	19.78
(degrees)	sd	5.47	4.88	5.21	5.17
Varus (mm)		5.70	6.08	5.89	5.56
	sd	1.62	1.04	1.65	0.52
Valgus (mm)		4.03	4.98	5.53	4.54
	sd	1.44	1.31	1.63	1.65
Anterior 30		7.31	7.91	7.84	8.04
	sd	1.33	0.96	1.77	2.89
Anterior 90		4.60	3.46	7.14	7.47 *
	sd	2.87	1.43	3.35	2.20
Posterior 30		4.39	4.11	2.13	1.82 *
	sd	0.45	0.96	0.93	0.75
Posterior 90		3.02	2.33	3.42	2.79
	sd	2.02	1.52	2.10	2.02

* P < .05

TABLE 3
MEANS AND STANDARD DEVIATIONS FOR LAXITY VALUES
OF EACH TESTER

		Technician	
		Tester I	Tester II
Internal Rotation		16.52	16.18
(degrees)	sd	3.54	2.07
External Rotation		19.67	19.35
(degrees)	sd	4.97	5.04
Varus (mm)		5.73	5.04
	sd	1.19	1.33
Valgus (mm)		5.03	4.52
	sd	1.67	1.42
Anterior 30		7.63	7.96
	sd	1.15	2.34
Anterior 90		3.99	7.31 *
	sd	2.18	2.73
Posterior 30		4.24	2.13 *
	sd	1.29	0.93
Posterior 90		2.63	3.05
	sd	1.75	1.98

* P < .05

Conclusions

The knee laxity values obtained from the four subjects were demonstrated to be very stable between trials and day to day. In addition the two testers who were evaluated, did for the most part provide corroborating data. In the two cases where tester two demonstrated significantly different data both values were within the functional range of knee laxity. It seems reasonable to conclude from these data that the Genucom Knee analyzer provides reliable data between trials, from day to day and between testers.

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FORCE DISTRIBUTION IN VARIOUS RUGBY UNION SCRUMMAGING TECHNIQUES

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Probably no other aspect of play embodies the image of rugby union more than the set scrum. It has been extolled as "the very heart and soul of the game" (Irvine, 1978). The original purpose of the scrum was as a means of restarting play after an infringement, but it has since become both a powerful offensive and defensive manoeuvre. A consequence of this competitiveness has been the adoption of techniques that are contrary to the spirit of the game and in some cases, potentially hazardous to players.

Previously reported data on forces measured on the players in the front-row of the scrum (who transmit the forces exerted by the other scrum members) range from 2500 N (Cohen and Siff, 1979) for the front-row only, to "one and a half tons" (Scher, 1977) reported for the whole scrum. All these data have been limited to the forward direction without reference to the vertical or lateral shear forces.

The tightly bound scrum would appear to be an ideal mechanism for creating injury to the cervical spine. The front-rows charging together during formation could lead to hyperextension injury if the player's head is "popped" out of the scrum or a compressive fracture on clashing heads if the player is not correctly aligned. Compression and flexion/rotation injury can occur when the scrum collapses, "screws" or twists, or when players turn their head to look for the incoming ball. Although the risk of injury is not high, more than 50 serious cervical spinal injuries have been attributed to the rugby union scrum and these are almost exclusively restricted to front-row players. The purpose of this paper was to examine the forces experienced by the front-row players in a variety of scrummaging formations with a view to making recommendations to reduce or eliminate injuries.

Kinetic data on two rugby union forward packs were obtained using a Kistler force platform incorporated into an extended scrum machine. By systematically moving the front row along the scrum machine with each trial, data on all three front-row forwards could be obtained under each test condition. The three components of force at engagement were obtained, along with a representative force derived from the average force exerted during the time interval from one until three seconds after engagement. Data on body alignment and body motion were obtained from high speed film taken from an overhead and side perspective.

Both scrums were tested under their usual scrum formation as well as with modified techniques (hip and crotch binding, set and push technique, and sequential formation). To enable an estimate of each player's contribution, data were obtained under the following scrummaging combinations: front-row only; front-row plus second-row; full scrum less loose forwards; and full scrum.

A typical force-time history of each trial showed a large force at engagement followed by a considerable drop within the first second of the scrum being formed. It

is likely this drop in transmitted force was due to compression and realignment of the scrum during formation. The maximum combined forward (Z-direction) force transmitted by all three front-row forwards at engagement was 6545 N. Of this total, the hooker carried half the load (3380 N), the tight-head prop carried 2092 N and the loose-head prop carried 1073 N. This force was as much indicative of the speed at which the scrum contacted the scrum machine as it was of the ability of the players to exert forward force and indicates the need for corrected alignment and strength to maintain body position. The load experienced by the hooker exceeded the threshold for injury to the spine (2000 N in flexion and 562 N in extension) and all players should be made aware of the risk of injury during engagement. Once the scrum had settled, this combined force was reduced to 4605 N. Nearly half of this load (46 percent) was produced by the two second-row forwards.

Lateral shear forces have been identified as a causative mechanism in premature degeneration of the cervical spine of older rugby players (Scher, 1983). Results of the present study showed lateral shear forces (X-component) to be related to the experience level of the front-row players. More experienced players were better able to retain the integrity of the scrum. Vertical shear forces (Y-component) at engagement, which would tend to increase the risk of scrum collapse, were greater using the crotch binding technique. Eliminating this technique did not reduce the total forward force produced in the scrum. During sustained pushing, there was a tendency for the scrum to twist about its long axis, forcing some players upwards and possibly out of the scrum, as well as a tendency for the scrum to turn or "screw" in a clockwise direction.

Some recent technique and law changes appeared to achieve the aim of increasing scrummaging safety; eliminating crotch binding reduced the risk of scrum collapse, and the use of the set and push technique eliminated the large force at engagement. Other changes were counter-productive and potentially hazardous. The sequential binding technique did not reduce force and the requirement to have eight players in the scrum at all times could mean that it is potentially more dangerous to have an inexperienced and physically ill-equipped player in the scrum than an unbalanced number of players. Recommendations included the need for correct conditioning and coaching of players for the demands of front-row play, as well as specific technique and law changes to ensure player safety.

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Maximal Voluntary Contractions Of The Leg During Unilateral And Bilateral Work

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Inter- and intramuscular coordination effects are known to play a major role in the force production during maximal extension of the limb. This is particularly true for the intermuscular coordination in the case of a two- or more joint extension. A common observation related to maximal voluntary contractions of the leg concerns the restricted electrical activity and maximal strength in one leg during single leg extension as compared to the extension of both legs simultaneously. The objective of the presented study was to measure the characteristics of the rectified and averaged EMG-traces of selected extensor muscles participating in a multi-joint leg extension task. For this purpose we measured the EMG-time-records of nine female downhill skiers (regional level) during maximal voluntary contractions of one leg when working alone, and also simultaneously with the contralateral leg. EMG recordings were taken from m. soleus (SOL), m. gastrocnemius med. (GM), m. vastus med. (VM), and m. gluteus max. (GL) in the right leg in order to observe the muscle activation pattern. In seven of nine subjects the right leg was the stronger one concerning the isometric maximal strength. Beckman Ag/AgCl surface electrodes (type 650414) were placed in pairs (30-mm interelectrode distance) over the belly of the respective muscles in a way recommended by Zipp (5). The fixation of the electrode over GL was positioned so that no mechanical stress was applied to the electrode body while the subject was lying supine in the measurement apparatus. The quantification of the rectified and averaged (three cycles triggered by the force signal) EMG-data included the calculation of the integrated EMG over a fixed time interval as well as the calculation of the maximal amplitude and the highest rate of increase of the filtered traces (low-pass 10 Hz). As control force-time-signals were recorded for each leg and for both legs together via A/D-conversion (250 Hz) by a micro-computer and quantified online by means of the maximal amplitude and the highest rate of increase (Müller, Freiburg). Both parameters have been subsequently used to calculate the Bilateral Deficits (1).

Except for VM all muscles measured showed significantly more electrical activity (EMG-amplitude, IEMG) during the unilateral work. For the rate of increase in the filtered EMG-data as well as for the IEMG during the analysis period -50 ms to 100 ms, significantly less values were found in all muscles in the bilateral condition. The correlation between the rate of increase and the IEMG in this first analysis period was for all muscles constantly high (SOL unilateral : $r=.93$, bilateral : $r=.98$, GM unilateral : $r=.87$, bilateral $r=.91$, VM unilateral : $r=.90$, bilateral : $r=.97$, GL unilateral : $r=.72$, bilateral $r=.89$). Significant differences in the time lag of activation onset related to VM were merely found in GM.

In good accordance to previously published results (1,2,3,4) it is hereby indicated that the neural innervation scheme during bilateral maximal voluntary contraction is distinctly different from unilateral contraction. The observation of different intermuscular innervation characteristics as the significant time lag in GM and the unchanged activation level in VM whereas all other muscles measured exhibit reduced activity gives notion of different intermuscular coordination in both conditions. Furthermore by means of constantly reduced rate of increase in the filter EMG-traces of all muscles the view of Vandervoort et al. (4) can be supported who pointed out restricted activity of type II muscle fibres. This might however quite as well originate from a different mode of recruitment and/or reduced firing rates.

Tab. 1 means and standard deviations, t-values (onesided, * p < 0.05, ** p < 0.01), z-values (Wilcoxon-test) and point-biserial correlation coefficients for the IDMG-values, the maximal amplitudes and the maximal rate of increase of the filtered EMG-traces as well as the time lag between the onset of activity in the measured leg extensors (m. soleus, m. gastrocnemius med., m. vastus med. and m. gluteus max.) during unilateral and bilateral maximal isometric contraction in female downhill skiers (n=9)

	right leg unilateral	right leg bilateral	t-value	z-value	r _{pbs}
IDMG (-50 ms to 500 ms after the onset of the force development) [mV/ms]					
m. soleus	39.3 (16.73)	27.1 (9.12)	3.39**	2.55**	0.77**
m. gastrocnemius	18.2 (7.05)	13.5 (3.94)	3.40**	2.43**	0.77**
m. vastus med	124.6 (26.74)	124.6 (27.75)	0.00	0.18	0.00
m. gluteus max	39.7 (10.85)	32.7 (7.28)	3.09**	2.31**	0.74*
IDMG (-50 ms to 100 ms after the onset of the force development) [mV/ms]					
m. soleus	9.84 (5.04)	6.16 (3.55)	3.27**	2.66**	0.76**
m. gastrocnemius	4.81 (2.31)	3.26 (1.42)	3.55**	2.43**	0.78**
m. vastus med	32.59 (10.39)	29.47 (8.87)	3.32**	2.19**	0.76**
m. gluteus max	8.38 (2.39)	6.37 (2.07)	4.39**	2.67**	0.84**
maximal amplitudes in the filtered EMG-traces [mV]					
m. soleus	0.140 (0.063)	0.100 (0.041)	3.27**	2.52**	0.76**
m. gastrocnemius	0.064 (0.021)	0.044 (0.017)	5.44**	2.52**	0.89**
m. vastus med	0.335 (0.085)	0.342 (0.085)	0.42	0.47	0.15
m. gluteus max	0.114 (0.034)	0.090 (0.015)	2.75*	2.19*	0.70*
highest rate of increase in the filtered EMG-traces [mV/ms]					
m. soleus	1.809 (0.926)	1.352 (0.801)	2.12*	2.31**	0.60
m. gastrocnemius	0.807 (0.353)	0.576 (0.298)	2.79*	2.07*	0.70*
m. vastus med	4.366 (1.866)	3.869 (1.296)	1.78*	1.68*	0.53
m. gluteus max	1.273 (0.408)	0.943 (0.346)	2.92**	2.38**	0.72*
time lag of activation onset related to vastus med. [ms]					
m. soleus	43.9 (12.4)	51.9 (15.5)	1.20	1.13	0.39
m. gastrocnemius	47.2 (12.2)	57.7 (15.8)	1.77*	1.66*	0.53
m. gluteus max	33.3 (9.8)	32.3 (13.4)	0.23	0.24	0.08

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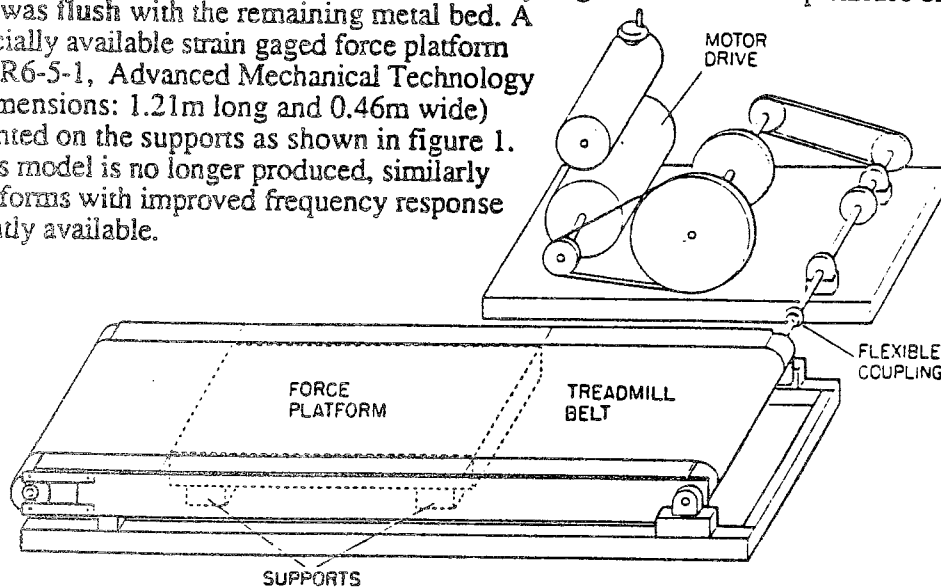
A TREADMILL MOUNTED FORCE PLATFORM

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Measuring ground reaction forces during locomotion with ground mounted force platforms is extremely time consuming. This is primarily because it is difficult to obtain repeatable constant speed trials (particularly with non-human animals). We have overcome this difficulty by mounting a force platform directly under the belt of a motorized treadmill. With this arrangement, we can now collect the vertical ground reaction force for an unlimited number of successive ground contacts over the full range of steady speeds and under various perturbations of normal gait. Simultaneous ground reaction force, film and oxygen consumption measurements can also be made.

There are several potential problems specific to incorporating a force platform into a treadmill: 1. mechanical cross talk from belt/plate horizontal friction forces onto the vertical force sensing elements, 2. erroneous force readings due to the belt either a) exerting upward vertical tension force on the foot, or b) applying extra downwards vertical force on the platform as the belt is pulled by the motor, and 3. motor and belt induced noise. Our design has overcome each of these problems while maintaining adequate force platform performance characteristics. This paper presents our design and evaluation.

DESCRIPTION- The motorized treadmill was fabricated from standard conveyor belt equipment. A rectangular section of the metal treadmill bed just large enough to accommodate the force platform was cut out. Rigid steel supports were then welded across the two side rails of the treadmill. These cross channels were carefully aligned so that the top surface of the platform was flush with the remaining metal bed. A commercially available strain gaged force platform (model OR6-5-1, Advanced Mechanical Technology Inc.). (dimensions: 1.21m long and 0.46m wide) was mounted on the supports as shown in figure 1. While this model is no longer produced, similarly sized platforms with improved frequency response are currently available.



EVALUATION- We verified that frictional forces were not causing any important error in our vertical force signal in several ways. First, a weighted, one-wheeled cart was placed on the belt above the force platform. The cart was bracketed with a pivot so that it would not translate or tip, but only roll and apply a vertical force of its weight. This created localized friction under the wheel between the belt and the top surface of the force platform. The force signal with the tread stopped and moving differed by less than 0.5% regardless of load or speed.

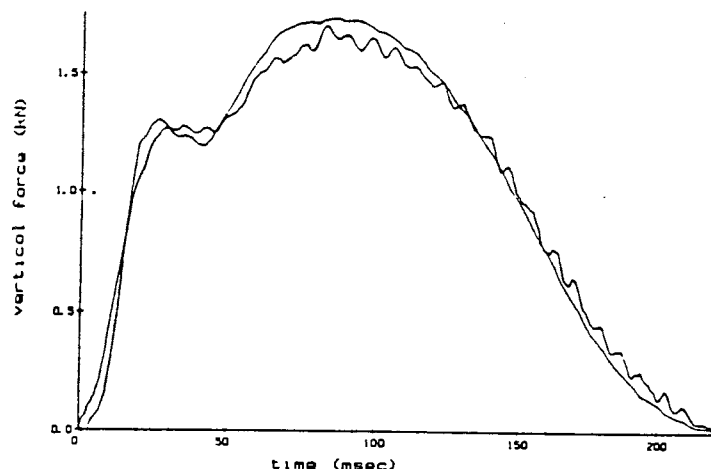
We performed two dynamic procedures to detect any erroneous belt forces caused by non-flush mounting. With the treadmill belt moving slowly, the force output was recorded as a person

lowered himself onto the treadmill belt in front of the platform, stood still, and rode the treadmill over and off the platform. The force output record from this procedure was a rising ramp, a level plateau and then a falling ramp. The average plateau force signal was within 0.5% of the person's weight and the fluctuations were less than $\pm 3\%$.

Motor and belt induced noise was minimized by mounting the motor on a separate platform and connecting the drive shaft to the driving roller with a rubber coupling. Noise incurred by both the moving belt and the motor was small in amplitude ($\pm 20\text{N}$), high in frequency and random. The unloaded natural frequency of vertical vibration of the platform was 160Hz as mounted in the treadmill; well above the frequency of the majority of the signal. It should be noted that this force platform was built as a prototype and that current AMTI production models have much higher natural frequencies. By mounting the platform directly under the belt, we have been able to maintain the unloaded natural frequency of the platform as high as possible. Other investigators have mounted an entire treadmill on top of a force platform (Martin and Gagnon 1987 ISB) or force platforms (Ewald Hennig pers. comm.) and thus added substantially to the "unloaded" mass of the platform.

Nonlinearity of vertical output (% deviation from a least-squares linear regression of output vs. load) was less than 5% over the range of 20 to 2000N. Variation in the force output with position of application was $< 2\%$.

Finally, we compared the force-time curves from a human subject running at $3.0\text{ m}\cdot\text{s}^{-1}$ over a ground mounted force platform (Kistler model 9261A) and on the treadmill mounted platform. The two records (collected at 400Hz and unfiltered) are shown in figure 2. The smoother trace is from the ground mounted platform. The contact time, peak force and overall shape are quite similar. Based on the positive results of all these tests, we conclude that this treadmill mounted force platform provides accurate measurement of vertical ground reaction force. Because the peak ground reaction force occurs at midcontact when vertical force is maximized and anterior-posterior shear force is near zero, this treadmill mounted force platform provides an accurate measure of the peak resultant force despite the lack of horizontal force measurement.



The most dramatic advantage of a treadmill mounted force platform is that so many contacts can be recorded in a very short period of time. It takes roughly 5 minutes to obtain a record from a single foot contact during a steady speed run using a floor mounted force platform runway system and a human subject. Using a treadmill mounted force platform, it is possible to collect 1000 foot strikes in five minutes. The time savings is even greater for non-human animal experiments. This work in part supported by NIH grant 5 R01 AM 18140. The loan of the prototype force platform by AMTI Inc. is greatly appreciated.

A GENERAL SIMULATION SYSTEM OF TWO-DIMENSIONAL MULTI-LINK MODEL OF HUMAN BODY

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1. Introduction

A kinetic analysis of a multi-link model of human body is one of the most important problems in order to investigate human movements. Therefore, there have been proposed some simulation systems of human movements. For instance, a long jump simulation was carried out by Ramey. Marshall formulated general Newtonian equations which govern the simulation of human movements on the basis of a model of the human bodies represented by open chain links.

This paper deals with a new two-dimensional simulation system applicable to various types of multi-link models without modification of the simulation programs. Newtonian equations are firstly established for individual links constructing the human bodies. Secondly, a method is proposed to simulate the human movements based on the Newtonian equations and the model of human bodies represented by the branching open chain links. Finally, the system is applied to the simulations of human movements in vertical jump and running motions.

2. Method

Newtonian equations of motion are formulated for each link of the human bodies. Muscle moments called 'joint torques' and forces acting to an individual link are illustrated in Fig.1, and Newtonian equations of motion of the link are given by the following formulas.

$$F_{i-1,i} - F_{i,i+1} + E + M_i \cdot g \cdot j = M_i \cdot a_i \quad (1)$$

$$T_{i-1,i} - T_{i,i+1} + V_{i,i-1} \times F_{i-1,i} - V_{i,i+1} \times F_{i,i+1} + V_{e_i} \times E = I_i \cdot \alpha_i \quad (2)$$

Since a pair of links are connected by one joint in the multi-link model, the following equation should be satisfied in order to maintain the connection between the links.

$$a_{i+1} = a_i - r_i \cdot \omega_i^2 \{ \cos \theta_i, \sin \theta_i \}^t + r_i \cdot \alpha_i \{ -\sin \theta_i, \cos \theta_i \}^t - r_{i+1} \cdot \omega_{i+1}^2 \{ \cos \theta_{i+1}, \sin \theta_{i+1} \}^t + r_{i+1} \cdot \alpha_{i+1} \{ -\sin \theta_{i+1}, \cos \theta_{i+1} \}^t \quad (3)$$

In the simulation system developed here, the equations above mentioned are automatically created in a matrix form from the model of human body stored in a Model Definition File (MDF). An example of the MDF is shown in Fig.2. The MDF includes the link constructive information, external force/torque information, and parameters of links. The link constructive information represents the number of nodes, links, and joints. If the link model has a constrained node, the ID number of the link including this node is specified in the first column of the external force/torque information as shown in the figure. The reaction forces acting on this node are automatically calculated by the simulation system. The input data files contain the joint torque histories, initial link coordinates and velocities, and external force/torque data.

One of the merits of the system is that the link construction can be changed easily by only modifying the contents of the MDF.

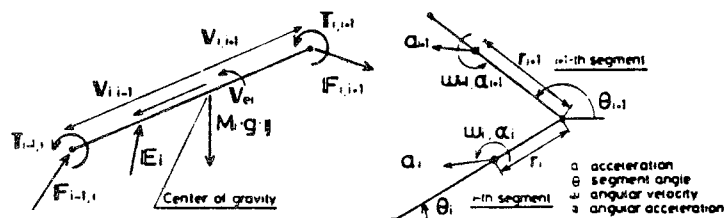


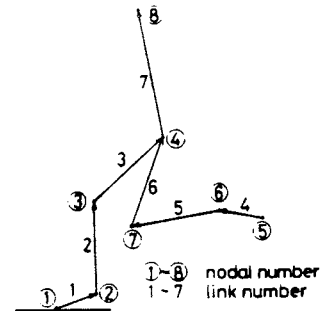
Fig.1 Force/torque acting on i-th link and relationship between neighboring links

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8      :the number of nodes
7      :the number of links
6      :the number of joints
1, 2, 3, 5, 6, 7, 4 :proximal point of segment vector
2, 3, 4, 6, 7, 4, 8 :distal point of segment vector
1, 2, 3, 4, 5, 6     :link connection information #1
-2, -3, -7, -5, -6, -7 :link connection information #2
-1, 0, 0, 0, 0, 0, 0 :external force/torque information
0.87, 2.79, 6.00, 0.87, 2.79, 6.00, 40.68 :link weight
0.50, 0.43, 0.43, 0.50, 0.43, 0.43, 0.63
      :ratio of position of CG from proximal side
0.0072, 0.0326, 0.0834, 0.0072, 0.0326, 0.0834, 2.6854 :inertia about CG

```

Fig.2 Examples of a Model Definition File (MDF)
and a link model



This system is implemented in FORTRAN-77 on an EWS (Apollo 590 Turbo), and Runge-Kutta method is employed to solve the differential equations given in Eqs. (1), (2), and (3).

3. Examples

The developed system has been applied to the simulation of vertical jump and running motions. Initial torque histories were firstly evaluated from the experiments of vertical jump and running motions.

In the case of the vertical jump, the torques histories which are higher than the initial ones in 5% were input to the system, and the simulation results shown in Fig.3 were obtained. The take-off velocity of center of gravity calculated by the simulation is higher than the initial one in 16% as shown in the figure.

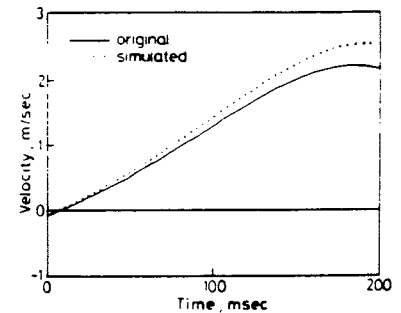


Fig.3 Results. (a) stick diagram, (b) take-off velocity

The simulation results of the running motion is shown in Fig.4. When the torque histories of the support leg were increased in 5% against the initial ones, the ground reaction forces were increased as shown in the figure.

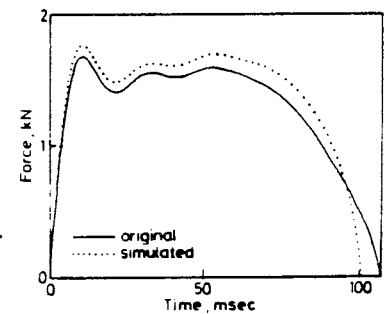
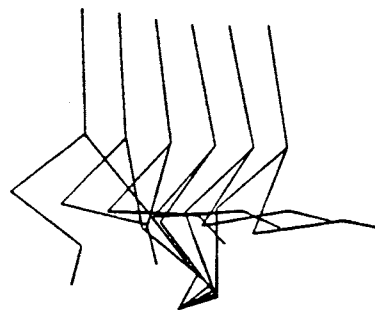


Fig.4 Results. (a) stick diagram, (b) reaction force

4. Conclusion

A new two-dimensional simulation system of the human bodies represented by the multi-link model was proposed and the effectiveness was verified through the simulations of the vertical jump and running motions. It is confirmed that the system is applicable to various types of multi-link models, and that it can be easily found how to improve the human performances by using the simulation system. An advanced simulation system adaptable to a closed-loop multi-link model is now under development.

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METHOD TO DETERMINE 3D COORDINATES WITH PANNING CAMERAS

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For three-dimensional movement reconstruction from film, a wide variety of methods are available. Application to a number of sports activities needs a reconstruction of coordinates in large volumes. If the displacement of the athlete disallows the use of two fixed cameras, then one has to use either more fixed cameras or panning cameras. This abstract describes a method to determine 3D coordinates with panning cameras as used during an investigation of speedskating. The method combines portability with relatively simple filming procedures; essentially the method is a combination of those presented by Dapena et al (1982 and 1987).

Method. The method demands known locations with respect to a generic reference frame R_0 , of two cameras and of background markers (markers for the lane indication can be used). Knowing the internal parameters of a camera, the coordinates of the image of the markers in the projection plan of the camera can be used to calculate the tilt-angle α (vertically) and the pan-angle φ (horizontally) of the optical axis i.e. the z-axis of a reference frame R_1 (or R_2) attached to the camera. If rotation about the optical axis is prohibited, the transformation matrix $M(\alpha, \varphi, \vec{c})$ is known ($\vec{c}$ is the location of the camera). The coordinates of the image of each point in the projection plane of a camera can be used to calculate the equation of the line that joins the camera with each object point. That equation is expressed in the camera-fixed reference frame. Knowing the transformation matrix $M(d\alpha, d\varphi, d\vec{c})$, dependent on the differences between the pan- and tilt angle and locations of the two cameras, the equation of the line can be expressed in the camera-fixed reference frame R_2 of the second camera. The coordinates of the image of the same point in the projection plane of this second camera enables the calculation of the equation of the line, that joins the involved object point with that camera. Intersection of both lines yields the coordinates in R_2 and with the help of matrix $M^{-1}(\alpha, \varphi, \vec{c})$, the coordinates in R_0 . Actually the midpoint of the perpendicular distance between the two lines will be used.

Concerning the internal camera parameters, the distance d from the camera to the theoretical projection plane, as appears in the collinearity equations of photogrammetry, can be obtained from a reference cross (Dapena et al, 1982). Disregarding a correction factor for vertical projected direction, the film shot of the known distance between two background markers at pan-angle $\varphi = 0$ can be used.

Validation. The origin of generic reference frame R_0 is located at the starting line of a 400-meter ice oval; the y-axis coincides with that line, the x-axis is determined by the markers as used as separation between the two lanes and the z-axis is vertical. The inter-marker distance is on the average 250 cm. The coordinates (in cm) of camera 1 is (970,1750,916) and pans along the x-axis in tilted position; camera 2 with (-3071,210,730) views along the x-axis and will tilt more than it pans (Locam II, Beaulieu zoom). A pole (175 cm) is placed at each marker position (largest x-coordinate is 1684 cm), horizontally parallel to the y-axis as well as vertically. The camera orientations were changed in such a way, that the pole was situated approximately in the middle of the film frame. Brief bursts of film were shot for each position of the pole, viewing at least two markers an arbitrary frame was analyzed. (NAC motion analyser). The coordinates of the image of the two markers in a frame were used to calculate the (averaged) value of α and φ . The distance d was obtained from the filmshots of the horizontal pole at the starting line (camera 2) or marker distance (camera 1) with $\varphi=0$. The (x,y,z) coordinates of the markers and of the pole in the generic reference frame R_0 were obtained from the calculated coordinates of the midpoint of the perpendicular distance of the two lines, as described above.

Camera 1 panned from -0.4 to 0.6 rad with $\alpha \approx 1.09$ rad. Camera 2 tilted from 1.36 to 1.45 rad with $\varphi \approx 0.01$ rad. The mean value and SD of all the involved perpendicular distances between the pairs of lines is 18 ± 14 cm. The deviation of the calculated x, y or z coordinates of the markers and the end of the pole from the real values is smaller than 10 cm. As expected these deviations are the lowest at small pan-angle of camera 1 (perpendicular then to camera 2) and increase systematically at higher pan-angles.

The deviation of the calculated length of the pole (horizontally as well as vertically) from the real value of 175 cm is 1.0 ± 1.2 cm i.e. within the limits of digitizing accuracy.

All the possible values of the direction cosines of the pole (horizontally and vertically) at any location show systematic deviation from real values (1 or 0). The extreme values were found at high values of the pan-angle φ of camera 1, if the pole is in vertical position: the angle deviation then is 0.08 rad. This deviation reduces drastically for all other positions.

Knowing these inaccuracies, the method can be used for motion analysis.

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ACTIVE AND PASSIVE DRAG IN SWIMMING

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Passive drag, the resistive force to which a swimmer is exposed when towed through the water in a fixed position, has been studied by many investigators. Values for the passive drag for a towing velocity of 1 m/s as given in the literature ranges from 25-35 N (Hollander et.al., 1987). The active drag, that is the resistive force experienced while actually swimming has often been indirectly estimated, yielding values ranging from 46-65 N at 1 m/s. Direct measurement of the active drag results in much smaller drag values (appr. 25 N at 1 m/s), values that are even smaller than the values for passive drag.

In the present study active and passive drag was measured in 12 male High school swimmers. The active drag was measured on the MAD-system (Hollander et.al, 1986). The passive drag was measured during towing in a prone position in a Swimming treadmill (United States Swimming, built by Unidyne, Minneapolis). The passive drag was measured using the force transducer of the MAD-system. The active drag at 1.5 m/s ranged from 50-75 N, whereas the passive drag at the same speed ranged from 20-40 N. The rather small values for the passive drag are due to the prone position, with the head down between the arms in the water, resulting in a very streamlined position of the body. Lifting the head so that nose and mouth are above the water surface results in a lower position of the arms and the hips and a threefold increase of the passive drag to 100 N at a speed of 1.5 m/s.

From this results it is concluded that passive drag values are very dependent on body position. In contrast the measured active drag gives a more realistic value for the resistance during swimming since the body of the swimmer during the measurements is in the same position as in real swimming.

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WALKING SYMMETRY AT DIFFERENT STEP FREQUENCIES: AN ANALYSIS BASED
ON THE GROUND REACTION FORCE PROCESSING.

by

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INTRODUCTION

Ground reaction force patterns have been widely adopted to point out inter-individual gait characteristics of normal and abnormal subjects [1,2,3]. Although such studies report a general discussion about the results obtained, specific details on intra-individual features of walking are often ignored. Particularly when collecting data from normal ambulation, that is always the base for comparison with pathological state, care should be paid to check the experimental consistency of some axiomatic aspects, including the symmetry of movement as well. The aim of this paper is to present the preliminary results of a quantitative investigation on the 3-D ground reaction force components, which was performed in order to obtain objective information concerning the symmetry of normal gait associated to different step frequencies.

METHOD AND EXPERIMENTAL PROCEDURE

The method consists of a suitable digital processing applied to those signals, generated by a force plate, proportional to the 3-D components of the ground reaction force acting beneath the foot during the stance phase of a step.

Each normal subject had to walk along a straight path-way about 15 m long at four different step frequencies imposed by the sound of a metronome. The force plate was located at about two thirds down the path way, where the subjects usually reached steady state progression speed, and fixed flush with the floor. Acquisition of the ground reaction force signals was performed at a sampling rate of 1 KHz which is more than adequate given the frequency content of the signals [4]. Signal processing [5,6] was essentially formed by three main frames: verification of the regular recurrence of measurement conditions, normalisation, estimation of walking symmetry.

The first frame was to check that the acquired data are not influenced by inertial predominance in the advancing direction. Steady state progression was accepted when the integral of the horizontal component along the sagittal plane gave a value within a suitable tolerance range around zero.

Normalisation was made on the 3-D force amplitudes, the application point displacements and the stance-phase duration.

The estimation of walking symmetry consists of a statistical estimation [6] of the differences occurring between the mean ground reaction patterns associated to the left and the right leg of each subject. Each mean pattern, identified by a series of 256 normalised values each variable, was compared through a two-tailed t-test at 0.01 and 0.05 level of significance.

To synthetize the results of the comparison process an index of estimated differences (μ) points out for each variable the percentage of the stance-phase where the left and the right mean pattern are significantly different [6] or, in other terms, where the hypothesis of walking symmetry is to reject significantly.

RESULTS

The results included in this paper were obtained by testing a preliminar group of 12 normal subjects. Each subject provided ten sets of data for the left and the right leg by walking at the four step frequencies (96, 108, 120, 132 steps per minute) imposed by the metronome.

Table 1 summarises the indexes of estimated (0.01 level of significance) differences associated to each variable of the ground reaction patterns: vertical component (Rz), horizontal component on the sagittal and the frontal plane (Rx, Ry), displacement of the application point on the sagittal and the frontal plane (Px, Py). As proved by the values shown in Table 1, normal gait is not featured by perfect symmetry; significant differences between the two legs concern all the variables and point out a final asymmetry that involves about 11% of the whole stance-phase. Such differences are particularly due to the displacement of the application point of force (about 17%) while the horizontal component of force along the frontal plane gives the best symmetry. The step frequencies providing the best symmetry (96 and 108 steps per minute) correspond to those which are more natural for normal ambulation.

Table 1

St.fr.	$\mu(Rz)$	$\mu(Rx)$	$\mu(Ry)$	$\mu(Px)$	$\mu(Py)$	mean
96	4.13 (2.68)	9.73 (5.92)	.78 (1.10)	11.41 (6.86)	14.06(19.85)	7.95 (5.19)
108	2.26 (2.60)	3.04 (6.27)	2.42 (2.30)	22.58(25.40)	16.56(26.45)	10.37(11.38)
120	3.44 (2.30)	10.39 (7.04)	1.87 (1.36)	22.57(15.80)	14.21(20.35)	10.50 (6.10)
132	23.67(16.34)	14.92(12.50)	3.28 (2.77)	14.30(19.88)	19.45(25.04)	15.12(11.71)
mean	8.37(11.97)	10.68 (8.12)	2.09 (2.06)	17.71(17.54)	16.07(21.32)	10.99 (8.76)

Note: numbers in parentheses refer to standard deviations.

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ON THE BIOMECHANICAL NONSENSE OF PRIMARY ARTHROSCOPIC ACL-SUTURES

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Objectives of the study: In the past 5 years an increasing number of papers deal with the unfortunate late results of primary suture of the anterior cruciate ligament (ACL). Today the majority of orthopedic surgeons use some augmentation techniques (2) "except in certain cases..". It is these exceptions that matter: Young, otherwise healthy pts., active in sports. Regardless of the known problems such as impaired vascular or neural supply of the torn ACL or biomechanical aspects of the reconstruction, there has to be a basic misconception - for conservative and operative treatment both may produce a comparable clinical outcome (1,7).

Material and methods: The following in vitro-experiments were performed: I) Knot holding capacity of 3 suture materials (n=21): nylon (polyamid), steel (stainless steel), vicryl (polyglactic acid). II) Primary suture retention of 3 ACL suture techniques (n=27): Marshall/Insall (4,5), Müller (6), lateral trap; primary fixation of pedicled patellar tendon grafts (PT) - see fig.

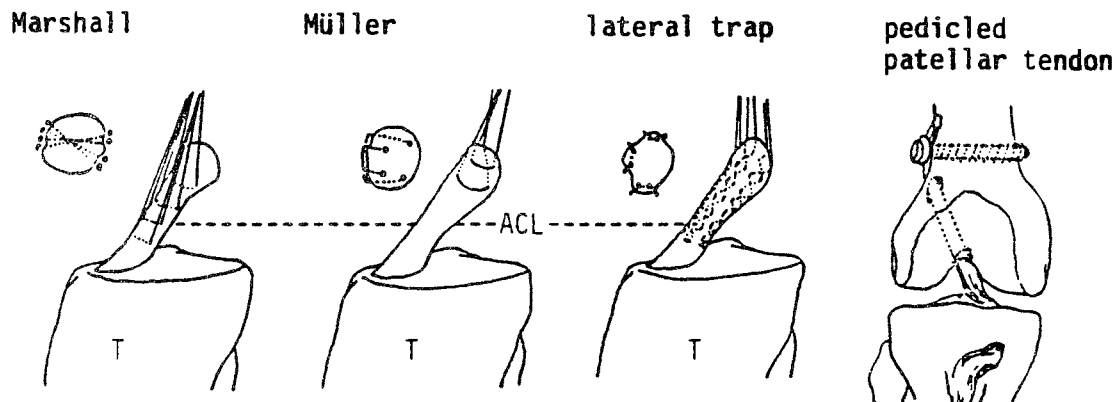


Fig.: Schematic drawing of the operative techniques (4,6). T=tibia, ACL with placed sutures; small circles: top view of ACL; PT: fixation with an AO-screw, patella and lig. patellae removed.

These sutures and the PTs as well as the biomechanical tests were done on post-mortem specimens of sheep knees within a 4-hour period following sacrifice. Parameters for evaluation: stiffness, knot holding capacity, primary suture retention, maximum load to failure, mode of failure.

Results: From a merely biomechanical point of view, the suture techniques given in the textbooks almost inevitably lead to an insufficient result, because the primary suture retention is extremely weak: 82.8 N, 33.7 N and

35-50 N respectively for the Marshall-, Müller- and anchor-technique (Schmid, 1988,9). Only the lateral trap technique shows values that make sense and correspond to the results obtained with the PT: 204.5 N vs. 161.2 N. The use of steel sutures with a high stiffness is unfavorable, the knot holding capacity with the lateral trap technique is lowered to 119.1 N. The stiffness of strands of vicryl and nylon is equivalent to the stiffness of reconstructive procedures where these sutures were used. On the contrary, steel sutures are more likely to cut through the tissue, the low force/lengthening ratio results in early failure. Some results are summarized below.

Tab.: Knot holding capacity and primary suture retention (μ /SD/SEM)

sutures		F max. (N)			N/mm		
nylon	4 strands	209.4	11.3	4.3	7.1	0.2	0.1
steel	2 strands	156.0	11.9	4.5	62.1	6.5	2.4
vicryl	3 strands	170.2	30.6	11.6	17.6	1.0	0.4
reconstructive procedures							
Marshall/Insall-nylon		82.8	19.4	11.2	6.8	1.3	0.8
Müller-steel		33.7	6.4	3.7	18.3	8.0	4.6
lateral trap-vicryl		204.5	18.4	8.2	17.4	3.9	1.7
lateral trap-steel		119.1	14.1	8.2	14.7	1.7	1.0
anchor-suture - polydioxanone		35-50 (range; 9)					

Discussion: In primary ACL reconstruction the use of an augmentation technique is mandatory (3). Arthroscopic placement of efficient sutures is impossible: Our biomechanical investigations clearly demonstrate, that the generally applied suture techniques (8) in most cases won't tolerate early functional after treatment. P.o.p. casting of the knee may be detrimental to the ACL and the cartilage, regarding morphology and function: A 4 weeks immobilization after ACL-reconstruction in sheep with a PT results in a 45 % reduction of the maximum breaking strength in comparison to the not immobilized controls - 448±70 N vs. 833± 274 N (survival: 6 months, n=14, done by AR). A lately recommended technique (9) using a PDS-anchor suture also exhibits an insufficient primary suture retention. The above mentioned "basic misconception" is biomechanical.

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STRAIN OF CARPAL LIGAMENTS DURING WRIST-JOINT MOTION

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INTRODUCTION

In order to understand the mechanics of the wrist-joint, information about the kinematics of the carpal bones and the behaviour of the ligaments during hand-motions is required. The kinematics of the carpal bones have been studied by de Lange *et al.* (1985). Prevailing concepts on ligament-behaviour are predominantly deduced from two-dimensional (2D) data on carpal kinematics. However, de Lange *et al.* (1985) proved that carpal bones display 3D motions while the hand is displaced in a 2D plane. Furthermore, these concepts lack the support of experimental verification and focus primarily on deviation-movements (Mayfield *et al.*, 1976; Bonjean *et al.*, 1981).

In this study 3D experimental data on ligament length changes are obtained in order to analyze their role in flexion and deviation of the hand.

METHODS

Radio-opaque markers are introduced into the carpal bones and 6 ligaments (see table 1) of fresh human cadaver wrist-joints. The bones are supplied with at least three tantalum markers each, varying in size for different carpals; the ligaments are represented by a string of 4 to 5 markers. On 3 wider ligaments, two of such strings are placed.

The prepared specimens (n=4) are moved through flexion and deviation cycles by displacement steps of 4 degrees. At each position the 3D positions of the markers are calculated, using Röntgen-stereophotogrammetric (RSP) principles.

Movements of the carpals and length changes of the ligaments in relation to the movement of the wrist are determined.

RESULTS

In table 1 the relative length changes for the ligaments in the four end-positions of movement are given. There is a clear difference between strain occurring during deviation- and during flexion-motions. In flexion-motions the length changes are considerably larger. Furthermore, it is noticed that the ligament lengthening is one-way directed. Ligaments tend to lengthen during either ulnar or radial deviation and during either palmar or dorsal flexion. A third remarkable finding is the absence of length change in any ligament during the movement from the neutral to the radially deviated position.

In figure 1 a representation is given of the ligamentous length-behaviour: the relative length change of a ligament is plotted against the motion of the hand. These results are partly in conflict with earlier, 2D concepts. For instance, in contrast to former studies the ligament RadioLunatum Palmare lengthens during ulnar deviation.

DISCUSSION AND CONCLUSIONS

The clear differences between our experimental data and the current concepts on ligament behaviour show that 3D information will contribute to a more accurate apprehension of the organisation of the wrist-joint.

The higher strains in the ligaments during carpal flexion, relative to deviation, is easily explained by the greater motion-range of the flexion-movement. These different strains imply higher stresses during flexion.

The fact that all the ligaments seem to strain in only one direction of movement supports the conclusion that each of them is able to stabilize the joint during just one part of the movement. The opposite part of the movement has to be stabilized by other (ligamentous) structures. In the motion from neutral to radial deviation, and vice versa, ligament length changes do not seem to occur,

so there can be no ligamentous constraint applied to the joint, and its integrity has to be taken care of by other structures. During the flexion-motion a comparable phenomenon is observed for the dorsal triquetro-trapezium ligament. This is hardly strained during palmar flexion and so it has to be ruled out as a possible intercarpal joint stabilator during this motion.

It can be concluded that to develop a proper concept on ligament-function, three-dimensional data are necessary, and that the ligaments tested can not be the sole stabilizing factors in the wrist-joint.

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Table 1: The mean (n=4; *n=2) length change of the tested ligaments in percentages in the four end-positions relative to the neutral position. The standard deviations are put between brackets.

LIGAMENTS	ulnar deviation	radial deviation	palmar flexion	dorsal flexion
RadioCapitatum Palmare	106.2 (2.2)	101.4 (2.2)	87.3 (7.5)	111.9 (4.5)
RadioLunatum Palmare prox.	101.7 (3.4)	100.0 (1.0)	87.2 (8.7)	103.3 (2.2)
RadioLunatum Palmare dist.	105.3 (2.7)	99.9 (0.6)	90.8 (2.9)	105.6 (2.9)
LunatoTriquetrum Palm. prox.	99.2 (1.6)	99.7 (1.7)	98.6 (2.8)	101.3 (0.7)
LunatoTriquetrum Palm. dist.	100.7 (0.5)	99.0 (0.9)	98.1 (2.4)	100.9 (0.2)
TriquetroCapitatum Palm. pr.*	105.4 (3.1)	97.3 (0.4)	101.8 (1.3)	103.9 (0.9)
TriquetroCapitatum Palm. di.*	96.4 (0.9)	99.4 (2.2)	98.9 (0.6)	101.8 (3.9)
TriquetroCapitatum Palmare*	88.8 (1.8)	96.8 (1.1)	94.0 (1.4)	101.8 (1.1)
RadioTriquetrum Dorsale	95.8 (3.3)	98.1 (0.7)	104.8 (5.3)	95.7 (0.8)
TriquetroTrapezium Dorsale*	106.4 (6.1)	100.0 (0.7)	98.9 (4.2)	98.0(2.3)

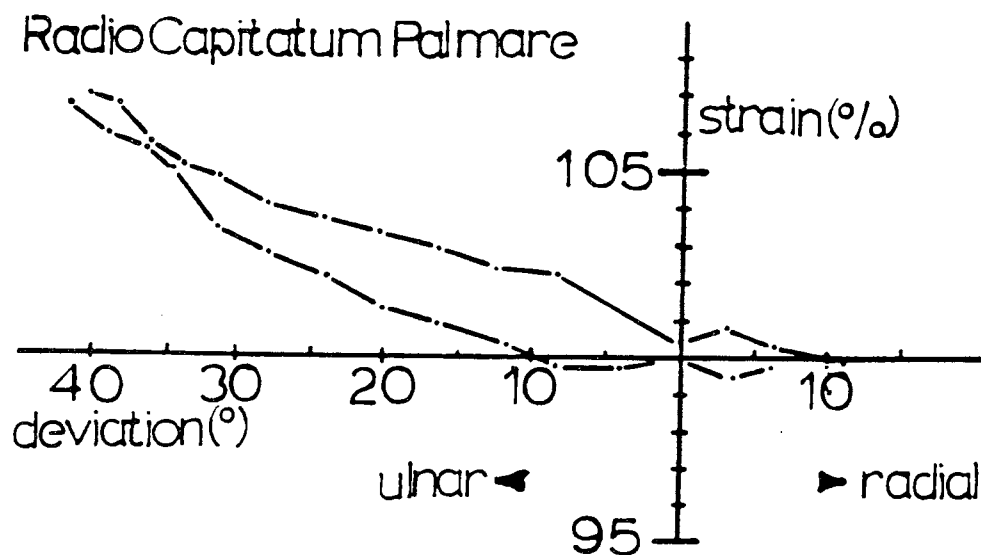


Figure 1: The relative length change of the ligament RadioCapitatum Palmare as a function of the deviation of the wrist-joint.

THE STRESS PROTECTION RATE OF BONE-PLATE SYSTEM

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Fracture healing has close relationship with its mechanical environment. The effects of stress protection generated by clinically used stainless steel plate might cause the ill-effect of bone atrophy.

MATERIAL AND METHOD

Finite element models (FEM) were made by investigators to find out the relationship between stiffness of the plate and its stress protection effects. However, the stiffness loss of the plate generated by the viscoelastic nature of the bone and incomplete anchor among the bone, plate and screw cannot be considered by the FEM. This loss will result in systematic error in the calculated results.

In this thesis, an experimental model consisting of a compression plate and a femur in vitro was established, and a three-dimensional finite element model was made accordingly. By modifying the traditional method, we

used the advantages of both FEM and experiment method to modify each other in order to obtain an ideal mechanical model.

The stress and deformation of the bone-plate system under various external loading conditions were calculated by using the modified model, and a comparison of stress protection effects between Ti-alloy plate and stainless steel plate was made by the modification of the elastic modulus of the plate on the FEM.

In addition, some elements with variable elastic modulus were established at the fracture site of the FEM. The elastic modulus of the intact bone, cartilage and granulation tissue was taken into account for calculation in order to get the effects of

change of bone tissue on the stress situation during fracture healing phase.

In this paper, through the biomechanical tested with four human wet cadaver femur. The load and strain were recorded automatically on the trip by X Y recorder. The distribution of stress, strain, displacement under pressure has been measured. It was carried out with the autograph machine whose measuring error was less than 1 percent.

RESULTS AND DISCUSSION

- 1, In this paper, appropriate agreement between results of experiment and calculation was obtained.
- 2, The stress protection rate (SPR) generated by the plate in this model at fracture site are between 17 - 52 % under various external loading conditions.
- 3, The SPR of Ti-alloy plate is 67 % as much as that of stainless steel plate.
- 4, The healing situations of

the tissue at fracture site affect the SPR, the harder the tissue the more stress stimulation the bone under the plate would have.

Results of this study indicated that the FEM established by this method is of practical significance.

KEYWORDS:

biomechanics fracture
healing stress protection
rate

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**PRESSURE DISTRIBUTION PATTERNS INSIDE OF A RUNNING SHOE
DURING UP- AND DOWNHILL RUNNING**

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The knowledge of the pressure distribution under the foot during locomotor activities is necessary for the understanding of the foot function. Twelve healthy subjects were instructed to run on a "Woodway" treadmill uphill, level as well as downhill at given speeds of 3 m/s and 4 m/s (8% inclination). One type of running shoe was chosen for all subjects. Eight piezoceramic sensors were taped onto the plantar surface of the foot at defined anatomical landmarks. The dimensions of the transducers were 5 mm x 5 mm. The peak pressures in the lateral and medial heel region were found to be considerably increased during downhill running. A shift of the relative load from rearfoot to forefoot could be observed when comparing uphill and downhill running. Increased pressure values for the hallux during uphill running demonstrate its important function to propel the body. High interindividual variability were found for all anatomical regions. This finding may indicate a pronounced individual running style of the subjects.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

The influence of individual movement pattern on coordinative efficiency in road cycling

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Introduction

After previous laboratory studies in bicycle ergometer exercise, it seemed desirable to investigate the coordinative behaviour of cycling on the road. There is only little knowledge (SODEN et al. 1979) about the movement pattern under different field conditions (climbing uphill, sprint, gear changes, road surfaces). So we have to investigate these influences on the movement pattern i.e. force-time curves in cycling motion.

Instrumentation

We attach two three - dimensional pedal - dynamometers (KISTLER Type 9251a) (Fig.1) with low-drift charge-amplifier (KISTLER Type 5023) on a custom ten - speed racing bicycle. Simultaneously the crank angle (digital angle coder) and the pedal angles are registered. All this data were transmitted by an eight channel PCM telemetry (MBB 3800). From this quantities the coordinative efficiency is determined by the propulsive forces and the resultant forces over a crank cycle (ZOTH 1899). To control the influences of the road surface we attached an accelerometer (ENDEVKO) at the seat tube.

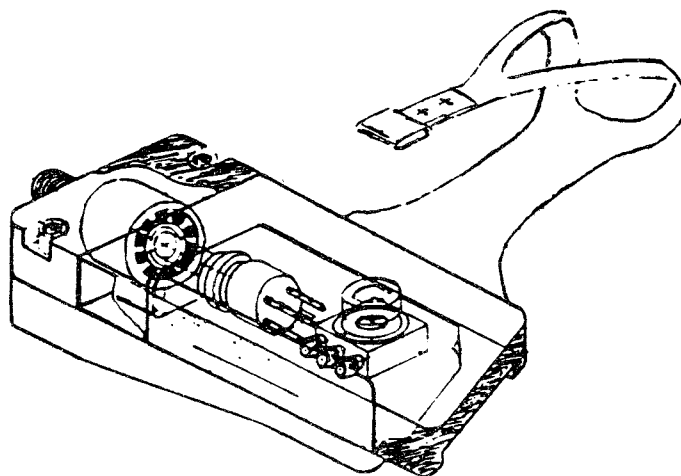


Fig. 1

Methods

We determined the dependency of the coordinative efficiency under different conditions of cadence and work load in ergometer study. Pedal frequencies of 40, 60, 80, 100 and 120 rpm and work loads of 100 and 200 Watt were chosen for this investigation. The subjects (n=7) were healthy sport students (age 24.7 ± 4.2 years SD; height 183 ± 5 cm SD; weight 76.1 ± 6.9 kg SD).

Results

While cadence increased (Fig. 2) from 40 to 120 rpm the coordinative efficiency decreased from $\eta = 74\%$ to $\eta = 34\%$ at a work load of 200 Watt and from $\eta = 67\%$ to $\eta = 19\%$ at 100 Watt work load.

If compared with field measurements ergometer exercise did not produce identical movement patterns due to environmental conditions like road surfaces, climbing uphill, sprint for example.

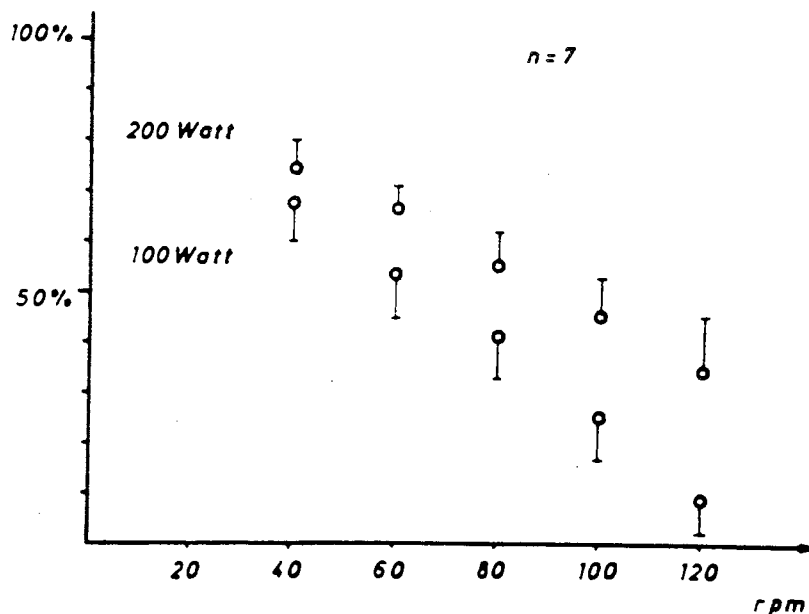


Fig. 2

In road cycling we found relatively high efficiency for competitive cyclists (national champion) in high cadences. At pedal frequency up to 120 rpm the efficiency of competitive cyclists never decreased below $\eta = 50\%$ whereas the efficiency of triathletes decreased down to $\eta = 20\%$.

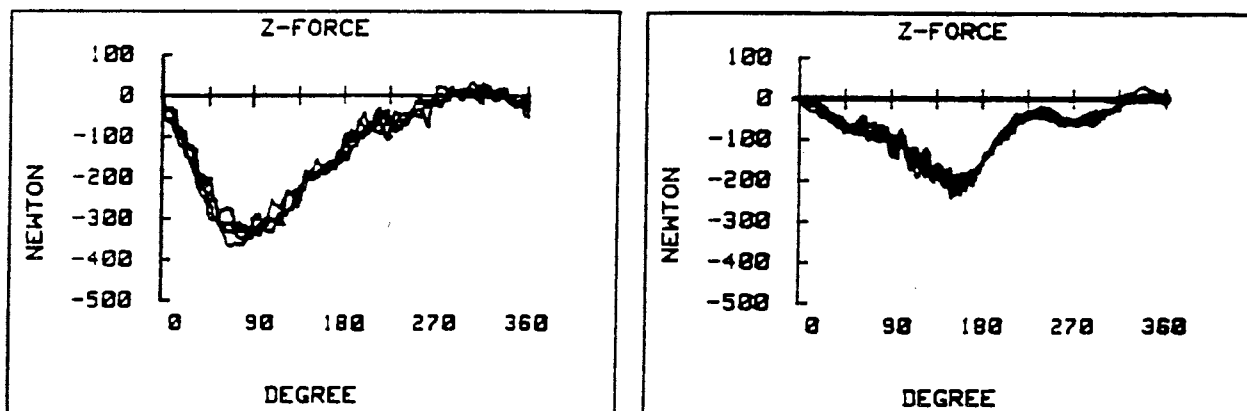


Fig. 3 a) competitive cyclist b) triathlete

For technique training we identify individual motion pattern (Fig. 3) like uneffective force application on the pedal (here z-direction). In an interactive process we are trying to improve the force-time pattern with the athletes. The movement pattern of top athletes serves as optimization criteria.

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Biomechanical study on sprint form and muscle strain

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Improper running form was pointed out as one of causes of hamstring muscles strain which frequently occurs during sprinting. A 100m sprint was usually divided into three phases : acceleration, maximal speed and endurance. Although clinical studies suggest that hamstring muscles strain during 100m sprint race occurs in any of three phases, there has been little investigation on the relationship between injury and running form during these phases.

The purpose of this study was to investigate the difference in the running forms at three phases, i.e. acceleration (30m), maximal speed(50m), and endurance(80m), during 100m sprint of the sprinters who had, and those who had not suffered from hamstring muscles-strain.

METHODS

Eleven male sprinters were divided into two groups : an injured group(N=6, 20.8 ± 1.2 yr, 1.78 ± 0.04 m, 642.7 ± 37.2 N) and an uninjured group(N=5, 19.4 ± 0.1 yr, 1.77 ± 0.05 m, 668.4 ± 34.1 N), according to their history of previous hamstring muscles strain. 100m sprint best record for each group was 10.77 ± 0.15 sec and 10.77 ± 0.23 sec, respectively.

They were filmed at the 30m, 50m and 80m marks during 100m maximal effort sprint with three 16mm high speed cameras operating at 120fps, 200fps and 200fps, respectively. At least one running cycle was filmed for each runner at each of the three marks. Coordinate data of the body markers smoothed with a digital filter were used to compute linear and angular kinematics. Muscle lengths of hamstrings were computed using the model described by Frigo and Pedotti(1978).

RESULTS AND DISCUSSION

Maximal running velocity was obtained at the 50m mark for both groups and that of the injured group tended to be greater than the uninjured group(9.88 ± 0.33 m/s vs 9.67 ± 0.27 m/s). Stride lengths of the injured group increased with distance progression while those of the uninjured group unchanged from the 50m to the 80m mark.

Muscle lengths of hamstrings(BFcl:biceps femoris caput longum, SM:semimembranosus, ST:semitendinosus) for both groups were beyond anatomical level during recovery phase at all three marks, indicating that hamstrings were under eccentric

contraction during late recovery phase. Maximal lengthening of hamstrings during recovery phase were obtained at the 50m mark for both groups, and that of BFcl was greater than other muscles for both groups(Fig. 1). The injured group tended to have greater hamstring muscles length during recovery phase than the uninjured at all three marks. These results suggested that hamstring muscles of the injured group were under more stress during recovery phase than the uninjured. One of the causes of these differences between two groups was the fact that the injured group tended to swing the lower leg more forward during recovery phase than the uninjured(Fig. 2). The lower leg movement shown by the injured group might reflect the position of the lower leg at the contralateral toeoff. The injured group tended to lift the upper leg higher at the contralateral toeoff, which also might have effects on the position and the movement of the lower leg at the contralateral toeoff and recovery phase.

From these results, the 50m mark while 100m maximal effort sprint might be the critical stage of the occurrence of hamstring muscles strain during recovery phase for the sprinter who breaks contact with the upper leg lifting higher and the lower leg swinging more forward, and also swings the lower leg largely forward during recovery phase.

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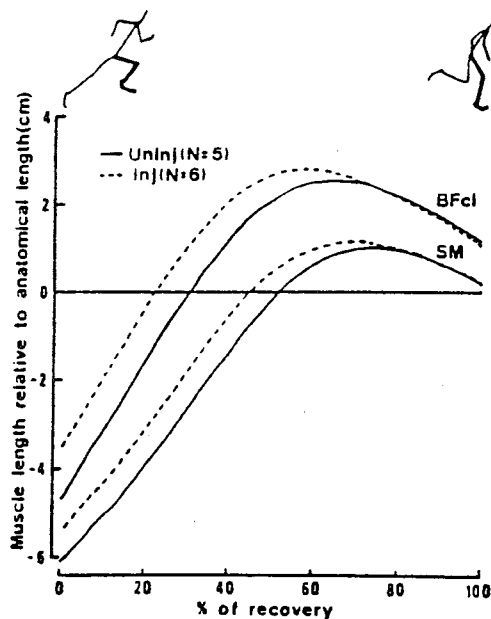


Figure 1 Hamstring muscles length during recovery phase at the 50m mark

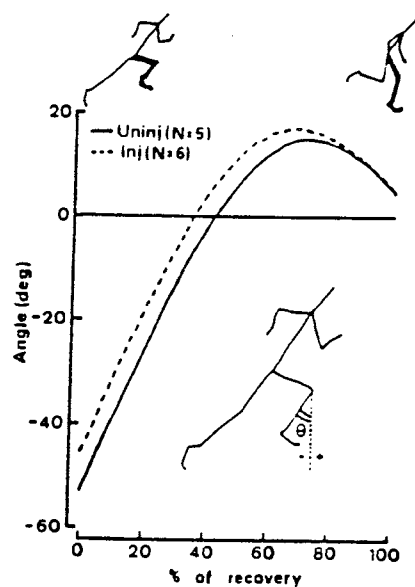


Figure 2 Lower leg angle(θ) during recovery phase at the 50m mark

BIOPHYSICAL CONTROL OF THE BONE TISSUE GROWTH

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The complex bone tissue growth control system (with the whole set of its dynamical data) suggested, worked out and tested by the authors permits finally to receive the necessary set of its physico-biological characteristics in the output. This system is built on the basis of the latest theoretical and experimental data concerning the bone tissue growth nature (the presupposed mechanisms and models) and also on the analysis of the main regularities and peculiarities of the growth, depending upon the patient's health and its individual characteristics (i.e. his age). Special attention was paid to the examination of the patient statistical data concerning the influence of patient's health defections of several types (i.e. professional ailments) on the process under discussion. The significance of the patient's health and that of the external psychological factors has been estimated by the authors within the framework of the results, known at the present time.

The comparative analysis and the demonstrated analogy of the bone tissue growth process and its principal mechanism obtained on the basis of the latest theoretical and experimental data have been made to well-known phenomena in various branches of natural science and technology such as crystal growth, cement hardening, water purification during its technical use in the several hydroconstructions (hydroequipment), the prolongation of the oil wells exploitation (hydrodrives of the heat engine) etc. The identity of these phenomena though different they may seem at the first sight as well as the identity of the main mechanisms of this process (sometimes the use of the model description schemas) and qualitatively identity of the main behaviour regularities of the above-mentioned schemes are of great importance. This brings about quite new possibilities of the bone tissue growth dynamic control (such as the

bone tissue layers growth rate), its space distribution, the packing density etc. It opens quite new possibilities for obtaining the predetermined optimal final values of physico-biological characteristics of the bone tissue (included the transition area between new and old tissue). There are such parameters as the tissue elasticity, durability, internal structure etc.

The main ways and methods of the bone tissue growth control are formed on the basis of borrowed reliable classical approaches within the above-mentioned branches of science, which have a strict theoretical background, confirmed by the multilateral computer modelling and by convincing experimental data, containing the heat, mechanical, electrical and magnet fields influence. In the common case (depending from the patient's individualities) we may use one of the above-mentioned fields influence as well as any one their superpositions. Special attention was paid by the authors to the using of the constant external magnet field and to the analysis of the influence its main physical parameters - intensity, influence duration, special orientation etc.

It is necessary to stress that the using of the magnet field for the realized bone tissue growth regulation has a number of the positive additional effects (the positive magnet field influences on the blood circulation system as a whole, including also the reduction of the deposits of any-types on the blood vessels, causing negative effect.

The constant external magnet field may be used for the influence on the special centres (or the groups of centers) stimulating bone tissue growth as well as other processes in the human organism, being of great importance.

It is remarkable that the using of the constant external magnet field in combination with other influence types permit to receive good results.

The application efficiency of the whole system worked out by the authors and the constant external magnet field as it part depends on the optimal choice of the whole set of influence parameters, depending from the viability characteristics of the concrete patient.

Change in functions of the lower limb muscles in 100 m sprint

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Some studies investigated the manner in which fatigue affected mechanics of sprinting by observing 400m sprint. However, little information is available on the changes in mechanics, especially kinetics of 100m sprint with the progression of distance from start. The purpose of this study was to identify the change in the functions of the leg muscles in starting, acceleration, maximum speed, and endurance stages of 100m sprint by using mechanical work as the primary variable.

Methods

Twelve male sprinters ran 30m and 100m from sprint start. They were filmed at the 1st and 5th steps from start of 30m run (film speed 64 and 100 Hz) and at 30, 50, 80 m marks of 100m run (125, 200, and 200 Hz) with high-speed cameras. Force platform technique was used for the 1st and 5th steps, but not for 100m run because of the difficulty of striking the force platform. Acceleration of the CG was used to estimate ground reaction force for 100m run. Digitized coordinate data were smoothed by a digital filter cutting off at 8 to 10 Hz. Planar link segment model was used to compute muscle moments, powers and works of the leg joints. Work done for movement phases was calculated by integrating the muscle power by the time of each phase.

Results and Discussion

Running speed reached the maximum at 50m and decreased thereafter (See Table). Stride length (SL) drastically increased from the 1-step to 30m and showed no large change from 30m to 80m while stride frequency (SF) slightly increased to 30m and was 96.5 % of 30m at 80m. This suggests that small decrease in running speed has been induced by the decrease in SF.

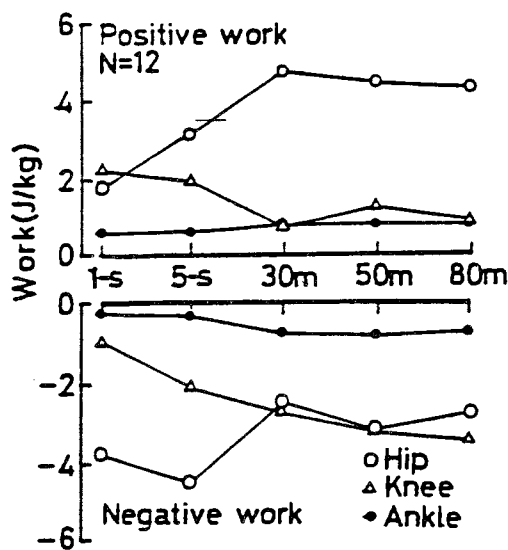
Performance descriptors in 100m sprint

Variable	1-step	5-step	30 m	50 m	80 m
Running speed(m/s)	4.58 (0.16)	7.07 (0.19)	9.79 (0.25)	9.83 (0.19)	9.63 (0.22)
Stride length(m)	1.09 (0.06)	1.63 (0.09)	2.12 (0.10)	2.18 (0.08)	2.18 (0.07)
Stride frequency(Hz)	4.20 (0.21)	4.36 (0.27)	4.62 (0.26)	4.51 (0.19)	4.46 (0.19)

Range of knee extension at the driving phase was significantly larger in the starting stage(49.1 and 33.0 deg) than the other stages(18.2, 18.3, and 19.7 deg). This suggests that knee extension during the driving phase is one of important factors to accelerate the body from sprint start. Recovery of the free thigh at touchdown and toef off was delayed at the starting stage because of short flight time. Free thigh was recovered early at 30m, but gradually delayed from 50m to 80m due to fatigue.

Figure shows positive and negative works(PW and NW) done by the leg muscles during one running cycle. There was no remarkable change in both PW and NW of the ankle with progression in distance. Knee PWs in starting stage were significantly larger than those of the other stages. Since these works were done during support phase, knee extensors would play a large part in starting sprint. Hip PW showed the reversed change of the knee PW. When running speed decreased from 50m to 80m, hip PW with knee PW decreased slightly. A close examination showed that the decrease in hip PW resulted from large decrease occurred in the early recovery phase. The Decrease in hip PW in this phase may result in the delayed recovery leg.

With distance progression knee NW increased, but decreased at the late recovery phase from 50m to 80m. Unexpectedly large hip NW in starting stage, done in both the driving and early recovery phases, was observed. This NW might play a part in recovering the leg in short time and in keeping the trunk lean forward. Most of the decrease in hip NW from 50m to 80m was attributed to the decrease in the driving phase rather than in the recovery phase. Combined with the decrease in hip PW of early recovery



phase, the decrease in hip NW in this phase may be one of the major limiting factors of delayed recovery leg and less forward thigh flexion at toef off, which lead to the decreases in SF and running speed eventually.

It may be concluded that function of the lower limb muscles varies from stage to stage of 100m sprint so that knee extensors play a major part in starting and acceleration stages and hip flexors in maximum speed and endurance stages.

Figure: Works done in the various stages of 100m sprint.

INFLUENCE OF THE DURATION OF INTERSTIMULUS INTERVAL ON MOTOR PREPARATION RECORDED ON THE CORTICAL AND SPINAL LEVEL

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In a warned reaction time (RT) experiment slow negative brain potential is present in the interstimulus interval (ISI) between the warning stimulus (WS) and the response stimulus (RS). This so-called contingent negative variation (CNV; Walter et al., 1964) consists of at least 2 components. The first wave is related to the WS, the second to the RS anticipation and movement preparation. Changes in excitability of spinal motoneurons during ISI can be estimated by means of Hoffman (H-) reflexes (Brunia, 1980; Bonnet et al., 1981). The purpose of this study was to investigate the influence of the duration of ISI on the time course of motor preparation recorded on the cortical (Experiment I) and spinal level (Experiment II).

Methods

In Exp.I, subjects (5 male, 23-26 years old) lay down on the bed in a prone position. They were asked to prepare the response when an auditory WS was given, and to perform a planter flexion of the right foot as soon as an auditory RS was given. A WS was followed by a RS with a constant interval in a series of 35 trials. All subjects performed 3 different series (ISI 1, 2, and 4s). An Ag/AgCl electrode was fixed to the scalp at C3' (2cm left to Cz). Linked mastoids served as reference. EMG recorded from the medial gastrocnemius muscle was used to measure the RT. EEG and EMG signals were amplified and stored on magnetic tape. Trials were discarded from analysis if artifacts were present during ISI, and in case subjects reacted too early or too late. EEG data were digitized with a sample frequency of 100Hz and averaged by a mini-computer (Nihon Kohden, ATAC-450) from one sec before WS to RS. In Exp.II, subjects (5 male, 25-27 years old) performed the same procedure as Exp.I except that H-reflexes were evoked in ISI and intertrial interval (ITI) instead of the recording of CNV. Surface electrodes were used both for stimulation and recording. Rectangular current pulses of 1 ms duration delivered by a stimulator (San-ei, 3F46) were applied transcutaneously to the tibial nerve of the right leg in the popliteal fossa. The anode was placed proximal of the knee. The stimulus strength was kept so that a small direct motor response was visible. Reflexes were recorded from the medial gastrocnemius muscle. Reflexes were evoked in a random order, at 11 different measuring points during the ISI. During each ITI a reflex was evoked. Using the mean (M) and standard deviation (s) of these reference reflexes amplitude values, the reflex amplitude during ISI were converted into z score; $z = (X - M) / s$ (Bonnet et al., 1981).

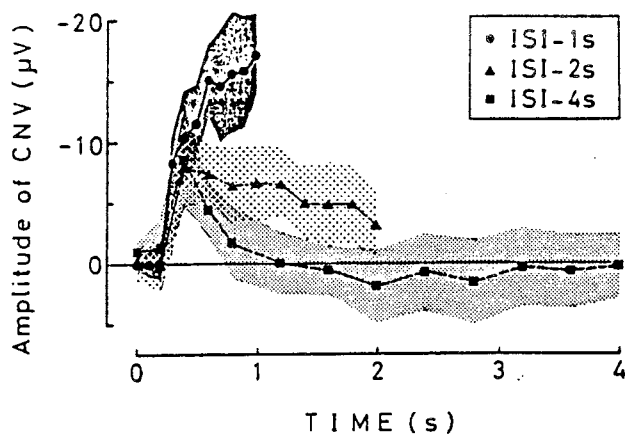


Figure 1 Time courses of the amplitude of CNV. Zero on time axis indicates the time at which WS was given. Screen areas indicate standard errors.

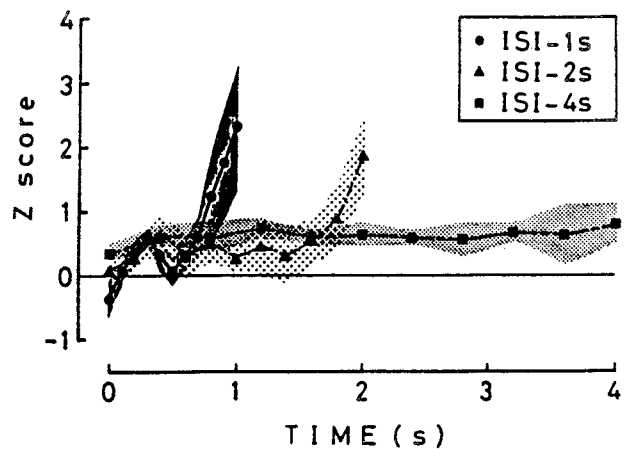


Figure 2 Time courses of the mean amplitudes of H reflexes, in terms of z score.

Results and Discussion

In Exp.I, the amplitude of CNV over the motor cortex changed with time in the different manners related to the durations of ISIs (Fig.1). The longer the ISI, the less the amplitude of CNV at the instant when the reaction stimulus was given. In Exp.II, the H-reflex amplitude of the involved leg changed just before the RS (Fig.2). The shorter the ISI, the larger the change. Both in Exp.I and II, the longer the ISI, the longer the RT.

These results indicate that the longer the ISI the less accuracy that subjects can estimate the time at which their preparation should be at a maximum. It is considered that as a result of inability to hold a state of preparation, RT increases with duration of ISI. In conclusion, the different duration of ISI may change the difficulty of time estimation and then the strategy and manner of motor preparation.

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SOME STRUCTURAL PARAMETERS OF M.GASTROCNEMIUS LINKED IN A MODEL

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It has been shown (Wagemans et al., 1988) that the macroscopic behaviour of muscle fibres during a contraction can be visualized by means of ultrasound. In this study an attempt is made to combine some structural parameters of a pinnate muscle into a mechanical model.

Therefore, twenty-one male subjects, aged between 19 and 28 years, have been tested. During a static plantar flexion in 5 different angles at the ankle, a simultaneous videoregistration is made of the exerted moment (M_o) and the ultrasound image of m.gastrocnemius (caput mediale). From the latter some structural parameters are determined: 1) the angle of insertion of the muscle fibres on the tendon plate (α), 2) the section of the muscle belly (MS) and 3) the fibre length (FL) (Wagemans et al., 1988).

Tabel 1 shows the correlations between these structural parameters and the moment in the different angles at the ankle. The highest correlation is found between the moment and the fibre angle, a correlation which decreases with increasing angle at the ankle.

	angle at the ankle (degrees)				
	60	70	90	110	120
$M_o(t) - \alpha(t)$	.87	.86	.82	.73	.71
$M_o(t) - MS(t)$	.74	.72	.77	.74	.72
$M_o(t) - FL(t)$	-.79	-.78	-.58	-.36	-.33
$M_o(t) - F_u(t)$	.94	.93	.90	.86	.85

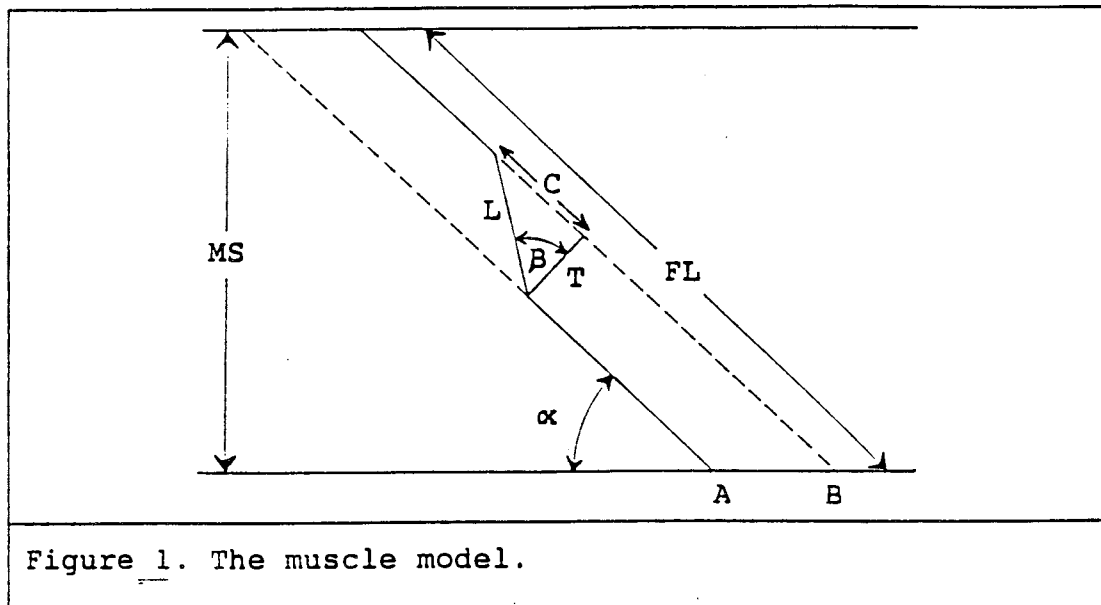
Tabel 1. Correlation between 1) the moment (M_o) and the structural parameters α , MS and FL , and 2) the moment and the predicted one (F_u), in 5 angles at the ankle. (N=42, 21 subjects test and retest)

In order to get a higher prediction of the moment at the ankle from the structural parameters and at the same time trying to give a mechanical representation of the behaviour of the fibres during a static contraction, a model is used. The muscle model constructed by Van Leemputte (1985) for parallel fibred muscles, is adapted for m.gastrocnemius, a unipinnate muscle.

In figure 1 a muscle with section MS is shown, consisting of N fibres, of length FL and thickness T , inserting at an angle α on the tendon plate. The distance of each fibre on the tendon plate (AB) is supposed to be constant and in connection with the adjacent one. In the model the relation between the fibre length (FL) and the thickness (T) is represented by a lever L , placed at an angle β : when the fibre shortens, β will decrease, resulting in an increasing T .

Within one fibre, VN of these succeeding elements are supposed. After a try out on 21 subjects, it seemed that only a few combinations of α , β and T are possible to admit all fibre movements in the different angles at the ankle. For a given $\alpha = 30$ degrees, $\beta = 65$ degrees and $T = 0.005$ cm, L and AB are determined for all further contractions, test and re-test, and only the angle of insertion α is used as input of the model. Finally, a serial visco-elastic component is added.

From the fibre angle α , the thickness T and subsequently β and C can be calculated. VN is verified for every subject and, by multiplying it with C , the fibre length and the muscle section are obtained. For every subject and every angle at the ankle, the highest model value of the fibre length is determined: FL_0 . When the difference of FL_0 and FL at every moment is multiplied with the cosine of α , the useful component of the force (F_u) is obtained.



The correlation between the exerted moment and the one obtained from the model is given in tabel 1. So, using this model, the shape of the moment during a static plantar flexion can be predicted from the changes of the angle of insertion of m.gastrocnemius with an accuracy up to 11.6, 13.6, 19., 26. and 28.6 % in the different angles at the ankle. Besides, the model enables the possibility to interpret the observed changes of the muscle structure.

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BRAKING HORIZONTAL MOMENTUM: A METHOD FOR DEVELOPING LEG POWER

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There is wealth of literature concerning the importance and the methods of developing the muscular power. One of the recent trends in the area is known as plyometrics (Chu 1983;1984, Costello 1984) which makes use of the vertical momentum of the free falling subject. It was the aim of the present study to investigate the effect of braking horizontal momentum developed during running on developing muscular power of the lower extremity, which may provide a practical method for developing this physical quality.

METHODS

Twelve students from the University of Bahrain volunteered to serve as subjects for the study. The mean height was 169.45 ± 8.20 cm., mean body weight was 65.44 ± 11.92 and ages was ranging between 19 - 21 years. Speed-Distance curves of running 50 m. were examined by means of ultra-sound measuring system (Hing and Nicol 1976) and especially designed computer program for fitting and differentiating the time - distance data. The 30m dash was selected as a training stimulus on the basis of providing a safe effective load in terms of the - horizontal momentum formula. The subjects were pre-tested in long and vertical jumps to provide basic data for indicating and calculating the leg muscular power (Mathews and Fox 1976) before exposing them to an 8 week program of three weekly sessions consisting of 36 trials divided into six sets (Veroshanski 1967, Gametta 1981). Each trial required the subject to accelerate for 30m as fast as he could and try to brake his movement at the shortest distance from the end line. A post-test was administered at the end of the training program. The same testing protocol was applied to select a control matching group ($n=12$).

RESULTS AND DISCUSSION

The analysis of the speed curves showed that the subject had developed an average speed of 6.89 ± 0.64 m/s at the end of the 30m dash which was found to be a proper stimulus. Trying to brake at higher speeds caused pain in the quadriceps. However, such speeds may be tried with subjects of lighter masses.

Table(1) Means and Standard Deviations of the Experimental Group in the Pre-Test.

Variable	Mean	S.D.
Broad Jump (m.)	2.29	0.17
Vertical Jump (m.)	0.40	0.09
Power (kg.m/s)	84.03	8.52

The mean deceleration time was 0.704 ± 0.36 s. Regarding the mean values the following parameters could be estimated: Horizontal momentum of 46 Kg.m/s , deceleration of 9.78 m/s/s, and the force needed to brake this momentum 65.35 N

The performance of the experimental group in the pre- test is menfisted in table (1). Statistical analysis as shown in table (2) revealed a significant positive effect of the applied method on the muscular power of the lower extremity in terms of the broad jump distance, height of vertical jump, and the mechanical power developed in the vertical jump when the pre-test was compared with the post test. The means of the differences between the two tests in the three applied measuremnts were significant at 0.01 level. To compare with the control group the the Mann-Whitney test was applied . The z values were 1.64, 1.81, and 2.04 for the broad jump, vertical jump, and power respectively, these values indicate significance of differences between the two groups at .05 level.

Table (2) Means ,Standard Deviations, and T value of Differences between Pre and Post Tests.

Variable	Mean	S.D.	t
Broad Jump (cm.)	4.09	4.53	3.00 **
Vertical Jump (cm.)	3.00	3.03	3.28 **
Power (Kg.m/s)	10.08	11.18	3.06 **

** Siginficant at = .01

The results suggest that the proposed method could serve as a helpful tool in developing the muscular power of the lower extremity.

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**BIOLOGIC COMPOSITE FIXATION OF POROUS-COATED SEGMENTAL BONE DEFECT
REPLACEMENT PROSTHESES IN LIMB SALVAGE SURGERY**

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INTRODUCTION When contemplating limb salvage after bone and soft tissue tumor resection, different reconstructive procedures are available. However, when extracapsular resection is involved and the patient has chosen joint function retainment, prosthetic replacement remains the only viable alternative. Previous experiences using custom-made metallic implants with movable joints were marred with significant complications mainly relating to prosthetic stem loosening and fracture, particularly in the young and functionally active patients. Noncemented fixation based on a press-fit stem, with or without porous coating, may also face a similar fate. In addition, custom-made implants were unable to optimally fit the bony skeletal defect due to tumor resection margin change at surgery. High cost and potential time delay marked other notable shortcomings of prosthetic replacement of bone and joint defects after surgical eradication of bone and soft tissue tumors. The hinged knee joint design imposes high rotatory stress at the implant/cement/bone interfaces which may contribute to stem loosening and fracture. For these reasons, a new prosthetic design and fixation concept was developed. This concept has been carefully investigated using animal models to establish its in vivo performance. Systematic bench tests and biomechanical analyses were also conducted to ensure its safety and efficacy. Initial clinical trials produced excellent results in cases followed beyond two years. Additional studies are in progress to investigate proper autogenous bone graft substitutes to achieve extracortical bone bridging and biological fixation and to identify other enhancing and retarding factors which may influence this new segmental bone and joint prosthesis fixation concept.

IMPLANT DESIGN The new segmental bone and joint prosthesis system adopts a design concept in which various components can be assembled for different anatomical regions and replacement dimensions using a conical coupling joint (Morse taper) with locking screws or key ways to prevent distraction or rotation. The entire prosthetic assembly consists of three major systems: 1) the proximal femur and hip joint system, 2) the proximal humerus and shoulder joint system, and 3) the distal femur and knee joint system. Additional components are available to allow replacements of the proximal tibia and knee joint, a diaphyseal bone segment, or segmental fusion of the knee. The shoulder portion of the prosthetic segment contains porous coating, either in the form of Ti wire mesh or Co-Cr-Mo beads. The intramedullary stems are solid with a fluted cross section for press-fit or cement fixation within the bone cortex. A rotating hinge design with two degrees of freedom was adopted for the prosthetic knee joint. The number of components in each system is flexible and depends upon the size of the stem and the incremental lengths of the bone segment replacement required.

BIOMECHANICAL ANALYSIS AND BENCH TEST Finite element analyses using an axisymmetric model under axial, bending, and torsional loads were performed to demonstrate the efficacy of extracortical bone bridging in achieving the locking strength of the conical coupling joint and the biological fixation between the implant segment and bone cortex at the shoulder junction region. Effective stress transfer from prosthetic segment to bone cortex was successfully demonstrated without the need for intramedullary stem fixation when extracortical bone bridging is achieved. If such bone bridging also occurs over the conical coupling joint, it tends to reinforce the locking strength of the joint and significantly reduces the interface stresses in the conical coupling components.

Conical coupling joint models were tested under bending and torsional loads in repetitive mode. Joint distraction force was used before and after such a fatigue test to study its effect on conical coupling joint locking strength. These tests were repeated with the addition of locking screws. Repetitive axial compression increases the taper joint locking strength. Bending fatigue tends to reduce the locking strength. Under constant axial compression, cyclic torsion did not have any significant effect. With three locking screws, the conical coupling joint nearly doubled its strength. After one million cycles in each loading mode, there was no sign of fatigue damage at the conical coupling joint components. Optimal taper angle was obtained based on the interface locking strength and stress distribution.

IN VIVO ANALYSIS USING CANINE MODELS A canine midfemoral segmental bone replacement model was used to study the effects of the method of initial implant fixation, age, and type of porous coating on the segmental portion of the implant. Demineralized bone matrix was used to substitute for the autogenous bone graft on extracortical bone bridging and biological fixation of the segmental bone prosthesis. The results showed that initial implant fixation and autogenous bone graft are of great importance to achieve successful extracortical bone bridging fixation. Age and type of porous coating did not seem to show any significant effects. Demineralized bone matrix was found to be less effective than the autogenous bone graft to achieve consistent bone bridging and ingrowth. More bone formation and bridging occurred on the compression side than on the tension side in the implant/bone system.

SUMMARY The current segmental bone and joint prosthetic system and fixation concept has several definite advantages: 1) it allows more effective transfer of stress from the implant to bone, 2) it reduces proximal stem and cement interface stresses, 3) it avoids the weakening effects in the porous-coated stem due to the sintering process, 4) cement fixation enhances initial implant stability, and 5) it allows easier prosthetic removal. Initial clinical results appear to support these theoretical advantages. Additional studies and continuous follow-up of the current patient series are necessary to ensure the long-term effectiveness of this concept of segmental bone/joint replacement in tumor surgery.

MEASUREMENT OF NECK RANGE AND PATTERN OF MOVEMENT

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INTRODUCTION Range of motion of the cervical spine has been measured using a variety of techniques. Inclometers, conventional goniometers, and oblique and anterior-posterior radiographs are just a few of the many methods that have been used. The problems of reliably taking accurate data stem from the complicated motion being measured and the lack of easily identifiable bony landmarks in this region. Clinicians need an accurate system for diagnostic purposes and to assess the functional improvement after surgical, therapeutic, and rehabilitative treatments. The purpose of this work was to develop an accurate and reproducible method to determine range and pattern of motion of the cervical spine.

MATERIALS AND METHODS The 3Space™ Isotrak™ System (Polhemus Navigational Sciences Division, McDonnell Douglas Electronics Company) was used to monitor cervical range of motion. The 3Space™ Isotrak™ System utilizes low-frequency magnetic field technology to determine the position and orientation of a sensor in relation to a source, providing full, six-degree-of-freedom measurements. Ten adult male and ten adult female normal subjects were studied. The subjects were seated in a chair modified to hold the 3Space™ source. The shoulders were restrained by a figure-of-eight adjustable strap to minimize trunk motion. The height of the back rest was adjusted so that the source was approximately at T2. The sensor was mounted on a modified leather helmet with a plastic platform which allowed control of the initial flexion-extension angle at the start of the data acquisition (Fig. 1). This system then monitored the head motion with respect to the chair or thoracic trunk which was defined as the cervical spine motion.

The stationary source frame was used as the initial coordinate axes (Fig. 2). The head motion was described using the Eulerian angle system which is based on a Z-Y-X rotation sequence. The first rotation (ϕ) corresponds to flexion-extension, the second rotation (θ) to lateral bending, and the third rotation (ψ) to axial rotation. The subjects were seated upright looking straight forward so that the sensor frame could be adjusted to a neutral position. The following motions were sequentially performed: (1) flexion and extension, (2) right and left lateral bending, (3) right and left axial rotation while the pattern, maximum range, and total excursion in each motion direction were recorded. Paired Student's t-test was used in statistical analysis.

RESULTS The excursion in flexion-extension is larger in female than male subjects ($p=0.007$), but the lateral bending and axial rotation are not statistically different in this study (Table 1). The range of motion for male and female subjects is illustrated for comparison (Fig. 3). A positive correlation existed between flexion-extension and lateral bending excursions ($r=0.745$, $p<0.001$). There exists a coupled motion phenomenon in the cervical spine, particularly in the directions of lateral bending and axial rotation (Fig. 4). Flexion-extension appears to be the only

motion component that can occur independently in normal subjects without significant coupled lateral bending or axial rotation. The displacement of the sensor has a similar pattern when compared to the angular rotation data. This displacement of the sensor has a similar pattern when compared to angular rotation data. These displacement data can be translated to any anatomic landmark on the head through a linear transformation of the reference axis origin.

DISCUSSION The majority of cervical motion assessment is still performed by subjective methods. The 3Space™ provides a means to objectively evaluate the range of motion of patients. This device also allows clinicians the possibility of analyzing the patterns of motion dynamically and comparing them with the characteristics of normal motion. It is important to note that the current method measures only the gross motion of the entire cervical spine. However, a combination of angular and linear displacement data may help to associate various abnormal motion with specific pathology of the cervical spine.

Table 1 Range of motion (Mean (S.D.), Degrees)

	Male (n = 10)	Female (n = 10)	Total (n = 20)
Flexion (ϕ)	-54.2 (9.3)	-63.6 (7.9)	-58.9 (9.7)
Extension (ϕ)	58.6 (10.1)	73.0 (14.4)	65.8 (14.2)
R Bending (θ)	-45.7 (5.8)	-47.5 (8.3)	-46.6 (7.0)
L Bending (θ)	43.0 (7.4)	52.4 (11.3)	47.7 (10.5)
R Rotation (ψ)	-74.3 (5.0)	-76.0 (10.1)	-75.1 (7.8)
L Rotation (ψ)	75.6 (9.4)	81.7 (9.9)	78.7 (9.9)



Fig. 1

Eulerian angles (Z-Y-X rotation)
1st rotation (ϕ): Flexion-Extension
2nd rotation (θ): Lateral bending
3rd rotation (ψ): Axial rotation

Fig. 2

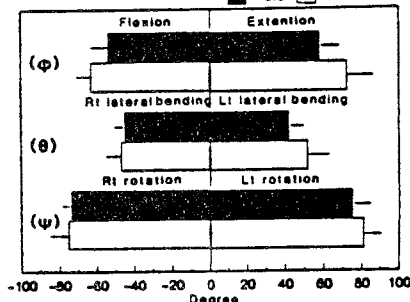
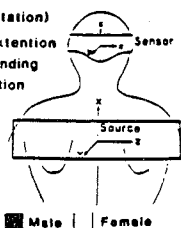


Fig. 3

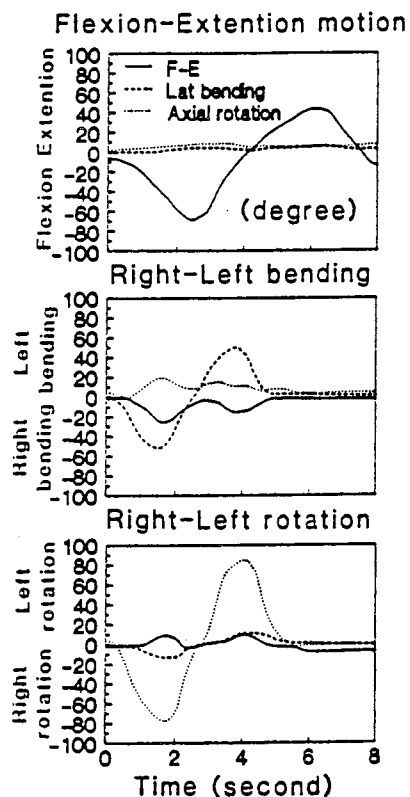


Fig. 4

**THE DEVELOPMENT OF A SIZE AND SHAPE MATCHING INDEX FOR
OSTEOCHONDRAL ALLOGRAFTS**

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INTRODUCTION The outcome of osteochondral allograft joint reconstruction procedures is dependent on many factors, not the least of which is selection of a graft of the proper size and shape to fit the recipient site. Graft bones have traditionally been chosen by visual inspection of radiographs of the recipient site and donor bones. Because it is unlikely that any one bone bank or consortium of bone banks will have a bone of identical geometry to that of the recipient, it is necessary to identify those geometric parameters in the bone that most significantly affect the biomechanical performance of the joint and to weigh the parameters according to their significance so that an optimal graft can be chosen. This study investigates the three-dimensional geometry of the distal femur and simulates hemicondylar allograft reconstructions with the goal of producing a clinically useful size and shape matching index.

MATERIALS AND METHODS The study has four stages: the collection of linear measurements and parametric descriptions of the distal femur in two and three dimensions, the simulation of hemicondylar reconstructions during which pressure distribution between the articular surfaces is calculated, the simulation of joint kinematics using the same reconstructions, and finally, the generation of a selection index or equation with which to make optimal graft selections.

Ten frozen femora were scanned in a CT scanner with section separations of 1.5 mm throughout the full axial depth of the condyles. From these CT scan data, the femora were graphically reconstructed in three-dimensional projections and shaded surfaces on a computer work station running the Analyze[®] software package. Linear measurements were taken from a distal-proximal view of each femur: medial condylar width, intercondylar width, lateral condylar width, greatest AP depth of each condyle, maximum condylar width, and the angle of inclination between the lines of greatest depths of the condyles (Fig. 1). The three-dimensional coordinates of the articulating surfaces of each condyle were extracted from the CT data and supplied to the computer program that approximated the radii of spheres which best fit each condyle, the radii of circles that best fit sagittal views of each condyle, and the major and minor radii of tori that best fit each condyle. These operations utilized least squares fitting techniques.

Simulated hemicondylar reconstructions were performed utilizing three-dimensional reconstructions of the knee joints of several subjects. Hemicondylar resections were performed on each computerized knee model, and grafts from the computerized bank of cadaver femora successively placed in the same recipient knee. In each reconstruction, the pressure distribution in the contact regions was calculated by software utilizing the "rigid body - spring model" technique developed by Kawai (1). Resultant forces and the spatial orientation of the knee model were calculated assuming a one-legged stance.

RESULTS AND DISCUSSION The various geometric parameters measured in ten frozen femora are tabulated (Table 1). A simulated hemicondylar reconstruction with super imposed pressure distribution is shown in Figure 2. The parameters describing the morphology of knee specimens cannot be ranked according to significance without some performance criteria. Restoration of normal joint pressure distribution within the knee is one such criterion, but improvements are clearly needed to make the simulations more realistic. Kinematic analysis needs to be combined with existing pressure distribution analysis so that the forces across all the contact surfaces throughout a complete cycle of gait or other activity can be modeled. Then, statistical techniques will be used to relate the performance grades of the simulated reconstructions to the measurements taken from the recipient and graft bones and a ranking and weighting of the most relevant measurements will result.

Table 1

<u>Measurement</u>	<u>Mean</u>	<u>S.D.</u>
(A) Maximum condylar width	81.90 mm	3.81 mm
(B) Medial condyle width	29.99	2.45
(C) Medial condyle AP depth	68.44	4.08
(D) Medial sagittal radius	34.07	2.38
(E) Medial spherical radius	44.71	3.26
(F) Medial torus major radius	38.57	5.80
(G) Medial torus minor radius	17.49	2.10
(H) Lateral condyle width	32.14	2.29
(I) Lateral condyle AP depth	69.48	6.01
(J) Lateral sagittal radius	35.48	2.93
(K) Lateral spherical radius	43.47	4.07
(L) Lateral torus major radius	40.40	6.59
(M) Lateral torus minor radius	17.67	1.99
(N) Intracondylar separation	20.53	1.98
(O) Condylar inclination	12.05 deg	2.32 deg

FEMORAL MEASUREMENTS

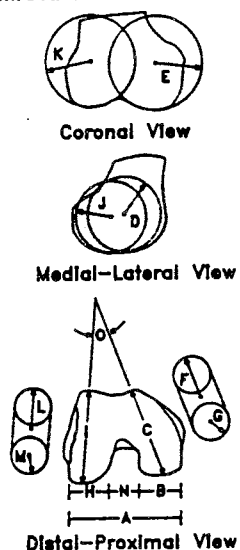


Fig. 1

LATERAL CONDYLE RECONSTRUCTION

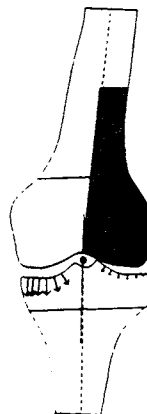


Fig. 2

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ACKNOWLEDGEMENT This study is supported in part by NIH grant CA 40583.

THE EFFECT OF RUNNING SPEED ON LOWER EXTREMITY
JOINT MOMENTS DURING THE SUPPORT PHASE

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Many running injuries are linked to anatomical and/or biomechanical factors associated with the support phase (1). Increased running speed requires changes in the running mechanics that serve to increase the musculoskeletal stresses (2). Joint moments (JM) are useful indicators of the amount of physical stress placed on the neuromuscular system. The purpose of the study, therefore was to investigate the effects of running speed on lower extremity joints generated during the support period to gain a better understanding of the individual adaptation strategies utilized during the impact phase and the motive strategies employed during the propulsive period.

Four equally skilled male runners volunteered as subjects. The experimental setup consisted of a force platform (552 Hz) embedded in the middle of a 40 m runway and a 16 mm (100 Hz) and an 8 mm (50 Hz) high speed movie camera. Appropriate anthropometric data were obtained using a modified Hanavan model (3). Eight trials of data were obtained for each of 4 running speeds (3.06, 3.57, 4.09 and 4.60 m/s $\pm$ 6%). Kinematic data were smoothed and higher order derivatives were generated using optimally regularized Fourier series. Kinematic and kinetic data were normalized in time by generating 101 estimates. Moment values for the ankle (AJ), knee (KJ) and hip (HJ) joint were calculated using a standard Newtonian model and normalized to body weight and height for comparison purposes. Implied muscular activity patterns were derived by calculating power values (4). Peak JM values and relative and absolute times to critical JM and muscular activity events were statistically evaluated ($p < .05$) using a within-subject statistical procedure (Model Statistic) (5).

The percentage of significant comparisons (SC) is shown in Table 1. Overall, JM parameters were found to increase with increased speed; however the pattern of the increases was subject and parameter dependent. A trend of increased plantar flexor JM was demonstrated by all subjects but was significantly demonstrated by only two subjects (S1 and S2) (Figure 1). The trend of increased plantar flexor moments not only served to decrease the braking and vertical ground reaction forces, but enabled a greater amount of elastic energy to be stored during the eccentric period that could then be utilized during the propulsive phase when the plantar flexor moment reached its maximum value.

Only 1 subject (S1) exhibited significant increases in maximum KJ flexor moments although S2 and S4 exhibited non-significant increases. The observed concentric KJ flexor moments could function to accommodate the vertical ground reaction forces and assist the hip extensors in reducing the relative horizontal velocity of the foot at contact and during the initial support phase. Maximum KJ extensor moments increased (16.7-66.7% SC) for all subjects. Eccentric extensor activity serves to arrest the negative vertical velocity of the body. Increased concentric KJ extensor moments also contribute directly to increased propulsive forces.

The hip JM parameters exhibited the greatest number of significant increases with speed, (59.3% SC) compared to the knee (27.0% SC) and ankle (25.0% SC). The first maximum HJ extensor moment exhibited the greatest number of SC (79.2%) of all the JM parameters, suggesting that the HJ activity performs an important role during footstrike.

Increased concentric HJ extensor moments were used to reduce the relative braking velocity of the foot during impact. Speed-related increases of HJ extensor moments exhibited during the entire loading phase substantiate the hypothesis that the hip joint musculature 'pulls' the trunk and other segments over the foot in preparation for the propulsive phase. Increased HJ extensor moments are also required to transport the thigh through a greater range of motion at the faster running speeds. Increased eccentric maximum HJ flexor moments (16.7-66.7% SC) were generated during the latter portion of the support phase to counteract the increased momentum of the lower extremity. The observed increased range of motion also could increase the relative contributions of passive elements.

Absolute times decreased significantly (50.0% SC), although most of the relative times remained the same for the speeds evaluated (13.2% SC) suggesting that the relative timing of events was invariant with speed. The timing of the critical events moved from the distal to the proximal joints during the initial support phase and from the proximal to the distal joints during the propulsive phase, indicating efficient strategies for attenuating impact forces and increasing propulsive forces.

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Table 1. Percent significant increases of JM parameters

Parameter	Subject			
	1	2	3	4
Max plantar flexor	50.0	50.0	0.0	0.0
Max knee flexor	83.3	0.0	0.0	0.0
Max knee extensor	16.7	33.3	66.7	16.7
First max hip extensor	83.3	83.3	83.3	66.7
Max hip flexor	16.7	33.3	66.7	66.7

Note. Percentage based on comparisons of all speed condition combinations.

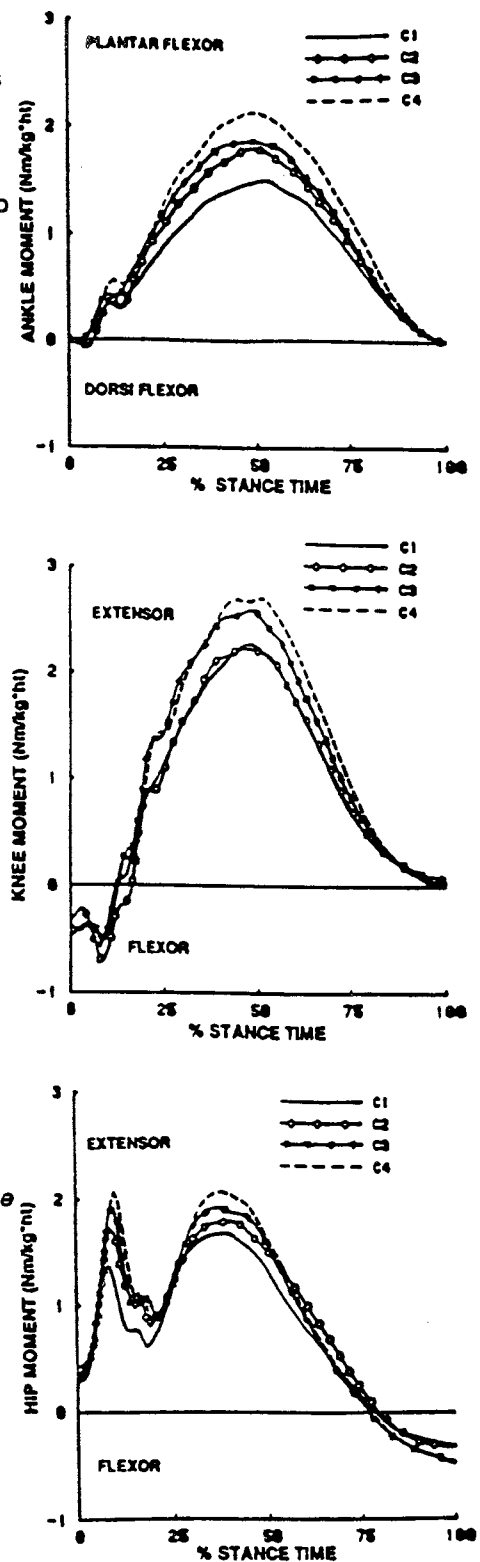


FIGURE 1 Ensemble joint moment curves for all speeds for subject 1.

CALF MUSCLE WORK AND SEGMENT ENERGY CHANGES IN RUNNING

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In this study the function of the calf muscle group (*M. triceps surae*) in running was investigated by comparing the work delivered by the muscles with changes in potential and kinetic energy of the body segments.

In previous work on walking (Nauta and Hof, 1987) it was shown that positive work of the calf muscles is mainly instrumental in increasing the kinetic energy of the leg at the onset of swing. There was no relation with trunk energy, except at very large step lengths. The so-called 'push-off' of the calf muscles, therefore, does not produce an acceleration or upward movement of the trunk in walking.

The present study deals with the same problem in running.

METHODS

Potential, kinetic and rotational energy were measured for a 7-segment body model by means of potentiometers and tachometers driven by strings attached to various points on the body. Calf muscle work was determined by EMG to force processing (Hof and Van den Berg, 1981).

There were four male subjects aged 27, 28, 29 and 52 years, with weights between 63–79 kg and statures between 1.73–1.89 m. They practised sports, but had no specific training in running. They ran on a treadmill at speeds between 1.75 and 2.75 m/s, which may be called jogging speeds.

RESULTS

In contrast to the situation in walking, in running potential and kinetic energy of the HAT segment are in phase (Cavagna et al., 1976), resulting in total energy changes of 50–130 J twice each stride (Fig. 1, upper curve). The changes in energy of the leg, the sum of foot, shank and thigh energies, ranged from 31–102 J. Remarkably, the major leg energy increase comes only after toe-off, in contrast to walking, where leg energy is already maximal at that time.

Calf muscle activity is restricted to the stance phase. It begins with negative work, 15–81 J, and ends with a still greater amount, 44–89 J, of positive work (Fig. 1, middle). Work and energy values showed large individual differences.

When work and energy curves are compared, it turns out that positive calf muscle work corresponds closely in time and magnitude with the last 2/3 of the increase in HAT energy. The first half of negative calf muscle work corresponds more or less with the decrease of both HAT and leg energy; for the latter half there is no such correspondence.

DISCUSSION

Regarding the positive calf muscle work, it is obvious that it serves to move the trunk upward and forward in the final phase of stance. A similar relation for the negative work phase remains less clear, unless the action of the knee extensor muscles is considered as well. In running these muscles perform an eccentric-concentric contraction similar to the calf muscles, but slightly earlier in time (cf. Nilsson et al., 1985).

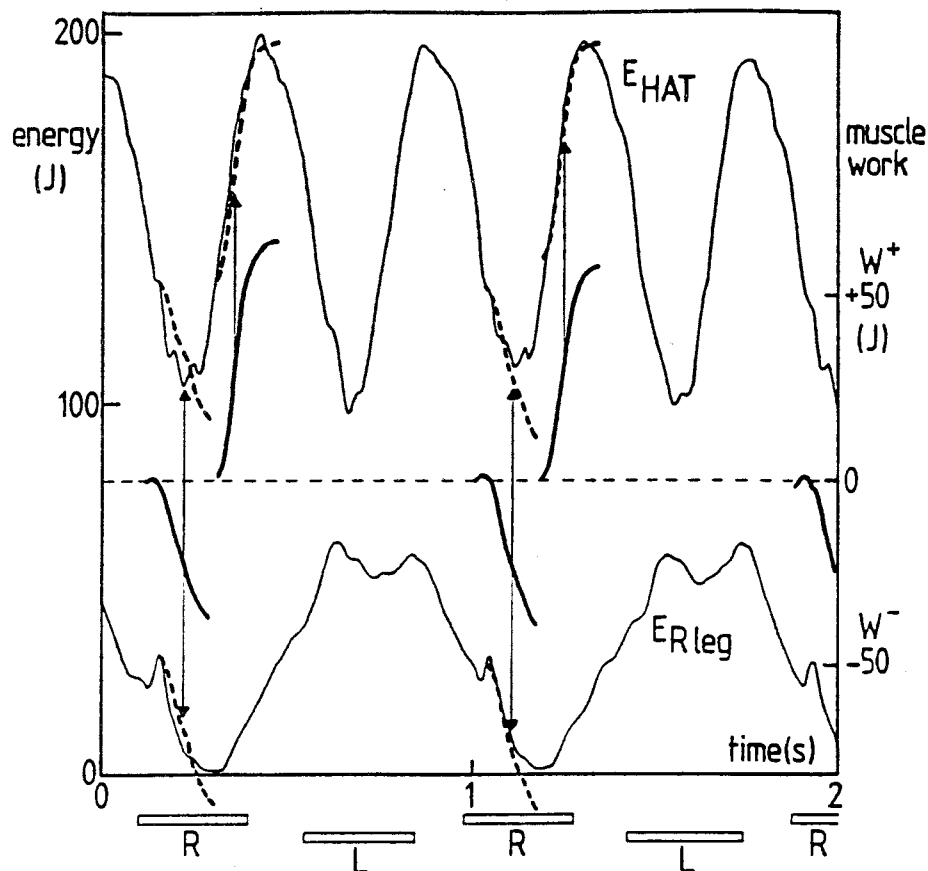


Fig. 1 Energy levels of Head-Arms-Trunk segment (upper curve) and right leg (lower curve). Calf muscle negative (W^-) and positive work (W^+), middle. Foot contact left and right foot below. The W^- and W^+ curves have been shifted upward and downward to see if they coincide with segment energy changes (dashed lines).

The picture emerging is that in stance the falling energy of HAT and legs is absorbed first by the knee extensors and later by the calf muscles. In the upward and forward movement, corresponding to the increase in HAT energy, again the knee extensors are first, followed by the calf muscles. In the time in between there may be a transfer of positive knee extensor work via the moving HAT segment to negative calf muscle work. Such a sequential activation from proximal to distal muscle groups is also seen in jumping (Bobbert and Van Ingen Schenau, 1988).

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DETERMINING MUSCLE FORCE AND ENERGY IN THE LEG DURING HUMAN WALKING
USING THE INVERSE DYNAMICS METHOD

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INTRODUCTION

The walking patterns of leg-amputees using prostheses, differs from the gait of non-amputees.

Important questions in analyzing these patterns are: why they are different, what are the consequences of these differences and what are the influences of the construction, fitting and alignment of the prosthesis on gait. This requires not only a consideration of the differences in body-movement, joint moments and forces, but also the muscle activity and forces has to be known to predict metabolic energy consumption and endurance time.

In the past many methods were developed to derive the individual muscle forces from the instant joint moments and forces. However most of these did not take into account the influence of previous time steps on momentary muscle forces and the fact of biarticularity of some muscles.

To predict muscle forces, muscle stimulation patterns and energy consumption from joint moments and forces a dynamic optimization technique as proposed by Davy e.a. [2] was used for one whole leg during one complete locomotion cycle. Three different muscle models were used and the results were compared with data from literature.

METHODS

The calculated torques represent the requirements for dynamic equilibrium placed upon the muscles and ligaments of the musculo-skeletal system. Since the lower limbs possess a mechanically redundant set of muscles [4], direct solution of the muscle force, moment-arm joint-torque equations and consequently the muscle force-joint reaction equations, is not possible. An approach of dealing with this problem has been the use of optimization methods in which the equipollence relations at any point in the motion are solved on the basis of optimizing some criterion such as the total muscle effort.

Given the joint moments at any instant, the problem becomes that of finding the muscles activities that creates the necessary moments. The unknown muscle activities are found using optimization techniques. The optimization problem can be stated as follows. Minimize an object function corresponding to a measure of muscle effort, and subject to equality and inequality constraints corresponding to the joint moment equipollence relations, and the tensile nature of the muscle forces respectively. The model incorporates 47 muscles groups in each limb [1]. At the onset the swing phase of the right leg begins and the period [0,1] is equal to one total locomotion cycle. The three-dimensional problem is very large and therefore hard to solve, so here only the flexion/extension actions were considered (the problem decreased to only 16 muscle-groups). When the subject has a symmetric gait, the problem can be reduced to only one limb.

However the prediction of the optimal process highly depends on the contraction and excitation description of the muscle model. The dynamic

muscle model incorporates a single control parameter and several features of excitation and contraction dynamics [2] [4]. In the context of this problem three muscle models of varying complexity have been considered. The models incorporate both passive and active characteristics of muscle behavior. The differences in complexity represent differences in the number of features and realism of these features in the light of observed physiological behaviors. The active muscle force generator is known to be affected by the length and the contraction velocity of the muscle, as well by the degree of excitation. The normalized force-velocity relationship is given by either a linear or a hyperbolic force-velocity relationship [3] [4]. The excitation dynamics are described by Hatze [4] or Davy [3] and so history dependent. The instant muscle control parameter was calculated from fourier series. By using this technique the instant individual muscle force distribution can be calculated.

RESULTS

Using a fourth order Runge-Kutta method, the optimal problem was solved using a quasi-newton method. The integration step-size depends of the model equations. The calculations of this reduced problem were successful. However the numerical instability of some models prevents a fast solution. The optimization requires about 200 steps to generate the solution. The calculated total muscle energy is 260 W/cycle (speed 1m/sec), which seems realistic.

.. muscle moments
— required

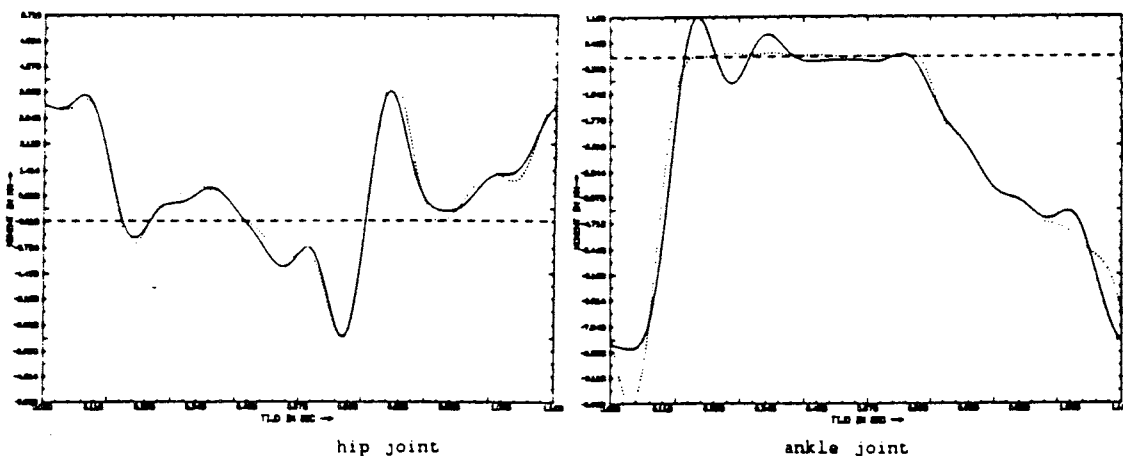


Fig. 1 The flexion/extension joint moments in the right leg during normal walking (1 m/s)

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SYSTOLIC AND DIASTOLIC FORCE MEASUREMENTS OF MITRAL ANNULUS

Zhang Ru-yi, Wang Ke-pang, Yu Jian, Wang Yun-ru, Tang Yue, Wang Ji-hai

Since the mitral annulus was used for mitral valve correction, the recurrence of mitral regurgitation has been reduced. However some problems still remained such as the magnitude of force acted on the mitral annulus at systolic or diastolic condition, the dynamic performance of mitral annulus. It will be very important for designing a new type mitral annulus.

A new method for measuring the force affected to mitral annulus is described in this paper. The experimental research is made by using seven dogs. In this experiment the systolic and diastolic forces of mitral annulus at various hemodynamic efficiencies are obtained. The results of the experiment exhibit the regularity and validity.

THE METHOD

A special designed circular strain gage type transducer whose elasticity and size are similar as that of mitral annulus is developed and calibrated before testing. The prime testing object is the dog. After an operation of anaesthesia and external body circulation had made and heart failure had appeared, the left auricle was cut and a transducer would be set to the mitral annulus of dog's left atrium. (see Fig.1) Sewing up the incision, the heart recovered and the signal of transducer is input to the signal processor. The systolic and diastolic force of mitral annulus can be measured according to the calibration data. It is to be noted that the single is negative (tension) at diastolic and positive (compression) at systolic.



Fig.1 set transducer to the mitral annulus

DATA ANALYSIS

The experimental research is made by using seven dogs (weight 20 Kg). The systolic and diastolic force and electrocardiogram curve of dog after

recovered are shown in Fig.2. The signal waveform of force measured by transducer is a dual-peak curve and both peaks correspond to the systolic times of ventricle and atrium at a cardiac cycle. The systolic and diastolic forces at various hemodynamic conditions such as lack oxygen, normal, increasing myocardial contractility, increasing resistance and volume load exhibit regularity. The average values of systolic and diastolic forces of seven dogs at various hemodynamic efficiencies are shown in table 1.

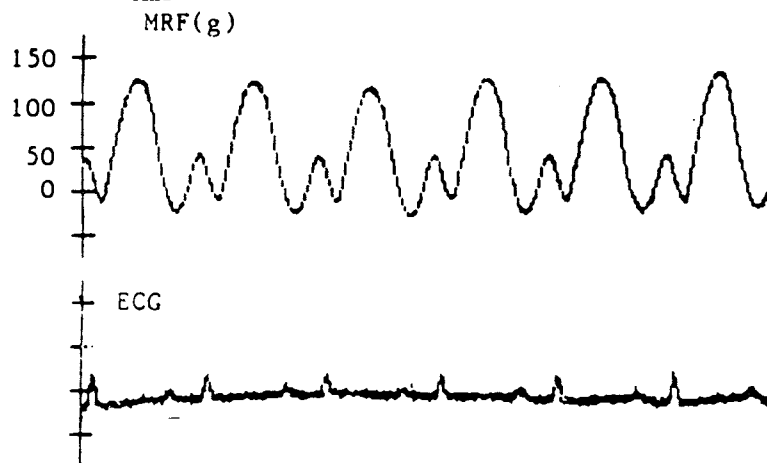


Fig.2 Force-time curve and electrocardiogram

	recorve	normal	myocardial	increasing resistance	increasing value
ventricle systoles	73.9	147.4	276.4	232.6	194.3
atrium systoles	33.6	59.3	83.7	75	51.7
ventricle diasloles	-33.4	-64.4	-130.6	-85	-99
ventricle systoles to diasloles	107.3	211.9	406.7	317.7	293.3

Table 1

(Force Unit g)

CONCLUSION

(1) the pattern of the variation of systolic and diastolic forces presents as a dual-peak curve and both peaks correspond to the systolic times of ventricle and atrium at a cardiac cycle.

(2) the force acted on mitral annulus when ventricle systoles is much large than that acted on mitral annulus when atrium systoles.

(3) all the results exhibit regularity. It will provide an important basis for new mitral annulus designing.

THE ROLE OF THE CORACOCALVICULAR MECHANISM IN THE MOTION PATTERN BETWEEN SCAPULA AND CLAVICLE.

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Acting as a stable but movable base for the humerus, the shoulder girdle plays a leading role in the mobility of the upper limb. Scapular 3D-rotations during the elevation of the humerus do not only enlarge the total range of motion, but also enable the glenoid to remain in a position of firm opposition to the humeral head. The shoulder girdle can be considered as a mechanism consisting of three links. The sternum together with the thorax is the first link, whereas the clavicle and the scapula are the remaining two links. To a large extent, the motions of the scapula are restricted by the clavicle and the thorax. Due to the serratus anterior and the rhomboideus muscles, the medial border of the scapula glides over the scapulothoracic wall. Therefore, the shoulder girdle can be viewed as a closed chain. The system of muscles surrounding the scapula is customary considered to be the source of motion of the shoulder girdle. In that way, the scapula functions as the driven part of the mechanism. The scapula is connected to the clavicle at both the acromion and the coracoid, respectively, by the acromioclavicular joint and the coracoclavicular ligament. What role these connections play in the motion pattern between scapula and clavicle is rather indistinct. Because of the architecture of the coracoclavicular ligament, it is usually divided in two separately functioning parts, viz. the conoid ligament and the trapezoid ligament. The function of the coracoclavicular ligament is analysed on the one hand in studies of prepared specimens (e.g. Liberson, 1939; Fukuda et al., 1986). Then, only the motion range between clavicle and scapula is determined and not the influence of this ligament on the real motion pattern between both bones. On the other, the function of the coracoclavicular ligament is based on 2D-motion studies of the scapula and the clavicle (e.g. Inman and Saunders, 1946; Dvir and Berme, 1978). However, in view of the 3D-structure of the shoulder girdle the results of the 2D-motion studies are more speculative, since motion effects in the third dimension can not be distinguished. In this study, the role of these ligaments is studied by means of 3D-simulations with the help of a kinematic computer model of the shoulder girdle (Pronk, 1988). In this model the clavicle is described as a beam between the sternoclavicular and acromioclavicular joints, whereas the scapula is modelled as a triangular plane with the angulus acromialis, the trigonum spinae and the angulus inferior as angular points. The motions of both bones are derived from palpation measurements of the 3D-positions of

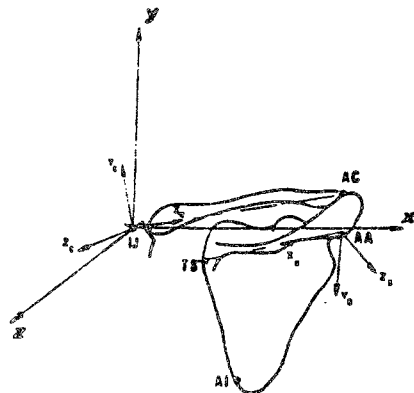


Figure 1:

The definition of the local coordinate systems fixed to the scapula (x_s, y_s, z_s) and fixed to the clavicle (x_c, y_c, z_c), with regard to the global coordinate system (X, Y, Z).

IJ = Incisura Jugularis,
AC = AcromioClavicular joint,
AA = Acromial Angle,
TS = Trigonum Spinae,
AI = Angulus Inferior.

the scapula and clavicle at several angles of humerus elevation (Pronk, 1987). Based on the measurements of 18 male subjects, the 3D-motion pattern of the scapula and the clavicle relative to the humerus elevation could be calculated. The rotations of the scapula and the clavicle are described as rotations of local coordinate systems fixed to the bones about axes as defined in figure 1 and in the sequence Y, z_s, x_s and Y, z_c, x_c , respectively, in correspondence to the common medical definitions. The relations between these rotations and the humerus elevation that were found, are presented in table 1. According to these relations, the motions of the scapula and the clavicle of 7 cadavers were simulated. After measuring the 3D-coordinates of the attachments of the conoid and trapezoid ligaments with regard to the local coordinate systems, the length changes of both ligaments could be calculated during the simulations. Since the axial rotation of the clavicle could not be recorded, it is assumed that the clavicle rotates axially about 40° as stated by Inman and Saunders (1946). Then, it appeared that the length of both ligaments hardly changed. In another simulation following the assumption of Dvir and Berme (1978) that both ligaments are already taut at the begin of the motion, it could be analysed that the conoid ligament mainly influences the magnitude of the axial rotation of the clavicle. Thus, as a result of the lateral rotation of the scapula, the clavicle is axially rotated forced by the conoid ligament. Since the angle between clavicle and the scapular plane measured at 18 male subjects varies between 60° - 75° during elevation of the humerus, it can be concluded that the conoid ligament probably restricts the torsion in the capsule of the acromioclavicular joint. In view of the orientation of the trapezoid ligament, it can be hypothesized that this ligament restricts translations in the acromioclavicular joint. However, by assuming the acromioclavicular joint in the model as a spherical joint, this hypothesis could not be proved by analysing the small changes in the length of the ligament.

Table 1: The relations between the three rotations (in degrees) of both the scapula and the clavicle relative to the elevation angle α (in degrees) of the humerus, based on 18 male subjects.

		forward flexion	abduction
scapula	protraction (Y -axis)	$-2.86+0.35\alpha-0.0024\alpha^2$	$-8.7-0.04\alpha-0.0008\alpha^2$
	lateral rot. (z_s -axis)	$0.54-0.11\alpha+0.003\alpha^2$	$-1.28+0.13\alpha+0.0015\alpha^2$
	tipping rot. (x_s -axis)	$-0.13-0.04\alpha-0.0004\alpha^2$	$-2.67+0.13\alpha-0.002\alpha^2$
clavicle	protraction (Y -axis)	$-2.3+0.22\alpha-0.002\alpha^2$	$0.17-0.18\alpha$
	elevation (z_c -axis)	$-0.05-0.47\alpha+0.001\alpha^2$	$-1.68+0.19\alpha-0.0007\alpha^2$
	axial rot. (x_c -axis)	not measured	not measured

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HUMAN JOINT HYBRID CONTROL MODEL

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Functional electrical stimulation (FES), both in efferent and in afferent mode, has been subject of intensive study and research in the course of the last 20 years or more. Encouraging results were obtained, some of them practically applied, although wider spread out notably in the case of more complex and prolonged motor activities were limited by two main factors:

- non-repeatable response of the stimulated muscles to FES,
- fast fatigue of muscle groups during prolonged activity.

The idea of the so-called "hybrid joint" is not a new one [1]. But since now, some attempt to develop a hybrid control stayed unknown.

Hybrid model proposed herein includes three separate parts: mechanical part of the system $S\{M\}$, external actuator $S\{E\}$ and internal (muscular) actuators $S\{I\}$. $S\{E\}$ is represented by a d.c. servo motor [2].

The EFES-response mapping of a skeletal muscle can be stated in the form of the force generator time behaviour $S\{I\}: R_t \rightarrow \text{Hom}_R\{S, C\}$ where: t denotes stimulation time, S and C correspond to the left R -moduli of stimulation and contraction respectively. The mapping F can be considered as an effect of a fifth-order diffusion transmission cascade: neural action potential, synaptic potential, muscular action potential, excitation-contraction coupling and muscle tension generating. These summary processes (for all included motor units in the particular contraction) can be described by fifth order recurrent distributed parameter diffusion system

$$C_i \frac{\partial V_i}{\partial t} = \frac{1}{R_i} \frac{\partial^2 V_{i-1}}{\partial x_i^2} - J_i(V_i), \quad (i=1, \dots, 5)$$

with boundary condition at $x=0$

$$V_i(0, t) = V_0 \cdot \sin(2\pi f t) = \text{STIM}(t).$$

For muscle-mechanical purpose, the presented distributed force generator can be approximated with the corresponding lumped-parameter one at $x=\text{Tend}$

$$\begin{aligned} \dot{x}_i &= \frac{1}{T_i} (b_i x_{i-1} - x_i), & x_i(0) &= 0, \quad (i=1, \dots, 5), \\ T_i & & x_0 &= \text{STIM}(t), \quad x_5 = F(t). \end{aligned}$$

Newton equation of uniaxial joint motion representing an automorphism $S\{M\}: R_0 \times R_0 \times R_t \rightarrow R_0$ on the Lie's $SO(2)$ group is proposed in the linear viscoelastic form

$$J\ddot{\Theta} + B\dot{\Theta} + C\Theta = \sum_{k=1}^2 \delta_k (F_k \cdot r_k - mgh \sin \Theta) + F(t-T_p),$$

$$\Theta(0) = 0, \quad \dot{\Theta}(0) = 0,$$

where: J-combined moment of inertia of the actual body segment and external actuator output shaft, B-combined viscous friction of the actual joint tendon and motor output shaft, C-tendon elasticity coefficient, $\delta_k=1$ ($k=1$) for agonistic muscle and $\delta_k=-1$ ($k=2$) for antagonistic one, $r_k=r(\Theta)$ - joint lever arm of the actual muscle tendon, h - segment length and T - the motor torque time lag const. corresponding to 70% of $F(\max)$.

In the present study, the mathematical model of a single joint hybrid control was given in the form of the 13.th order linear differential Cauchy's system. A joint with predominantly one degree of freedom was selected (elbow or knee). Basic energy was obtained by efferent electrical stimulation of the natural actuator in the given joint. Simultaneously and constantly stimulation (STIM) of both the m. agonist and m. antagonist was performed, whereby m. agonist was stimulated by Heaviside's rectangular input and m. antagonist by concave exponential excitation. External energy was actuator supplied by an electromechanical actuator of Institute "Mihailo Pupin", already realized for the active orthosis.

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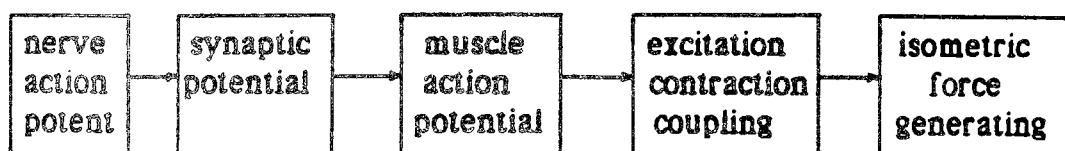
RECURRENT DIFFUSION MODEL OF ISOMETRIC MUSCLE ACTION SIMULATIONS

Vladimir Ivančević and Radoslav K. Andjus

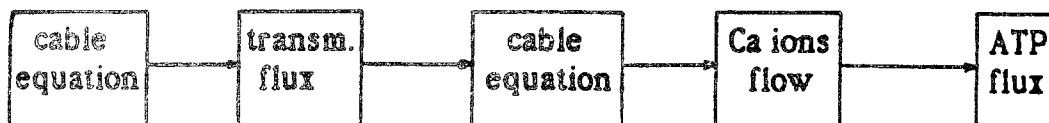
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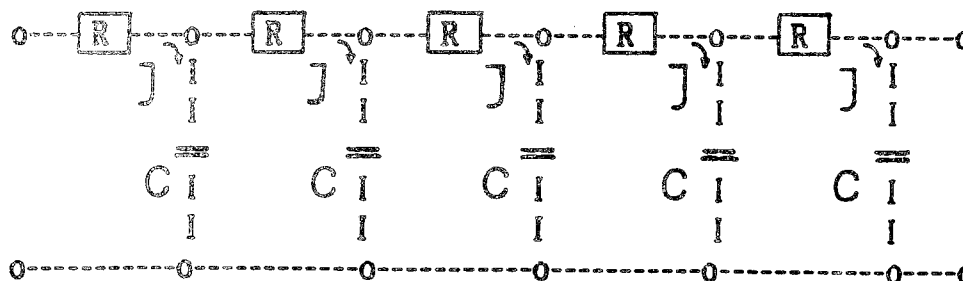
A solution of both electro-physiological and mechano-chemical excitation-contraction control process existing inside skeletal muscle isometric contraction is proposed in the form of the fifth-order recurrent (distributed parameter) diffusion system and its lumped-parameter approximation in the ordinary Cauchy form. The model includes five cascade elements: neural action potential, synaptic potential, muscular action potential, excitation-contraction coupling and isometric tension generating, according to the block-diagram



where stimulation (STIM) is applied either indirectly (to the nerve) or directly (to the muscle). We state the following hypothesis according to [1]: all of these transmission processes can be considered as being some kind of diffusion processes. Therefore, we have the following distributed parameter transmission cascade



and corresponding (the most simple case) lumped parameter approximation cascade



where the last circuit corresponds to the low-pass filter representing the contraction process itself, and the second one corresponds to the Eccles circuit model of synaptic activation.

The isometric stimulus-response contraction mapping can be stated in the form of the force generator time behaviour $F: R_t \rightarrow \text{Hom}_t(S, C)$, where: t denotes stimulation time, S and C correspond to the left R -moduli of

stimulation and contraction respectively. These summary processes (for all included motor units in the particular contraction in the form of either single twitch, incomplete or complete tetanus) can be described by fifth order recurrent distributed parameter diffusion system

$$C_i \frac{\partial V_i}{\partial t} = \frac{1}{R_i} \frac{\partial V_{i-1}}{\partial x^2} - J_i(V_i), \quad (i=1, \dots, 5),$$

where $J_i(V_i)$ denote nonlinear terms analogous to those described in [2] or [3], with boundary condition at $x=0$

$$V_i(0,t) = V_0 \sin(2\pi f t) = \text{STIM}(t).$$

The single element behaviour is now given by

$$V_i(x,t) = V_0 \exp(-x_i/\lambda) \sin(2\pi f(t-x_i/\gamma)), \quad \lambda = 1/(\pi C R f), \\ \gamma^2 = 4\pi f/(C R).$$

For muscle-mechanical and state-space control purpose, the presented distributed force generator can be approximated with the corresponding lumped-parameter one at $x=\text{Tendon}$

$$\dot{x}_i = -\frac{1}{T_i} (b_i x_{i-1} - x_i), \quad x_i(0) = 0, \quad (i=1, \dots, 5), \\ x_0 = \text{STIM}(t), \quad x_5 = F(t).$$

The single muscle behaviour in the lumped approximation form is given by the sum of its transient and weighting term

$$x_i(t) = b_i(1 - \exp(-t/T_i)) \cdot x_{i-1} + x_i \exp(-(t-\tau)/T_i),$$

where $T = CR$ is the element time characteristic (possible time-invariant) and b is the input gain (or conversion factor).

The simulations are performed in two experimentally oriented situations: direct stimulation and indirect (neural) one, with both sinusoidal and Heaviside rectangular excitation (5v, 10ms). Simulation results range from the single twitch to the complete tetanus. The simulations are executable on the IBM PC family computers.

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SOME FACTORS AFFECTING ECCENTRIC CONTRACTION STRENGTH

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A special dynamometer was designed to measure isometric, concentric and eccentric contraction force exerted on various joints of a human body. This dynamometer has a motor-driven lever-arm which rotates around an axis at a constant speed. The purpose of the present study was to investigate properties of the force exerted during eccentric contractions using this dynamometer.

Method

Subjects were six healthy men. Each subject was seated on a chair with the right upper arm rested on a support bench horizontally and the forearm supinated. The lever arm was attached to the right wrist with a cuff. A force transducer was placed on the wrist to measure elbow flexion force. Each Subject was asked to flex their right elbow at the maximal voluntary contraction level. Immediately after an isometric contraction of 2 seconds, the elbow joint was eccentrically stretched about 15°, then again put in an isometric state at that elbow angle. Starting elbow angles were randomly assigned every 15° from 70° to 130°(full extension 180°) and stretch speeds ranged from 2.5 to 20 rpm.

Results and Discussion

Elbow flexion force during an eccentric contraction was always greater than those during isometric contractions. The angle (PA) at which elbow flexion force reached its peak during eccentric contractions was $92^{\circ} \pm 13^{\circ}$ (mean $\pm$ S.D.). The difference between starting isometric force and eccentric peak force was larger at a higher speed. In trials where final elbow angles were smaller than PA, peak-force elbow angles during eccentric contractions were closer to the final elbow angles, whereas in other trials where starting elbow angles were larger than PA, peak-force angles during eccentric contractions were closer to the starting elbow angles (Fig.1). If a non-linear least square regression was performed on isometric values, the gaps between peak elbow flexion force during eccentric contractions and the point on the regression curve were almost the same in the tested angle range (Fig.2). Increase of stretch speed enlarged the gaps. After an eccentric contraction, elbow flexion force gradually decreased to isometric value although the elbow was no more being stretched (Fig.3). This may be due to the release of stored elastic energy and/or enhanced contractile component (Cavagna et al., 1968, Thomson and Chapman, 1988), which needs further investigation.

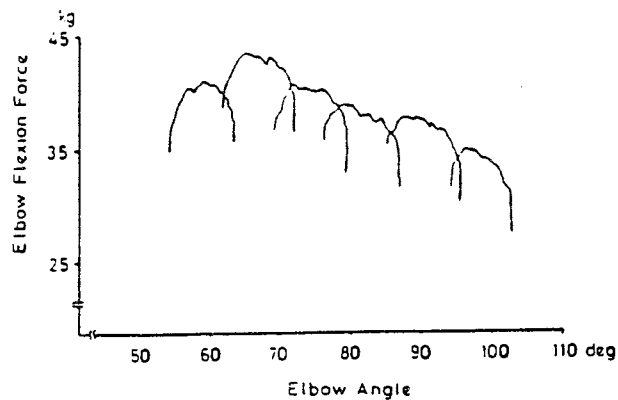
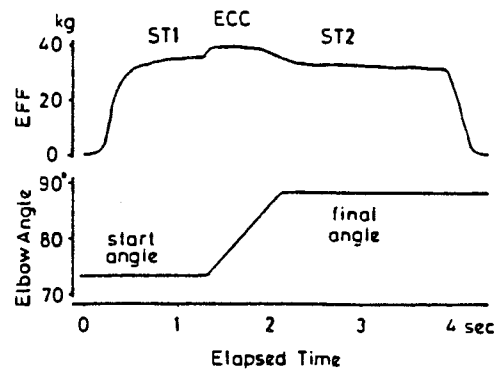


Fig.1



ST1, ST2 : Isometric Contraction
 ECC : Eccentric Contraction
 EFF : Elbow Flexion Force

Fig.3

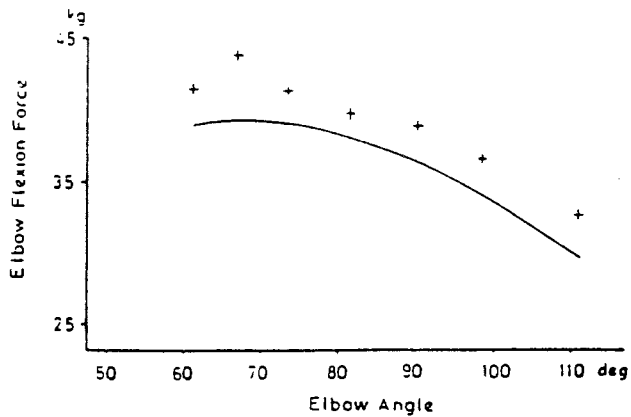


Fig.2

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MOMENT ABOUT THE VERTICAL AXIS DURING THE BASEBALL BATTING

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Studies have shown that the moment about the vertical axis generated by the feet (M_z) has an important role in the hitting skill (Carlsöö, 1967; Cooper et al., 1974). Therefore, it is considered to be essential for the effective hitting performance to investigate the factors which influence the magnitude of M_z . The purposes of this study were to examine the relationship between the M_z and the body weight transfer during the baseball batting, and to reveal the effect of muscle strength on the peak M_z .

Procedures

Twenty six right-handed university baseball players (mean $H_t=1.74m$, mean $W_t=71.2kg$) participated as subjects in this study. Each subject was asked to hit a fixed baseball with a bat ($1.02kg$) on a $0.6 \times 1.2m$ force platform (AMT Inc.). Stride length (SL) was measured as the distance in the pitcher-catcher direction between both balls of the foot after the swing. The displacement of the center of pressure (CP) which acted on the platform during the swing was defined as the body weight transfer. The M_z in relation to the position of CP was plotted every 10msec from the backswing phase to the follow through. The ratio of the total displacement of CP to the SL and the position where the peak M_z was observed in the displacement of CP were also calculated. In addition, the isometric muscle strength in extension at knee joint (joint angle= 110°) was measured by using the Cybex machine.

Results and Discussion

SL was $0.79 \pm 0.06m$ (mean $\pm$ SD) and the total displacement of CP corresponded to $84.0 \pm 13.8\%$ of SL. Typical pattern of M_z in relation to the relative position of CP is illustrated in Figure 1. As SL depended on the body size, relative value was adopted for the position of CP. The peak M_z was observed at 100-200msec before impact when the CP was transferred at $71.1 \pm 10.6\%$ of the total displacement. Whereas, negative value in M_z was observed at impact. These results indicated that with loading the body weight on the left foot, the batter generated the M_z in the early phase of the forward swing. Mason (1987) reported that the peak in the anterior force by the left foot of the batter was larger than that in the posterior force by the right foot. Therefore, it was suggested that the left foot contributed more to the magnitude of M_z than the right foot.

The relationship between the isometric muscle strength in extension at knee joint and the peak M_z is shown in Figure 2. Significant correlations ($r=.532$; $p<0.01$) between these two variables were obtained. According to Messier and Owen (1985), left knee extension continued during the swing, with a rapid increase occurring during the 0.10sec prior to impact. But, as the displacement of the knee angle was small, isometric muscle strength was considered to be the index of the peak M_z .

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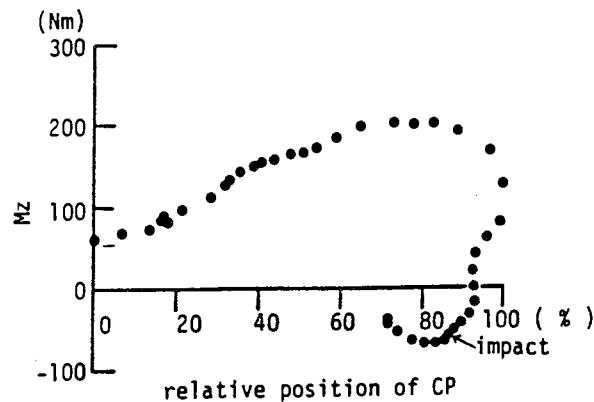


Figure 1. Typical pattern of Mz in relation to the relative position of CP.
0 and 100% indicate the backward-most and the forward most position of CP, respectively.

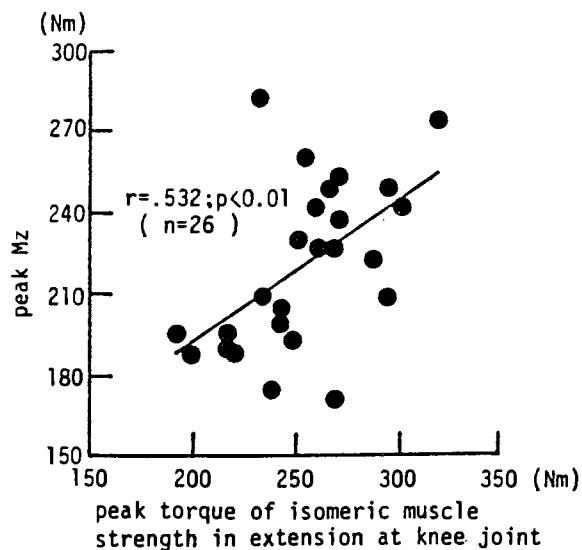


Figure 2. Relationships between the isometric muscle strength in extension at knee joint and the peak Mz

THE DEVELOPMENT OF GAIT FROM THE VIEW POINT OF TEMPORAL AND DISTANCE FACTORS

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The purpose of this study is mainly to investigate the course of development of gait from elementary school students to junior high school students using temporal and distance factors of walking. To compare the gait pattern of the elementary and junior high school student with adults, gait studies on twenties and thirties were performed, and the time of maturation of gait was discussed. The number of investigations on the gait of the old is small(1), therefore this study attempts to show the differences of gait between twenties and the old.

METHODS: One hundred and forty-six normal male subjects old were studied. The distribution by age and their physical characteristics are shown in Table1. The subjects were asked to walk with bare feet at normal pace over an approximately 5-m walkway system. The variables, measured automatically by the system, are cadence, walking velocity and step length. Reproducibility of step length was also calculated.

RESULTS AND DISCUSSION: Cadence tends to decrease with age, but the rate of decrease is more rapid between age of 11 and 13yr. There are significant differences between twenties and 7,9,11 and 13yr. Walking velocity tends to increase rapidly in elementary school students, and gets gradually constant between age of 13 and 15yr. In the pattern of this trend, the significant change is found between 9 and 11. There are significant differences between twenties and age of 7 and 9yr. The velocity of the elder group are significantly small in comparison with that of twenties. Step length increases rapidly from age of 7 to 13yr and gets gradually constant between age of 13 and 15yr. In the pattern of this tendency, there are significant increases between age of 9 and 11yr and between age of 11 and 13yr. Step length of the old decreases significantly. The value of the reproducibility is one of the indications of walking stability, and it tends to decrease with growth. In the pattern of this tendency, there is significant decrease between 7 and 9yr and also significant decrease between 11 and 13yr. This shows that the stability of walking advances with maturation.

All the variables of twenties are nearly the same values as those of 15yr. While many excellent studies of gait maturation in children(2,3), we propose that the maturation of gait, as determined by our investigation on temporal and distance factors of walking, is 15yr because there are no significant changes between twenties and the age of 15yr. The gait characteristics of the old are short step length and slow walking velocity. The values of the

other variables are not significantly different from twenties.

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Table. Physical characteristics of sample population

Age group	7	9	11	13	15	20s	30s	Old
N	13	16	22	22	21	23	11	18
Age	7.2 (0.35)	9.2 (0.30)	11.3 (0.28)	13.2 (0.33)	15.2 (0.35)	23.8 (3.23)	36.2 (3.02)	75.7 (6.04)
Height(cm)	121.6 (6.2)	128.4 (6.5)	141.4 (5.1)	157.1 (10.3)	168.8 (5.4)	170.2 (5.2)	172.2 (5.7)	162.0 (5.2)
Weight(kg)	24.4 (3.7)	26.6 (3.8)	36.6 (5.8)	50.4 (12.8)	55.7 (8.0)	68.2 (13.3)	65.8 (8.8)	60.0 (6.4)

Values are mean $\pm$ SD.

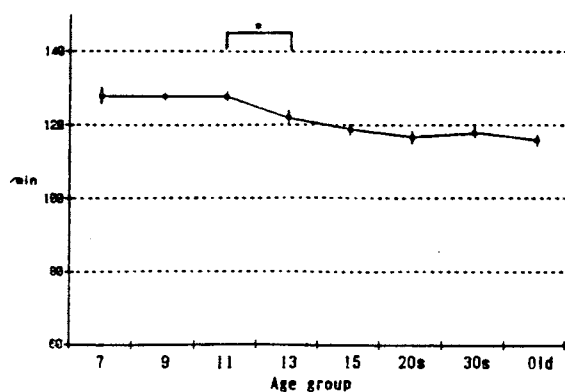


Fig 1: Cadence versus Age. ($p < 0.05$)

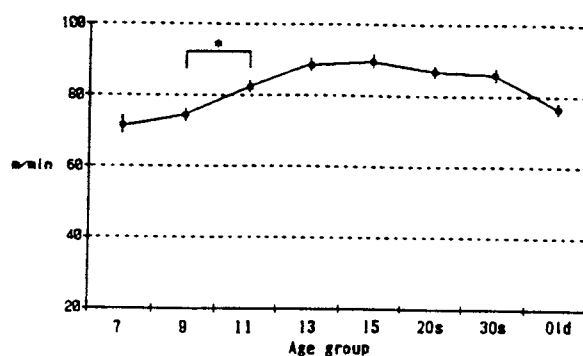


Fig 2: Velocity versus Age. ($p < 0.05$)

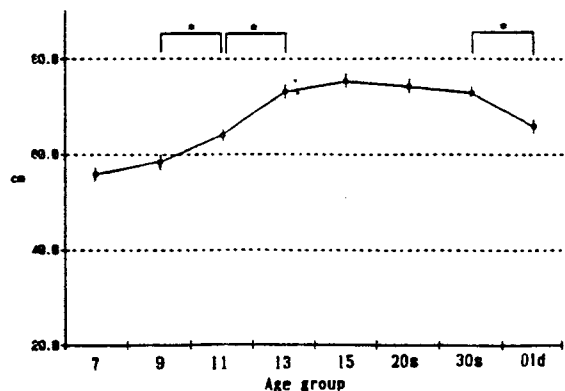


Fig 3: Step length versus Age. ($p < 0.05$)

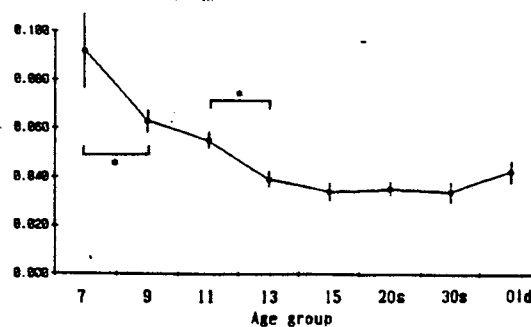


Fig 4: Reproducibility versus Age. ($p < 0.05$)

**CONTRIBUTION OF ANGULAR MOTION AND GRAVITY TO
TIBIAL ACCELERATION.**

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INTRODUCTION

Repetitive impact loadings have been shown to be linked to lower limbs fatigue injuries such as stress fractures (1) and osteoarthritis (2). Force platforms and accelerometers have been used by several authors to quantify the magnitude of the foot-ground impact during locomotor activities (3,4,5). Walking and running tibial acceleration has been recorded by means of skin mounted (4) and bone mounted (6,7) transducers. Independent of types and sites of attachment, it has been assumed that the transducers output reflected the acceleration resulting from ground contact (A_g). In the present study a bone mounted accelerometer was used in conjunction with high speed cinematography to evaluate the contribution of tibial angular motion (A_m) and gravity (A_g) to the axial component of tibial acceleration (TAA).

METHODS

Six symptom free volunteer subjects participated in the present study. A 6 gram Entran strain gage triaxial accelerometer was secured to a traction pin that was inserted into the right tibial. The 4.7 mm diameter traction pin was implanted 5-7 cm deep into the lateral aspect of the bone and 3 cm below the proximal articular surfaces. The subjects were required to walk and run five times across a Kistler force platform while being filmed from the side at 200 fps by a Photosonic high speed camera. The 3 components of ground reaction force and the 3 accelerometer signals were sampled simultaneously at 1 kHz by a DEC data acquisition system. A low pass digital filter (60Hz) was applied to the acceleration signals. The film coordinates of the accelerometer and those of the proximal and distal ends of the tibia were smoothed with a 15 Hz low pass digital filter.

The contribution of the tibial angular motion (A_m) to the tibial axial acceleration (TAA) was computed as follows:

$$A_m = d^2R/dt^2 - \omega^2 R \quad (1)$$

where R is the distance from the accelerometer to the hinge point and ω is the angular velocity of the tibia in the sagittal plane. Equation 2 shows how the contribution of gravity (A_g) to TAA was computed.

$$A_g = g \cos\theta \quad (2)$$

where g is the acceleration resulting from the gravitational force and θ is the angle between the tibia and the vertical.

RESULTS

Typical patterns of tibial axial acceleration (TAA) for one subject during walking and running are illustrated in Figures 1 and 2. Solid lines represent the total axial acceleration as measured by the bone mounted accelerometer and dashed lines represent the axial acceleration that result from ground contact impact (A_i). The bone mounted transducer underestimated A_i by 45 % (4.3g vs 7.9g) during running while it overestimated A_i by 16 % (2.9g vs 2.5g) during walking.

CONCLUSIONS

The results of the present study indicate that tibial angular motion and gravity contribute significantly to tibial axial acceleration during walking and running. Furthermore, the magnitude of the contribution of tibial angular motion to tibial axial acceleration is position dependent as revealed by the second term ($-\omega^2 R$) of Equation 1. Therefore, care should be taken in inferring ground impact magnitude from tibia attached accelerometer measurements.

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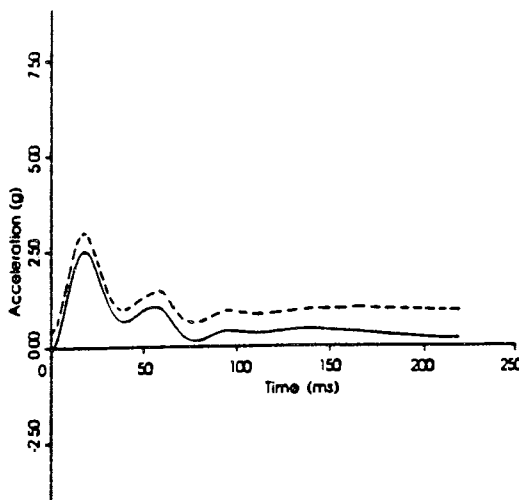


Fig 1. Walking tibial axial acceleration.

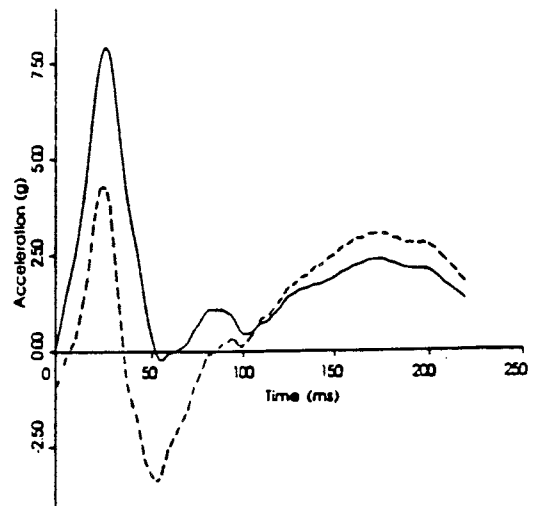


Fig 2. Running tibial axial acceleration.

TURNING OUT OF RESONANCE DURING WALKING AND REQUIREMENTS OF FOOTWEAR

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The concept of resonance is traditionally studied and used in biomechanics of locomotions. The resonance is useful for the minimization of energy expenditure by the inertial movement. It is undesirable by the unsufficient impact damping.

In this paper, we consider the new aspect of undesirable resonance, namely, the situation when it hampers the fixation of joints satisfying the movement programm. The fixation is realized, as a rule, by the muscles whose effort minimizes the energy expenditure of other muscles and favours thereby the minimization of total energy expenditures.

All this brings to the assumption that there exists the mechanism not only turning but also turning out of resonance by the locomotion control.

METHODS

Consider the bipedal walking of a polylinker whose dynamics is described by the system of differential equations

$$A\ddot{u} + B\dot{u} + Cu = f \quad (1)$$

where A, B, C are symmetric matrices of kinetic energy, dispersion and potential energy respectively and $f(t)$ is a controlled vector function of rigidity of elastic ties in hinges.

The stride can be subdivided into phases in accordance with the mobility in metatarsal, ankle and hip joints (Pitkin, 1975, Hof et al., 1987). In each of these phases a human body is representable by the system of one freedom degree under the mobility restriction in all but one hinges. In the limits of each configuration its natural frequency ω^* is resonancelly dangerous with respect to the conservation of its integrity.

The problem of walking control is replased by the problem of the control over the spectrum of an operator pencil corresponding to (1). The problem is solved variationally in terms of abstract Rayleigh systems (Abramov, 1983). The search for the solution $u = x \exp(\omega t)$ ($x \neq 0$) for the homogeneous equation corresponding to (1) results in the equation $\omega^2 Ax + \omega Bx + Cx = 0$ generating the pencil of matrices

$$L(\omega) = \omega^2 A + \omega B + C \quad (2)$$

The problem of control over the spectrum: given the Rayleigh system $R_\lambda(L_\lambda, P_\lambda)$ depending on the controlling parameter $\lambda \in U$ and the set $\Omega \ni \omega$; it is necessary to choose $\lambda \in U$ so that the spectrum $\sigma(R_\lambda)$ does not meet Ω .

In our statement, the Rayleigh system depends on the controlling parameter linearly and the spectrum of the pencil (2) breaks down into two sets of values of the Rayleigh functional (the condition of strong damping (Duffin, 1960))

$$p_{\pm}(x) = [-(Bx, x) \pm ((Bx, x)^2 - 4(Ax, x)(Cx, x))^{1/2}] / 2(Ax, x).$$

The solvability criterion for the problem of turning out of the resonantly dangerous domain $\Omega = [\beta, +\infty)$ (since the natural frequency of each configuration is real positive) has the form: $L(0) > 0$ on the set $V = \{x_i\}$ which gives after the substitution of zero in the equation (2):

$$(C_i x, x) \geq 0 \quad (i - \text{are the indices for which } \lambda_i \geq 0) \quad (3)$$

$$(C_i x, x) = 0 \quad (i - \text{are the indices for which } \lambda_i < 0)$$

If $\Omega_+ = [\beta, +\infty)$, $\Omega_- = (-\infty, \beta)$ and $q_+(\lambda)$ ($q_-(\lambda)$) is the maximal (minimal) point of the spectrum of L_λ , then the distance ρ from q_+ to Ω_+ is given by the formula

$$\rho = \sup_{\Lambda} \text{dist} \{ \sigma(R_\lambda), \Omega_+ \} = \inf_{x \in V} |p(x) - \beta| \quad (4)$$

where $\Lambda = \{\lambda\}$.

As we consider $\Omega = \Omega_+$ the formula (4) is transformed in the following way: $\rho = \inf |p(x) - \beta|$ and since $\beta = 0$, finally

$$\rho = \min_{V_+} [-(Bx, x) - ((Bx, x)^2 - 4(Ax, x)(Cx, x))^{1/2}] / 2(Ax, x). \quad (5)$$

RESULTS

The substitution of entries of matrices , , in (4) and (5) allows to obtain the unknown values of the components of piecewise constant controlling vector function on each phase of the stride.

Experimental shoes and insoles with calculated viscoelastic properties are presented (Pitkin et al., 1986, 1987), (Mendelewich, Pitkin, 1987).

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ISOKINETIC TORQUE EXPRESSED RELATIVE TO KNEE VERSUS LEVER ARM ANGLE

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INTRODUCTION

Muscle strength measured with isokinetic dynamometers is typically expressed as maximal torque irrespective of angle or as angle specific peak torque (Gregor *et al.* 1979). Functional strength of the knee extensors and flexors is routinely assessed by measuring the hamstring (HAM)/quadriceps (QUAD) torque ratio in clinical settings. Because the HAM and QUAD muscles oppose each other at any given knee angle during locomotion it has been suggested that this ratio be measured specific to a given angle (Figoni *et al.* 1988). Moreover, it is believed that more valid measures are obtained when a lever arm angle relative to horizontal is used rather than an angle expressed relative to an anatomical reference (Wickiewicz *et al.* 1984). However, soft tissues are compressible and the alignment of the rotational centers of the knee and the dynamometer are displaced during loading (Nosse 1982). Therefore, it is unlikely that the lever arm angle of the dynamometer reflects the knee joint angle. Accordingly, this study compared torque during QUAD and HAM muscle actions when values were expressed relative to a given angle determined by the dynamometer's goniometer versus actual knee joint angle. Knee joint angle was determined with an external goniometer.

METHODS

Physically active males ($n=6$) with asymptomatic knee function volunteered. Their mean (SD) age, height and weight were 28(6) yrs, 180(6) cm and 72(2) kg. The subjects visited the laboratory on two different occasions. The first was intended to familiarize the subject with the protocol and the dynamometer (for a detailed description see Tesch *et al.* 1989) utilized in the study. The subjects were positioned in a chair providing support for the thigh, pelvis, back and shoulders. The axis of rotation of the knee joint was aligned with the center of rotation of the dynamometer. An external goniometer ($GONIO_{ext}$) was aligned with the rotational center of the knee, the trochanter major of the femur and the malleolus lateralis. The goniometer of the dynamometer ($GONIO_{int}$) and $GONIO_{ext}$ were calibrated at 90° of flexion, (assessed with a set square) and at 30° of flexion using a spirit level and protractor with a 1° resolution. After a 10 min warm-up on a cycle ergometer followed by stretching of the thigh muscles, QUAD and HAM torques were measured in the following order: QUAD concentric (CON), HAM CON, QUAD eccentric (ECC) and HAM ECC. CON and ECC muscle actions were performed at constant angular velocities in this order: 0.52, 1.57 and 2.61 rad/s (30, 90 and $150^\circ/s$). Torque and angle for both goniometers were recorded on a strip-chart recorder (Mingograf 804, Siemens Elema). Mean and standard deviation (SD) were calculated and potential intra- and inter-group differences were tested using analysis of variance (ANOVA). The level of significance was set at $p<0.05$.

RESULTS

The torque-angle relationships for QUAD CON and HAM CON muscle actions at 0.52 rad/s are shown in Figs. 1 and 2. Peak torque during QUAD muscle actions occurred at greater ($p<0.05$) angles for $GONIO_{int}$ versus $GONIO_{ext}$. For HAM muscle actions peak torque occurred at lesser ($p<0.05$) angles for $GONIO_{int}$ versus $GONIO_{ext}$. Therefore, the HAM/QUAD torque ratio at a given angle measured deviated substantially ($p<0.05$) for $GONIO_{int}$ versus $GONIO_{ext}$ at most angles. This was true for both CON and ECC muscle actions irrespective of angular velocity. The mean (SD) HAM/QUAD torque ratio at different angles during CON muscle actions at 0.52 rad/s is depicted in Table 1.

DISCUSSION

This study showed that angle specific peak torque expressed relative to $GONIO_{int}$ was different from that relative to $GONIO_{ext}$. QUAD peak torque occurred closer to full knee extension and HAM peak torque occurred closer to 90° of knee flexion when assessed with $GONIO_{int}$ compared with $GONIO_{ext}$. As a result, the HAM/QUAD torque ratio decreased as $GONIO_{int}$ angle increased (i.e., as full flexion was approached). This ratio did not change as $GONIO_{ext}$ angle increased. Thus, it appears that the torque of HAM and QUAD is not measured at the intended joint angle if a dynamometer goniometer measuring lever arm angle is used.

The error introduced is probably due to soft tissue compression. This source of error should be considered when comparing individuals possessing large differences in muscle mass and fat tissue of the limb. It is also likely that changes in soft tissue mass following interventions such as inactivity or training, per se, will influence angle specific torque if related to lever arm angle rather than knee angle.

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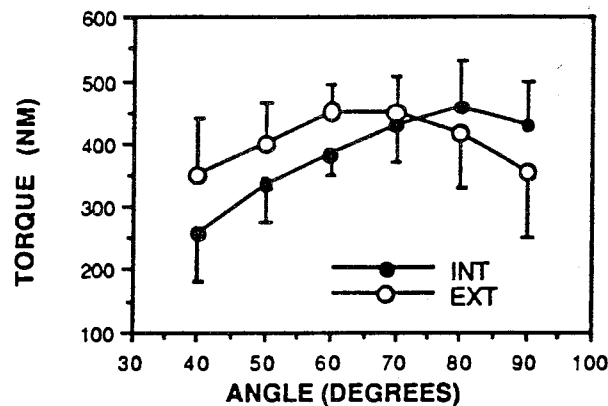


Fig. 1: The torque-angle relationship for QUAD CON muscle actions at 0.52 rad/s. Vertical bars indicate plus or minus the standard deviations.

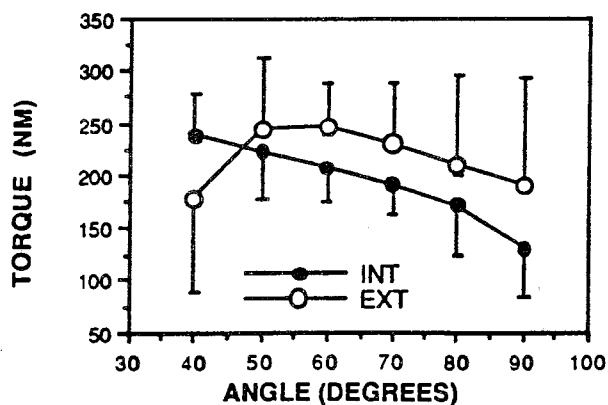


Fig. 2: The torque-angle relationship for HAM CON muscle actions at 0.52 rad/s. Vertical bars indicate plus or minus the standard deviations.

Table 1. The HAM/QUAD torque ratio at specific angles measured by GONIO_{int} and GONIO_{ext}. Values are mean (SD). * = significant difference; ns = non-significant difference.

Angle	GONIO _{int}	GONIO _{ext}	
40	1.05 (0.17)	0.54 (0.21)	*
50	0.73 (0.20)	0.62 (0.09)	ns
60	0.54 (0.10)	0.56 (0.07)	ns
70	0.44 (0.08)	0.53 (0.10)	*
80	0.36 (0.06)	0.52 (0.12)	*
90	0.31 (0.10)	0.60 (0.30)	*

THE ROTATION CENTER OF THE GLENOHUMERAL JOINT.

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Mekelweg 2, 2628 CD Delft, The Netherlands.^{*}: Faculty of Human Movement Studies, Free University, Amsterdam.Introduction and method.

The location of the rotation center of the humeral head is of great importance to gain insight in the functioning of the glenohumeral joint. On the one hand there are clinical implications: Considering the dimensions of the glenoid and the loose capsule, the glenohumeral joint is an unstable joint. Therefore, muscle action is needed to prevent the joint from dislocation. On the other hand, in developing a musculoskeletal model of the glenohumeral joint the location of the rotation center has to be known to calculate the moment arms of muscles crossing the joint.

The location of the rotation center of the humeral head must be derived with respect to the scapula. Thus far, only 2D-motion studies of the humerus and the scapula have been used to determine the rotation center. Observed translations of the rotation center then are likely to have been caused by motions of the scapula. Another method is to calculate the rotation center from the shape of the articular surfaces. For that, 2D-roentgenograms and consecutive transversal cross-sections like CT-scans have been used, but these methods are not accurate. Saha (1958) used a spherometer to measure directly the radius of the surfaces, but he was not able to determine the rotation center.

In this study the geometry of the shoulder mechanism, including thorax, clavicle, scapula and humerus, is recorded. Seven cadavers have been measured, totalling 14 shoulders. The 3D-positions of morphological structures as articular surfaces, muscle origins and insertions, ligament attachments and muscle paths, are recorded with the palpator and expressed in a coordinate system of which the axes are parallel to the anatomical axes (x: medial-lateral, y: caudal-cranial, z: ventral-dorsal). The measurement error of the palpator has a standard deviation of 0.93 mm per coordinate or 1.49 mm over all coordinates (Pronk & van der Helm, 1988).

In this paper the results will be restricted to the articular surfaces of the glenohumeral joint. A large number of datapoints is recorded on each joint surface (humeral head: range 35 to 71; glenoid: range 25 to 49). Due to the near resemblance we have chosen to model the articular surface as a sphere. The center M (Mx, My, Mz) and radius r of the sphere are estimated by minimizing the distance e_i of the datapoints (x_i, y_i, z_i) to the surface of the sphere, using a least squares criterion:

$$e_i = \sqrt{(x_i - Mx)^2 + (y_i - My)^2 + (z_i - Mz)^2} - r$$

And the criterion to be minimized:

$$V(x_i, y_i, z_i, M, r) = \sum_{i=1}^N e_i^T e_i$$

An iterative Gauss-Newton method is used to obtain a solution.

Results and discussion.

Table I shows the accuracy of the method. Comparing the mean residual error $\bar{e}$ to the measurement error of the palpator, the accuracy of the fit is

amazing. This is an indication that the articular surfaces of the glenohumeral joint can be modelled as spheres. In addition, the standard deviations indicate that the obtained parameters are very precise. Together with the results of Saha (1958), the results of this experiment are presented in table II. Compared with the humeral head, the $\overline{sdr}$ of the glenoid (table I) is relatively large. This can be explained by the small surface of the glenoid by which the datapoints do not provide enough information to estimate the radius accurately. Therefore, the estimation of the center of the glenoid is repeated with a constant radius equal to the radius of the humeral head. The results of these calculations are shown in table III. The increase of $\bar{e}$ (glenoid) is small (0.06 mm.), while $\overline{dm}$ decreases. These results indicate that the joint surfaces of the glenohumeral joint can be assumed to be two concentric spheres. Thus, we assume that the rotation center of the humeral head is the center of the sphere fitted to the glenoid, for if this is not the case this would result in an unstable joint.

The resultant muscle force vector will compress the joint surfaces onto each other. The line through this force vector passes through the rotation center of the humerus. If the direction of this force vector points outside the glenoid, the reaction force of the glenoid can not fully counteract this force. Hence, a translation of the humeral head will follow. This results in an unstable joint, because the rotation center will shift towards the edge of the glenoid. This is not a preferable situation. It is likely that the function of the rotator cuff is to control the direction of the resultant muscle force vector so that it points inside the glenoid. The insertions of the rotator cuff as a halfcircle around the humeral head suit this purpose very well. If these assumptions are correct, it is very unlikely that a translation of the rotation center with respect to the scapula will occur during the range of motion of the humerus.

Saha, A.K. (1958). 'Zero-position' of the glenohumeral joint. Its recognition and clinical importance. Ann. roy. Coll. Surg. Engl. 22, 223-236.
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Table I: The accuracy of the method used to derive the shape of the glenohumeral joint (in mm., N=14).

$\bar{e}$: mean residual error of the fit of the datapoints to the sphere.
 $\overline{sdM}$: mean standard deviation for the estimation of the center M.
 $\overline{sdr}$: mean standard deviation for the estimation of the radius r.

	$\bar{e}$	$\overline{sdMx}$	$\overline{sdMy}$	$\overline{sdMz}$	$\overline{sdr}$
humeral head	0.71	0.11	0.24	0.16	0.17
glenoid	0.26	0.67	0.18	1.00	1.47

Table II: radius r (in mm.) as obtained in this study, averaged over 14 shoulders, compared to Saha (1958), standard deviations in parentheses.

	This study		Saha (1958)	
	$\bar{r}$	range	young adult	old person
humeral head	25.7 (2.4)	22.0 - 31.0	± 22	± 23
glenoid	28.3 (4.2)	21.6 - 36.9	± 22	± 21

Table III: mean distance $\overline{dm}$ between estimated center M of the humeral head and the glenoid and mean absolute difference $\overline{dr}$ between the radius of the humeral head (r_{hum}) and the glenoid (r_{gl}), standard deviations in parentheses.

$\overline{dm}$	$\overline{dr}$	$r_{gl} - r_{hum} (\overline{dr} = 0)$	
		$\overline{dm}$	$\bar{e}$ (glenoid)
6.2 (4.0)	3.2 (3.1)	4.9 (2.8)	0.32

STANDING RADIOGRAPHIC MEASUREMENTS VS FORCE PLATE GAIT ANALYSIS OF MODERATE GONARTHROSIS

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In surgical treatment of gonarthrosis the operation is planned from a static X-ray of the patient standing. In the present study radiographic measurements of angular deformity of the knee in the frontal plane during standing were compared to analysis of adduction moments during walking obtained by a Kistler force plate and a video recording system. Twenty patients with moderate unilateral medial gonarthrosis were investigated. The mean Hip-Knee-Ankle (HKA) angle measured on the radiographs was 11 degrees of varus and the mechanical axis between ankle and hip was located medial to the medial tibial plateau in seven patients. The mean mid-stance adduction moment normalized to height and weight was 0.33. No significant correlation between magnitude of HKA angle and mid-stance adduction moment was found.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

MATHEMATICAL MODEL OF KNEE JOINT EXTENSOR MUSCLES FATIGUE

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Introduction

Human muscles fatigue has been defined in this study as the failure to maintain the required or expected force [1]. Sustained contraction of a muscular group induces changes of maximum force, torque, power etc. [2,3,4,5], however after rest-time the recovery can be observed [5]. The purpose of this study was to simulate knee joint extensor muscles fatigue with a mathematical model.

Model

Movement of a shank from vertical to horizontal position for sitting subject and a phenomenon of extensor muscles fatigue were described by the following set of equations :

$$T\ddot{M} + 2T\dot{M} + M = K_w(\alpha)U(t) \quad (1)$$

$$T_z\dot{y} + y = U(t) \quad (2)$$

$$M_f = M(1 - y) \quad (3)$$

$$M_f + M_z + M_H + M_g = J\ddot{\alpha} \quad (4)$$

$$M_H = k(\alpha)\alpha + c(\alpha)\dot{\alpha} \quad (5)$$

$$M_g = mgr \sin(\alpha) \quad (6)$$

$$M_z = f(M_f, \alpha) \quad (7)$$

where M is the muscular torque in knee joint without fatigue, M_f is the muscular torque with fatigue, $U(t)$ is the excitation parameter, y is the variable of fatigue and recovery, M_z is the external torque, M_H is the braking torque, M_g is the gravitation torque, α is the angle between a shank and vertical line. T, T_z, J, m, r are constant parameters, $k(\alpha)$ is the coefficient of elasticity, $c(\alpha)$ is damping coefficient and $K_w(\alpha)$ is the maximum static muscular torque.

Simulation

The set of equations has been solved using Runge-Kutta-England method of numerical integration. Two different examples of simulation results are shown in figures 1 and 2.

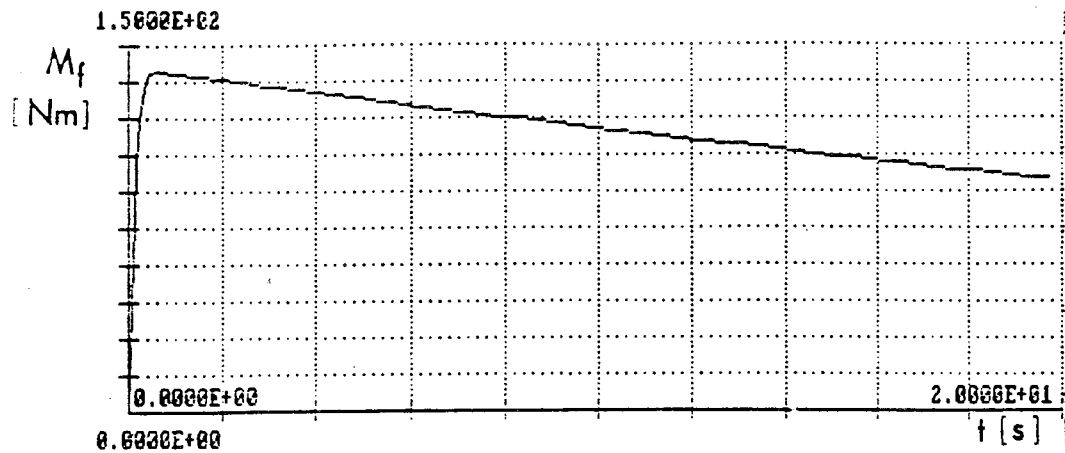


Figure 1. Relation between torque M_f and time during 20s of maximum voluntary contraction in static conditions.

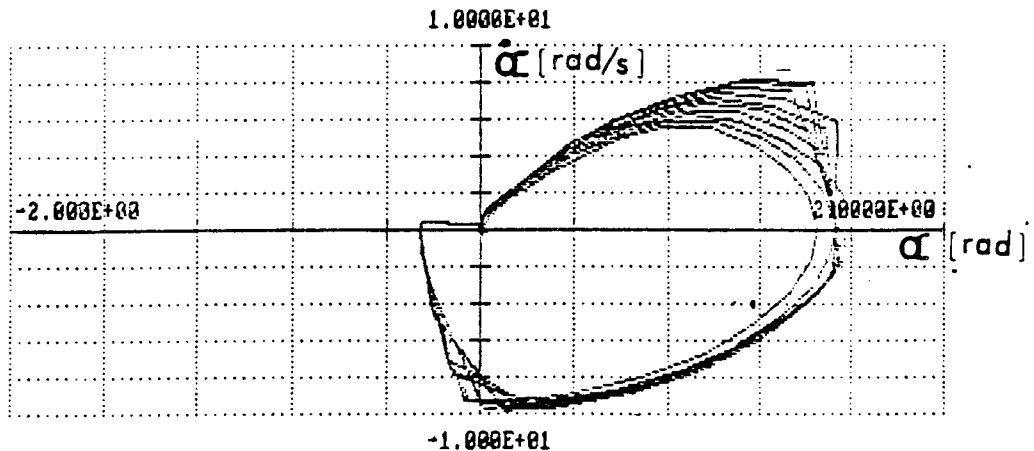


Figure 2. Phase trajectory $(\alpha, \dot{\alpha})$ for 10 maximum voluntary contractions performed during 2s period.

This model was verified for right knee joint extensor muscles in static and dynamic conditions.

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A MECHANICAL MODEL TO SIMULATE AND PREDICT HUMAN WALKING PATTERNS

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INTRODUCTION

The provision of a prosthesis to a leg amputee and the rehabilitation procedures are very expensive and do not always lead to an on long-term successful use of the prosthesis. The possibility of calculating in advance the functional results of a prosthetic provision would lead to a fitting optimally adapted to the amputee, resulting in a shorter rehabilitation period and an increased possibility of success. Most of the present mechanical models to simulate human walking are not adequate for this purpose since they can only be used for analyzing measured movements and not for predicting movements. Therefore we developed a model that does have these potentialities.

METHODS

The basis of the model is formed by a 7-segmental three-dimensional rigid body model, with segments for the upper legs, lower legs, feet and pelvis including head, arms and trunk. Anatomical data are used to define the mechanical properties of the segments, using the regression relations of Chandler et al.¹ To define the muscles in the segments, the muscle attachment model of Brand et al.² is applied. With the constraint that during stance the foot must be on the floor and no slipping between foot and floor may occur, the forward movement of the whole body can be calculated from the input joint rotations only (fig. 1). For any arbitrary set of input joint rotations of the model, the forces and moments of force needed to bring about that movement are calculated. These include the ground reaction forces. The predictive capacity of the model is obtained by applying optimization routines to the model. The variables that are optimized can be the step-time parameters and the variables describing the joint rotations. As a result of the analysis, the value of an objective function is calculated. This objective function will be minimized during optimization (fig. 2).

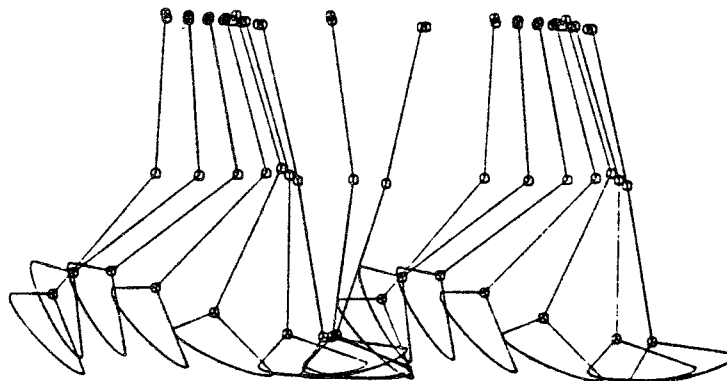


Figure 1: Forward movement of the body, only one leg is shown. Note the roll-over of the foot.

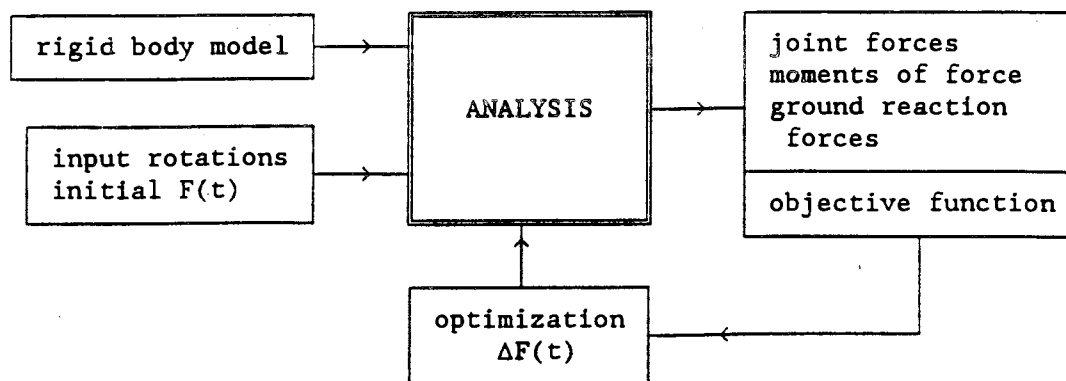


Figure 2: Block diagram of the model.

RESULTS

The model is applied to normal level walking. For the input rotations some measured knee, hip and ankle rotations in the sagittal plane are chosen. From the constraints that the body must be in balance, the pelvic rotations and the hip adduction are calculated. With an objective function describing the energy consumption during walking, step-time parameters such as the step frequency and step length can be predicted as a function of the forward velocity. From literature it is known that the energy consumption should be a quadratic function of the velocity. In figure 3, the calculated energies are plotted with the best fitting curve of the form $E = A + B v^2$ (with $r > 0.99$).

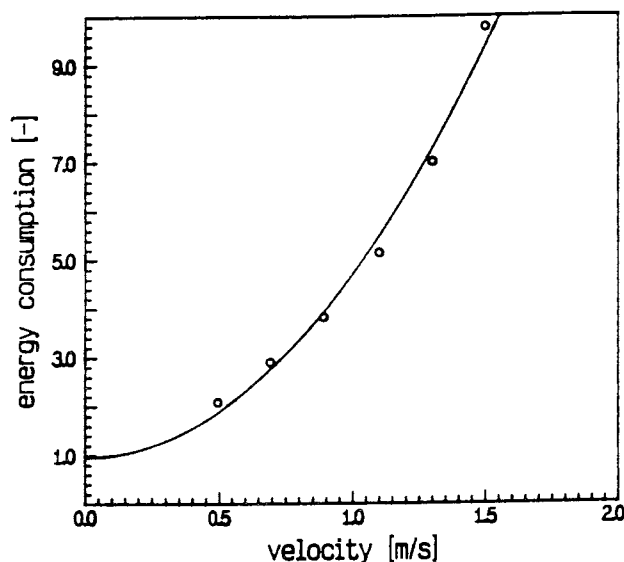


Figure 3: Predicted energy consumption (arbitrary units) as a function of the forward velocity.

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Segmental Energy Contributions in Walking and Running

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Introduction

Segmental mechanical energy analysis techniques have been used to describe and understand various aspects of human locomotion. Previous studies have described the energetic utilization of individual limbs or segments in both walking (e.g. Bresler and Berry, 1951) and running (e.g. Chapman and Caldwell, 1983). A topic which has received little attention is the manner in which the individual segmental energy patterns change as the speed of locomotion is altered. What are the specific roles of these segmental patterns in walking and running at different velocities? Are the segmental energetics specific to the gait mode (ie: walking or running), or are they related more closely to progression velocity?

The purpose of this study is to examine segmental energy patterns in the trunk and leg segments in both walking and running, and to demonstrate how these energetics change as progression velocity is altered.

Methods

Three subjects were filmed at a nominal rate of 65 frames per second while locomoting over level ground. Reflective markers attached to the fifth metatarsal, lateral malleolus, knee joint centre, greater trochanter and shoulder were digitized and used as endpoints to form a four segment model consisting of the foot, shank, thigh and trunk. One complete stride was analysed for each of a variety of locomotion velocities ranging from a slow walk (about 0.5 m/s) to a medium speed run (5.5 m/s). Marker displacement data were smoothed using a dual pass, second order Butterworth digital filter. Frequency cutoffs were determined individually for each marker (3 to 8 Hz). Velocity - time data were calculated using finite differences from the smoothed displacement records.

Translational kinetic (KE), rotational kinetic (RE), potential (PE) and total segmental energy (SE) time histories were calculated for each segment. These energy variables were used to calculate the work rate of each segment and the total model using different assumptions regarding segmental energy transfers (Winter, 1979; Norman, Caldwell and Komi, 1985). The resulting work values W_y (no transfers), W_{yb} (transfers within segments) and W_{yb} (transfers between and within segments) were used to calculate energy transfers within (ET_y) and between (ET_b) segments as a percentage of W_y . Individual transfers within each segment were expressed as a percentage of the work output of that specific segment. Transfers were also calculated between individual segments.

Results and Discussion

As would be expected, W_y , W_{yb} and W_{yb} all increased with velocity of progression. While the magnitude of total energy transfer (ET_{yb}) showed no

clear trend with velocity, ET_v and ET_h magnitudes are offsetting functions of velocity (Figure 1: one subject). It is unclear if these differences are strictly velocity related or reflect basic differences in energy transfer mechanisms between walking and running. The individual transfers within and between each segment provide more detailed explanation of the total model results. For example, the trunk (ET_{tr}) and thigh (ET_{th}) within segment transfers show a definite trend toward lower values with increased velocity as does the total model ET_v , but it is unclear whether differences exist between walking and running. However, Figure 2 depicts distinct differences between the two gait modes in the energy transfers within the shank (ET_s) and foot (ET_f). Within each gait mode ET_s and ET_f decrease with increasing velocity, but in the change from walking to running the transfer values are suddenly elevated, forming a clear bifurcation point. These results are discussed with regard to basic energetic mechanisms in walking and running.

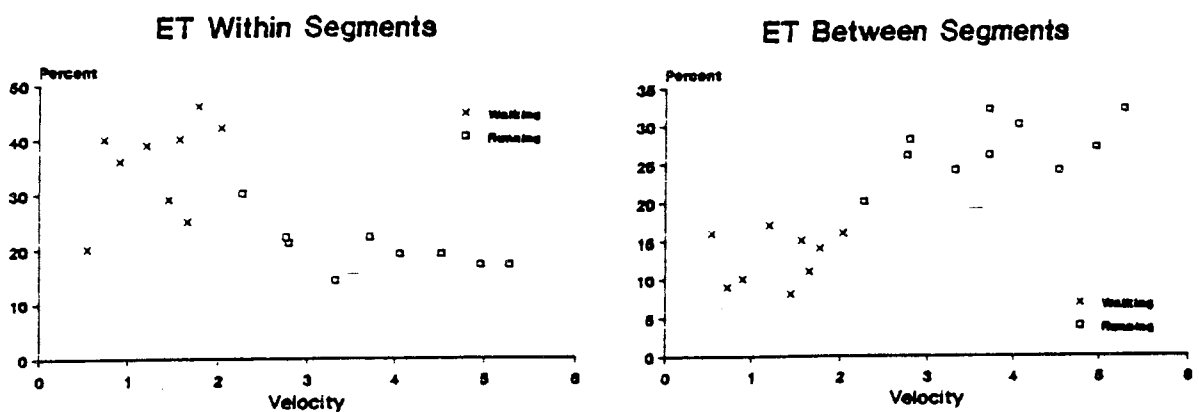


Figure 1: Energy transfers within (ET_v) and between (ET_h) segments

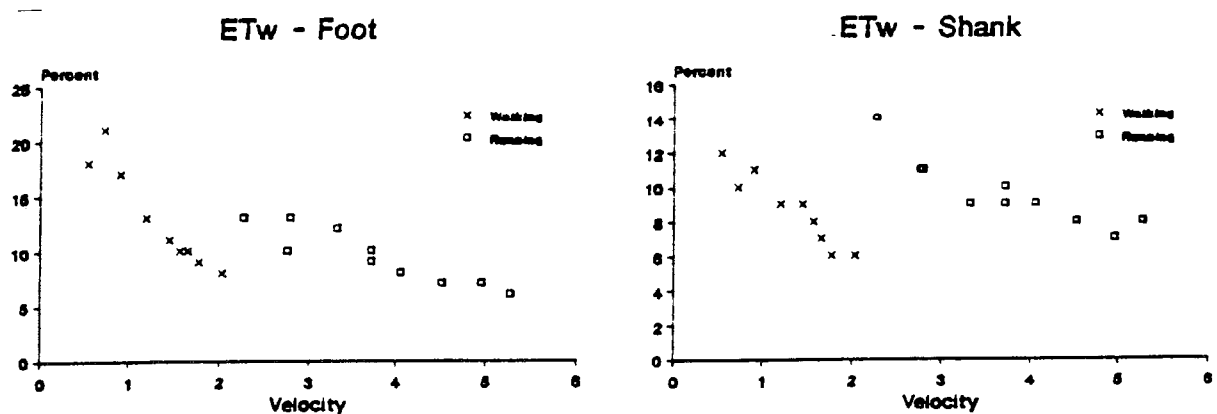


Figure 2: Energy transfers within the foot (ET_f) and shank (ET_s)

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CONTROL OF RAPID MOVEMENT ONSET UNDER CONDITIONS OF VARYING RESISTANCE

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Previous investigations have examined neuromuscular aspects of rapid movement initiation with conflicting results. While all have agreed that the onset, amplitude and duration of activity in agonist and antagonist muscles are important determinants of limb movement, the interaction of these parameters with mechanical aspects of the task in determining movement kinematics remains unclear (Haycraft, 1898; Berger & Mathus, 1969). Resistance to movement and variations in load are two factors which should influence rapid movements, yet there are conflicting and incomplete experimental data for their actual effects. For example, investigators have disagreed over the extent to which electromyographic (EMG) patterns in antagonist muscles are determined by central or proprioceptive influences (Angel, 1975, 1977; Jacobs et al, 1980; Maton et al, 1980; Berardelli et al, 1984). The fundamental issue appears to be how centrally-generated activation patterns may be altered under different mechanical conditions, either by planning or proprioceptive feedback. This investigation examined relationships between the activation pattern of movement-related antagonists and known mechanical constraints of load and resistance to movement, as reflected in the neuromuscular activity and the movement dynamics.

Seven normal adult female volunteers without prior training were tested, after providing informed consent. Subjects were seated and stabilized; the right arm was attached to a lever instrumented to record movement dynamics. The experimental task required the subject to respond to a visual cue with an immediate maximum velocity elbow flexion through a range of 90 degrees, starting at an angle of 135 degrees and stopping at an angle of 45 degrees. A variable strength magnet was used to provide resistance to initial movement at 0, 40, or 70 percent of the subject's maximum isometric flexion torque. Weights were added to the lever to double or triple the moment of inertia of the forearm and lever system. After one day of initial testing and familiarization, data were collected from each subject on three different days: on each day the load (moment of inertia) was constant (X 1, 2, or 3) and the three initial resistance (magnet) values were tested in blocks of ten trials each. The order of treatments was balanced across subjects. Short rest periods between trials (10-15 s) and longer rest periods between blocks (5 min) prevented fatigue effects. Analog records of each performance were collected on a laboratory microcomputer and stored on FM tape. Surface electromyographic (EMG) recordings were obtained from bipolar electrodes over the biceps brachii and triceps brachii. Torque exerted at the elbow prior to movement onset was obtained from a strain gauge in series with the resisting magnet. Elbow angle (position) was transduced by a linear potentiometer attached to the lever axle, and angular velocity and acceleration were electronically derived and stored also. The EMG records were rectified and low-pass filtered, and critical points were selected from the analog records using a microcomputer-based digital oscilloscope.

The data were analyzed using ANOVA to determine whether differences existed between different loads or different resistance for several kinematic and EMG variables. Increasing the load from 1 to 2 to 3 times the moment of inertia resulted in consistent increases in reaction time (RT), movement time to 45 and to 90 degrees (MT45, MT90) and total movement time (MTTOT), while the final angle actually achieved did not vary. These movement time increases were accompanied by increases in time to peak velocity (VELTIME), decreases in peak velocity (VELPK), no change in time to peak acceleration (ACCTIME), decreases in peak acceleration (ACCPK), and increases in the duration of the flexion acceleration (ACCDUR). That is, increased loads yielded slower movements. The initial burst of biceps brachii activity (B1) showed some corresponding differences, as the time to peak (B1PK) and burst duration (B1DUR) steadily increased, the initial rate of EMG increase (B1SLOPE) was lower with the heaviest load, and the premotor latency (B1PMT) increased with the heaviest load. The triceps burst had a greater duration and greater time

to peak in the loaded conditions. Increasing the resistance to movement initiation from 0 to 40 to 70 percent of the subjects' maximum torque yielded different results. As with increased loads, increases in resistance yielded increasing RT and no change in final angle achieved. However MT45 steadily decreased, and MT90 and MTTOT decreased significantly at the highest resistance. These movement time decreases were accompanied by decreases in VELTIME, ACCTIME, and ACCDUR, and an increase in ACCPK. Thus the result of increasing resistance to movement initiation was delayed onset but faster movements. No significant differences related to level of resistance were found in the first bursts of biceps brachii or triceps brachii EMG activity.

These data suggest that this rapid movement initiation was affected by different mechanical constraints in different ways. Under conditions of varying load which remained in effect throughout the movement (i.e. moment of inertia), the pattern of muscle activation was altered. However, when resistance only to initial movement was varied, the muscle activation pattern did not change, and changes in the movement pattern could be attributable to elastic properties of muscle.

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AN ANTHROKINETOMETRIC STUDY OF MALE MONTREAL OLYMPIC ATHLETES
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A large number of Kinanthropometric studies have been undertaken on Olympic athletes in order to study the variations in body size, shape and composition of different categories of athletes. However, the landmarks and measurements of body segments constituting various body levers are yet to be properly defined. In the absence of any such data on Olympic athletes, an attempt has been made in the present study, to indirectly predict some anthropometric variables of Olympic athletes from the anthropometric measurements to find the variations in these parameters of different categories of athletes.

Subjects and Methods: Anthropometry of 309 male Olympic athletes reported by Carter (1982) has been used to predict the upper arm and fore arm links according to Dempster (1955). Distances of centers of masses of arm segments from their respective axes of rotation at shoulder, elbow and wrist joints have been computed by applying Clauser et al (1969) formulations. Segment volumes have been computed based on the cylindrical model of arm segments. The mechanical advantage is given by the ratio of effort force arm to resistance force arm of segmental limb levers. As the muscles are inserted on arm bones quite close to shoulder and elbow joints, the length of effort force arm has been taken as a constant distance of five cm (Le Veau, 1977). The ratio of segment volume to resistance force arm has been called as the 'Effort Force Factor (EFF)' while the product of EFF and mechanical advantage gives the Resultant E.F.F. Similarly the product of segment volume to resistance lever arm has been considered as 'Resistance force factor of the respective joint.'

Results: Anthrokinetometric variables of male Montreal Olympic athletes as computed from the average values of their anthropometric variables have been enlisted in Table 1. It is interesting to note that the values of speed enhancing parameters i.e. longer lever arms, lesser mechanical advantage, lesser segmental volume per centimeter are in general favourable in athletes performing timed athletic events like Rowing, Canoeing, Cycling etc. While the athletes performing strength activities like weightlifting, wrestling have shown the smaller resistance force arms without loosing the bulk of muscles. This adaptation suggests that in case of longer lever arms, the arrangement of sarcomere units in the muscles adjust more in linear fashion i.e. sidewise while in case of short r lever arms, the same number of sarcomeres adjust their arrangements to a greater extent in the transverse direction by lying one above the other. In both the cases the contraction of constant number of sarcomeres will result in the generation of same muscular pull which in the first case will be used to move comparatively lesser resistance to a greater distance while in the second case a greater amount of resistance force will be carried to a lesser distance. In other words mechanical advantage lost in strength is equivalent to gain in speed or distance moved. These principles of kinetics are well in agreement to the observations of the present study. For instance the lengths of arm links (26.3 cm in upper arm and 24.5 cm in fore arm) are smallest in weight-lifters who have the maximum mechanical advantage (0.5236 at elbow and 0.3444 shoulder axis). Contrarily, Rowers having longer link lengths and lesser mechanical advantage,

can move at a greater speed. It is interesting to note that the volumes of upper arm plus fore arm segments of weightlifters and rowers are quite comparable, 4760.1 cm³ and 4730.4 cm³ respectively. It is suggested that a new biomechanical specialisation which may be termed as Anthrokinetometry be developed so as to work out internationally accepted definition of landmarks and body measurement techniques of movable body segmental units.

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Table-1: Volumes, Centres of Mass (CM), Mechanical Advantages (MA) Lengths of Arm Links (LAL), Resistance Force Factor (RFF), Proportionate Effort Force Factor (EFF) Resultant E.F.F. (R.E.E.F.) of Upper Arm (UA) and Fore Arm (FA) for male Montreal Olympic athletes. S=Shoulder, E=Elbow, W=Wrist.

Sport	Volume cm ³	CM from		MA**	LAL cm	RFF		EFF	REFF
		S/W* cm	E cm			S/E	E/W		
Average	UA 2264.4	15.8	15.0	0.3164	30.8	35.77	33.94	143.3	45.34
Athlete	FA 1724.6	16.6	10.8	0.4643	27.6	19.22	30.05	165.7	76.93
Boxing	UA 1866.1	15.0	14.3	0.3329	29.3	28.03	26.61	124.2	41.36
	FA 1588.8	16.0	10.2	0.4692	26.2	16.24	25.41	155.5	76.50
Cycling	UA 1850.7	15.6	14.9	0.3193	30.5	28.98	27.52	118.2	37.73
	FA 1507.2	16.8	10.7	0.4664	27.5	17.23	26.97	149.9	69.93
Track & Field	UA 2057.3	15.7	14.9	0.3192	30.6	32.22	30.59	131.4	48.93
	FA 1730.1	17.0	10.9	0.4591	27.9	18.64	29.48	158.9	72.94
Fencing	UA 2138.0	16.2	15.3	0.3094	31.5	34.55	32.62	132.3	40.93
	FA 1769.3	17.3	11.1	0.4521	28.4	19.57	30.61	160.0	72.32
Hockey	UA 1917.1	15.5	14.6	0.3236	30.1	29.62	28.08	124.1	40.92
	FA 1565.5	16.7	10.7	0.4682	27.4	16.72	26.16	146.6	68.62
Canoeing	UA 2636.2	16.4	15.5	0.3051	31.9	43.21	41.02	160.6	49.07
	FA 1908.1	17.0	10.9	0.4591	27.9	20.78	32.51	175.2	80.45
Gymnastics	UA 2163.0	14.8	14.0	0.3361	28.8	31.99	30.39	146.2	49.45
	FA 1609.8	16.3	10.4	0.4794	26.7	16.79	26.27	154.3	73.99
Swimming	UA 2288.3	15.7	15.0	0.3175	30.7	36.04	34.23	145.3	46.13
	FA 1642.9	16.8	10.7	0.4664	27.5	17.61	27.57	153.3	71.49
Rowing	UA 2311.7	16.7	15.9	0.2985	32.6	43.75	41.53	155.9	46.54
	FA 2118.7	17.7	11.3	0.4821	29.0	22.26	37.48	187.3	82.80
Judo	UA 2506.8	15.1	14.4	0.3309	29.5	37.38	35.97	165.9	54.90
	FA 1740.0	16.9	10.1	0.4931	26.0	17.64	27.60	171.6	84.62
Wrestling	UA 2675.4	15.3	14.6	0.3259	29.9	39.31	37.50	167.9	54.72
	FA 1786.0	15.7	10.1	0.4970	25.8	17.97	22.00	177.5	83.24
Weightlifting	UA 2918.6	14.5	13.8	0.3444	28.3	42.38	40.22	201.0	69.23
	FA 1861.5	15.0	9.5	0.5236	24.5	17.78	27.33	194.9	102.06

* In case of UA from shoulder joint axis & in case of FA, from wrist joint axis.
 ** In case of UA, at shoulder joint & in case of FA, at elbow joint.

THE INFLUENCE OF SHOEING ON TENDON STRAIN IN THE EQUINE FORELEG

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INTRODUCTION

It is common practice to change the hoof shape of horses in order to change limb loading patterns or gait [1,2]. The present study was undertaken to investigate the influence of different hoof angles and soil conditions on the load distribution over the tendons in the equine foreleg. In five ponies tendon strain, ground reaction forces and kinematics were measured during walking before and after altering hoof angles. The effect of ground surface properties was also quantified.

MATERIALS AND METHODS

Mercury-in-Silastic strain gauges (MISS) were implanted in the superficial and deep digital flexor tendons (SDF, DDF), the suspensory ligament (SL) and distal check ligament (DCL) of five ponies weighing between 165 and 240 kg. Tendon strains were recorded while the ponies walked with unaltered hoof angles on an indoor flat rubber track and an outdoor sand track. Recordings were also made from the ponies with raised heels, raised toe and with an eggbar shoe (extended heels) while walking on the rubber track. Average strain patterns were obtained by ensemble averaging of 20-40 strides. Kinematic data were obtained using high-speed photography, synchronized with hoof acceleration recording [3]. From the kinematic data, obtained over a total of 20 strides, joint angles and length change of musculotendinous structures (MTS) during an average stride were calculated as described previously [4]. Forceplate recordings were made during all indoor measurements. Of each limb, a minimum of five strides was recorded and averaged. EMG's of the deep digital flexor (DDF) muscle and superficial digital flexor (SDF) muscle in the forelimb were recorded, and ensemble averaged over 20-40 strides.

Tendon loads were obtained from post mortem in situ calibration of the MISS [4]. Radiographs were made to determine geometric parameters, such as joint rotation centres and their distance to several tendons, describing the line of action of all musculotendinous structures [4].

RESULTS AND DISCUSSION

Fig. 1 shows the average tendon strains during normal walking in one representative subject. The corresponding MTS length patterns, calculated from the kinematic data, are shown in Fig. 2. In case of the completely passive structures (DCL and SL), there is a striking resemblance between tendon strain and MTS length. The DDF and SDF tendon strains are modulated by muscle contraction, which according to the EMG data occurs during the swing phase (60-100%) and at the beginning of the stance phase (0-20%). However, these tendons are very long compared to the corresponding muscles, so a large proportion of MTS length change will represent tendon strain.

The effect of shoeing on the peak tendon strains are summarized in Table 1. Across the whole group, the strain in the DCL is most affected by corrective shoeing. This is because it is a passive structure that

crosses the second interphalangeal (coffin) joint. Effects in the other structures are more easily compensated by muscle activity, which explains a less uniform response between different ponies. Within each individual however, the different shoeing conditions did have a significant effect, and the corresponding MTS length patterns suggest that this effect can be predicted from kinematic data. It can be concluded that the effect of corrective shoeing therapies may vary between individual patients. Automated movement analysis using the CODA-3 system may become a useful, non-invasive tool for individual evaluation of the effectiveness of such therapies.

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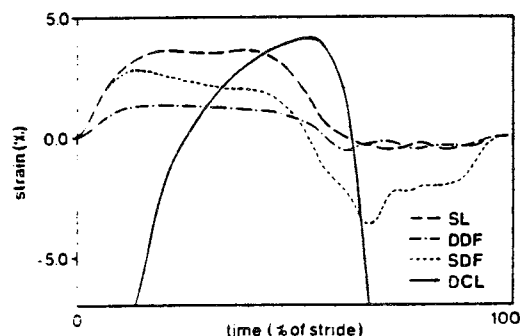


Fig. 1. Tendon strain patterns during normal walking in subject 4.

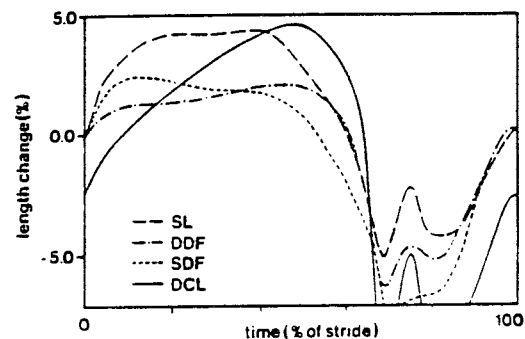


Fig. 2. MTS length patterns during normal walking in subject 4.

Table 1. Peak percentage strains in the suspensory ligament (SL), distal check ligament (DCL), deep digital flexor (DDF) and superficial digital flexor (SDF), measured with different types of shoeing. Values are average and standard deviation of 5 subjects. * indicates a significant difference from the normal situation ($p < 0.02$, two-tailed paired t-test).

	normal	extended heel	raised toe	raised heel
SL	3.7 ± 0.5	* 3.9 ± 0.5	3.7 ± 0.5	3.9 ± 0.5
DCL	4.7 ± 1.1	4.8 ± 0.8	* 5.5 ± 0.9	* 4.1 ± 1.2
DDF	2.2 ± 0.4	2.1 ± 0.5	2.2 ± 0.5	* 2.0 ± 0.4
SDF	3.4 ± 0.7	3.5 ± 0.5	* 3.2 ± 0.5	3.5 ± 0.7

SYSTEM APPROACH TO BIOMECHANICS OF PARETIC GAIT CONTROLLED BY
SUBCUTANEOUS ELECTRICAL STIMULATION

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INTRODUCTION. Functional electrical stimulation has been widely used in the rehabilitation of the extremities of paretic patients. There are two methods of stimulation: with surface and subcutaneous electrodes. In the first, desired movement depends on the daily positioning of electrodes done by a therapist or the patients themselves, while subcutaneous electrodes are attached onto the nerve by surgery. In the latter case, the proper position of electrodes it is very important in order to obtain functional and quality movement. For this purpose, quantitative methods for gait evaluation and prediction have been developed within the system approach paradigm. High correlation between gait corrections, EMG responses and the position of stimulating electrodes was found.

METHODOLOGY. Biomechanics of paretic gait behaviour controlled by subcutaneous electrical stimulation (1) was analysed within the system approach paradigm. Gait states represented by its variables, input function for which exist selective mapping into desired output function and criterion for gait performance estimation was defined. According to this, quantitative method for gait assesment under the influence of subcutaneous electrical stimulation was developed (2). It enables: prediction of gait correction of hemiparetic patients, elektrode positioning during surgery (3) and follow up of gait behaviour. Walking dynamics was determined by the vertical component of ground reaction force and its point of action(4), joint angles, space and time gait parameters, EMG response (M-waves) of main muscles and conduction velocity of stimulating nerve. The method of ankle movement prediction can be divided into three phases:

1. Preimplantation phase. Responses to single pulse surface stimulation of the common peroneal nerve were monitored in the ankle dorsal flexors, evertors and plantar flexors in order to obtain a strong activation in the tibialis anterior and a moderate one in the peroneus longus. Triceps surae was monitored to exclude unwanted plantar flexion. Besides single pulses, a 30 Hz train of pulses with the same amplitude and duration was applied during gait in order to estimate ankle dynamics with the electrodes in the same position. Gait dynamics without stimulation at the same gait velocity was also measured.

2. Implantation phase. The records of EMG from the previous phase were used during surgery as a guideline for determining implant position. Muscle activation was monitored during surgery while moving the electrodes along the nerve and the

responses compared to this EMG.

3. Postimplantation phase. Two to three weeks after surgery, quantitative gait evaluation with subcutaneous stimulation and EMG activity of stimulated muscles groups were measured and the results compared with those obtained before implantation. This phase is used to follow up on a patient's gait quality over time.

RESULTS AND DISCUSSION. Seventeen hemiparetic patients from a population of 31 already implanted were analysed. General conclusions were found:

A time domain analysis of ground reaction pattern, describing gait, indicated three characteristic groups of trajectories: a) a tendency to approach a normal gait trajectory; b) no change in trajectory; and c) following a period of stability in trajectories, some patients exhibited migration of the trajectory to a new undesirable state, suggesting a less functional stimulation. In the four patients who exhibited the response described in c), reimplantation was necessary. Following reimplantation in these patients, similar ground reaction patterns were observed following the initial implantation. The characteristic of ground reaction pattern caused by single channel peroneal implants are reproducible over time.

An almost linear relationship between electrode position, foot responses and gait correction was found. With subcutaneous stimulation, when it is indicated, gait performance could be achieved at least as good as with surface stimulation. Quality of gait pattern depends on electrode-nerve fixation. Carry over effects were observed in three cases.

Though desired foot movements controlled with functional electrical stimulation can be adequately predicted by the described methodology, the efficacy of the implanted stimulator may change with movement of the peroneal nerve electrode over a longer period.

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APPLICATION OF NONLINEAR FILTERING TO THE DECOMPOSITION OF SURFACE EMG SIGNALS

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Surface EMG-activity is the result of the superposition of biphasic action potentials which result in an interference pattern. Signal describes the active response to the neural stimulation of the muscle. Many severe problems affect the interpretation of the EMG, because especially during dynamic muscular work EMG-signals are nonstationary outputs of a basically nonlinear system and describe only the average contribution of the active elements of the muscle to the total tension /1,3/. Thus is it very difficult to model the relationship between EMG-activity and the external force produced by the muscle in all situations with a linear model, although monotonic relationships between EMG and force production have been found in restricted situations, such as during isometric work.

The running average of full-wave rectified EMG-activity has often in ergonomic studies been used for the evaluation of muscular strain. Furthermore, the accumulation of localized muscular fatigue has been monitored using the median frequency of EMG activity. In general, frequency-domain parameters appear to be more consistent than time domain parameters. Due to the inherent nonstationarities of EMG precautions have to be made when linear analysis methods, such as spectrum analysis, are being applied.

In ergonomic evaluations it is often important to be able to identify and classify the occurrence of certain potentially harmful work sequences during normal work. Especially high-intensity static work has been considered harmful, because the muscles cannot sustain high-level static contractions without muscular fatigue. During dynamic work the peak tissue strain is usually within the physical capability of the tissues, but prolongation of tissue loads can produce similar effects as in the case of static work. Long-term EMG measured during normal work can be used for the analysis and classification of muscular contractions. Pure static contraction is characterized by approximately constant levels of muscular activation and average EMG activity and nearly stationary EMG. In dynamic situations the muscular activation is changing rapidly leading to nonstationary EMG. Normal work involves a combination of static and various types of dynamic contractions. In physiological responses, such as EMG, the effects caused by static work are superimposed on the effects caused by the dynamic work.

We propose a new method for the analysis of surface EMG, where trend analysis is used for decomposing the signal into components describing separately the static and dynamic aspects of muscular strain. EMG-signals are decomposed into three components: static, dynamic and noise. Static components correspond to long trends (e.g. over 2 s) of the EMG amplitudes, dynamic to shorter trends (e.g. 0.05 - 2 s) and noise to the shortest transients, which cannot be directly connected to the force output of the muscle. Long trends of the signal describing the static work are also better correlated to the external force than the shorter transients /1/.

This type of decomposition cannot be done using normal linear methods. Application of e.g. long averaging filters for the analysis of long trends would distort the phase and amplitude characteristics of the signals due to sharp level changes and transients inherent in the EMG. For these reasons median filtering was used for the decomposition of EMG. Median filtering is a nonlinear filtering technique known for preserving sharp changes in signals and for being particularly effective in removing impulsive noise. In its basic form median filtering means choosing the median value of the samples in a moving window. The median filter has some interesting properties, which make it suitable for the analysis of EMG. When a signal is filtered with a median filter having a window length of n , all spikes of the signal shorter than $(n-1)/2$ are removed. This has been called the cut-off length of the filter. Despite this, even a sudden change in signal is preserved accurately [2,4]. These filtering properties can not be produced with any linear signal processing method.

In the proposed prefilter the EMG signals are first full-wave rectified and fed into a filter bank consisting of two median filters. Cut-off length t_s of the first median filter is set to correspond to the shortest allowed length of a static EMG component. Thus all trends longer than t_s are detected and the output of this filter is used for the estimation of static components of the signals. The cut-off length of the second median filter is set according to the shortest dynamic transient allowed, which usually is set to correspond to the fastest possible force transient produced by the muscle. Delays caused by the median filters are compensated and static part is subtracted from the output of the second median filter. Result corresponds to the dynamic part of the signal. Noise signal is formed through subtracting the compensated output of the second median filter from the full-wave rectified EMG signal.

The preprocessor has been implemented in software into a PC/AT-based system designed for the analysis of long-term EMG. In the system the decomposed EMG signals are used for later processing, where various potentially harmful work sequences such as long-lasting static work are detected and classified. Also the relative amount of static and dynamic muscular work is compared. The developed preprocessor is suitable also for the detection of high-intensity static sequences, where the original EMG-signal is nearly stationary and suitable for fatigue analysis.

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MUSCULAR FATIGUE IN MTM-PACED WORK

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The rationalization of working life had already introduced the system of measuring work movements in the 1930's. Work postures and movements of the body were measured in the MTM system in which each movement could be divided into TMU units (1/100 000 hour). Scientific evidence of how these TMU units correspond to physiological responses of the human body is not clear. The aims of this study are to evaluate muscular fatigue measured by EMG (electromyography) and to evaluate perceived exertion in a MTM-paced work.

Materials and Methods

Twelve healthy, female students volunteered to participate in MTM-paced arm work for one hour. Their mean age was 25,5 years (21-37 years). The task consisted of grasping a small cylinder from a box situated 45 cm away from the subject and to drop it down into a hole in the table. The movement was repeated 1,615 times during one hour. Surface electromyogram was recorded from the descending part of the right trapezius muscle. Amplification of the EMG signals was made linearly with a bandwidth of 5 to 500 Hz, using small Medinik amplifiers. The amplified EMG signals were recorded on a TEAC R-71 FM tape recorder with a bandwidth of DC-1250 Hz. The EMG signals were sampled with a frequency of 4 kHz and AD-converted with 12 bit accuracy (range of signal 5V) on a Masscomp 5500[®] computer. The Fast Fourier Transform (FFT) was used to obtain the power spectral density function (PSDF). FFT is based on a spectral resolution of approximately 4 Hz. Mean power frequency (MPF), i.e. mean of the spectral frequency, and root mean square (RMS), reflecting the values of the amplitudes in the signals, were calculated in parallel-processes. A scale, used by Borg, RPE (1) extending from 6 to 20, was used to measure subjective ratings of fatigue in various body regions proceeding, during and following the one hour of work.

Results

In the ratings, the perception of discomfort and fatigue, proceeding and following the work, increased significantly ($p < 0.02$) (Wilcoxon's matched-pairs, signed-ranks test) in the back of the head, the neck, shoulder, elbow, hand, upper and lower back. After nine minutes of work there were significantly higher ratings ($p < 0.05$) in the back of the head, neck, shoulder and elbow.

Regression analysis with time as the independent variable, and MPF and RMS, respectively, as dependent variables, were calculated for the myoelectric signals. A significant decrease ($p < 0.05$) in the MPF, and a simultaneous, significant increase ($p < 0.05$) in the RMS values indicating muscular fatigue, were found in eight persons. Continuous MTM-paced work could thus generate muscular fatigue in the trapezius muscle, as well as increased perception of discomfort.

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Segment timing and interaction in arm movement for
maximum velocity under differing load conditions.

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Several studies of striking and hitting movements have shown sequential patterns of segment action when maximum speed is a requirement (e.g. Milburn, 1982; Putnam, 1983). Other studies of vertical jumping and lifting have shown simultaneous or near simultaneous segment actions when loading is high (e.g. Koniar, 1973). On the basis of these findings a shift from sequential to simultaneous patterns of segment action with increasing load has been postulated by Hudson (1986).

The purpose of this study was to investigate segment timing and interaction in arm movements for maximum speed under different loading conditions. The study is one of several in progress aimed at developing the empirical basis for a linked segment sequencing theory (LSST), as an explanation of coordination in movement where coordination is defined as optimal timing and sequencing of segment actions.

Methodology

Three subjects performed a series of planar arm movements for maximum hand speed over a smooth flat surface under varying load conditions and spatial constraints. An NAC video camera operating at 200 Hz, placed overhead, was used to record trials. Reflective light markers were fixed to the shoulder, elbow, wrist and knuckles so that a kinematic description of the movement could be obtained using the Flextrac video motion analysis system.

In each experiment subjects performed 10 trials under five different load conditions (hand held loads of 0, 1, 2, 3, and 4 kg). Starting position for the movement was fixed, but finishing position and the accuracy requirement of the task were not constrained. The timing and sequencing peak segment angular velocities (PAV), were used to indicate if differences existed among trials.

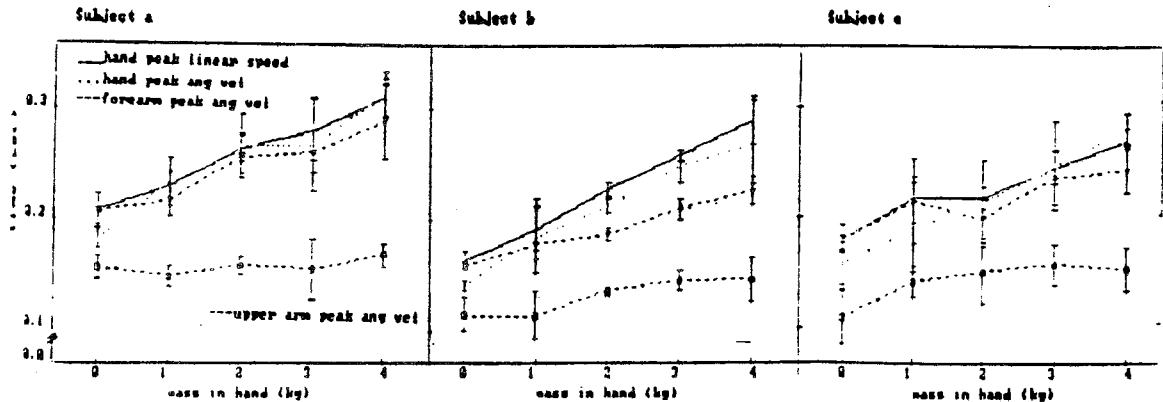
Results and Discussion

The times of the segment PAVs relative to the peak linear hand speed are shown in figure 1 for the three subjects (a, b, and c).

The angular velocity peaks of the upper-arm and forearm occur in a definite proximal to distal sequence. However, the distinction between times for forearm and hand velocity peaks is not as evident. A sequential pattern of timing of segment actions was observed for all conditions except for

hand PAV preceeding forearm PAV at zero load for all subjects with the opposite occurring at other loads. The sequential pattern of timing became more accentuated as loading increased as is shown in figure 1 by the divergence of the PAV-load curves.

Fig. 1. Timing of segment peak angular velocities and peak hand speed as mass is scaled.
means of 5 trials (error bars = std dev)



A repeated measures analysis of variance was conducted on the segment PAV times. Significant main effects were noted for segment [$F(2,16) = 19.42$, $p < 0.05$] and mass condition [$F(4,16) = 4.83$, $p < 0.01$], together with a significant segment by mass interaction [$F(8,16) = 3.46$, $p < 0.05$]. The significant main effects indicate that there were, as expected, differences in the timing of the PAV's of the segments and that the relative timing changed with increasing load. However, the segment by mass interaction indicates that the timing relationships between segments changed as load was increased.

The general principle of proximal to distal sequential timing was found to hold but there was no indication of a change from sequential to simultaneous timing as load increased.

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The influence of track hardness on gait kinematics in pacers

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Our previous studies (see: Wilson et al., 1986), have provided some evidence to suggest that pacing on a fast track surface requires different gait characteristics to pacing on a slow track under both training and racing conditions. The purpose of this study was to confirm our earlier observations on different tracks, quantifying track hardness, and to determine the extent to which pacers are able to modify their gait in pacing on surfaces of different hardnesses.

Methodology

Tracks: Two tracks used for training Standardbred pacers, Morrinsville (1200 m), and Curtin's Farms Racing Ltd, Morrinsville (1000 m), were used in the study. Track hardness was determined by a standard drop test. Peak impact force, time to peak force, force reduction and energy absorption were measured. Track surface composition, material and particle size, was also analysed.

Filming: Eight Standardbred pacers, four of which normally train on the Morrinsville track and four on the Curtin's Farms track, were filmed at warm-up speed and at speeds approximating mile ratings (min:sec/mile) of 2:10, 2:05, and their fastest pace on two different track surfaces. When the animals had completed the first film session on one track they were immediately walked (approximately 500 m) to the other track to begin the second session. All pacers were in full harness and were filmed in the finishing straight on a flat track surface. Drivers were instructed to drive at the required constant speed for a distance of 50 m through the filming zone. The camera was operated with a 1/4 shutter at 200 frames per second providing an effective shutter speed of 1/800 s. All pacers were classed as four years old and were in training for weekly racing.

Film Analysis: In this study the forward tip of each hoof at hoof contact and lift off, for a single stride, were digitally analysed at the appropriate film frames to produce displacement-time descriptions of the motion. Equipment used, scaling and timing procedures, and the reliability of single stride kinematic analysis of pacing gait have been described previously (Wilson et al., 1988).

Analysis of covariance on stride length and timing variables, with pacing speed the covariate, was used to determine if differences exist between gaits on the different surfaces.

Results and Discussion

The filmed section of the Curtin's track was considerably harder than that for the Morrinsville track (Peak force of 97 g vs 79 g and time to peak force of 3 ms vs 3.8 ms). These results were consistent with what might be expected from the particle size analysis of the track samples.

Animals paced at increased speeds with significantly increased suspension lengths ($p < 0.01$) on both tracks. There was also a significant difference in suspension length between the two tracks ($p = 0.03$). Suspension length increased by 26 % on the hard track compared to a 13 % increase on the soft track as pacing speed increased by 20 %. Stride lengths increased significantly ($p < 0.01$) by approximately 10 % with a 20 % increase in pacing speed and were significantly less ($p = 0.04$) on the hard track compared to the soft track. These changes in suspension length and stride length are consistent with a 26 % increase in suspension length and a 13 % increase in stride length for a pacing speed change of 22 % reported previously by Wilson et al. (1986) for a similar range of pacing speeds. There were no significant differences for any of the timing variables between tracks.

Animals trained on a hard track had shorter stride lengths, on 6 out of the 8 track and speed combinations, than animals trained on the soft track. These differences were observed when there have been no hopple length changes made for pacing on the different tracks and when hopple lengths were ostensibly the same. Further work is continuing to take account of animal size and hopple length differences between animals.

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EXERCISE-RELATED ADAPTATION IN GEOMETRY AND MECHANICAL PROPERTIES OF IMMATURE RAT TIBIA AND VERTEBRA

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INTRODUCTION

Wolff's Law indicates that bone will model and remodel in response to specific loading regimens. Immature bone appears especially sensitive to cyclic loading (Carter, 1984), and exercise during a period of rapid growth may alter geometry and mechanical properties of immature bone (Matsuda *et al.*, 1986). Because the skeletal system will experience differential loading during exercise, and trabecular bone may be especially sensitive to differences in loading (Rambaut and Johnson, 1979), adaptive changes will vary throughout the skeleton. Thus the purpose of this study was to quantify the geometrical and mechanical differences of tibiae and vertebral bodies as a result of endurance exercise in maturing rats.

METHODS

Female Sprague-Dawley rats (n=20, 8 wks, 150-170 g) were progressively trained (75-80% max aerobic capacity) on a motor driven treadmill 5 d/wk until they were running 1.8 km/h at a 10-12% grade for 1 hr/d. After 10 wk, the exercised rats and age-matched control rats (n=20) were killed. The right tibia and the sixth-lumbar vertebral centrum (L6) were isolated and equilibrated in buffer solution (50Mm potassium phosphate, pH 7.4, $37 \pm 0.5^\circ\text{C}$). As described previously (Shaw *et al.*, 1987), 3-point-bending tests to failure were conducted at midlength, with each tibia loaded in a medial-to-lateral direction, using a servocontrolled electromechanical testing system (Instron 1122). For all tibial mechanical tests, crosshead speed was 5 cm/min. Geometrical data were obtained by analyzing the contralateral tibia middiaphysal crosssections. L6 were compressed to 50% of their initial height at a 50%/s strain rate, while immersed in the buffer (Salem *et al.*, in press). Color slides were taken of the rostral surface of all L6 to determine cross-sectional areas. After digital conversion of the analog load-time data, computer programs (Run Technologies, Los Angeles, CA) were used to quantify the structural and material properties of tibia and L6.

RESULTS

The average weights of control (298 ± 25 g) and exercised (294 ± 18 g) animals at sacrifice were not significantly different. Tibia and L6 geometries are presented in Table 1.

TABLE 1. *Morphological Properties of Tibia and L6 Vertebra*

	Controls	Exercise
Tibial length (mm)	38.90 ± 1.30	37.80 ± 0.80 *
Tibial cortical crosssectional area (mm ²)	4.54 ± 0.56	4.28 ± 0.22
Tibial periosteal circumference (mm)	9.81 ± 0.88	9.27 ± 0.45 *
Vertebral-body cross-section (mm ²)	11.31 ± 1.71	11.00 ± 1.26
Vertebral-body volume (mm ³)	58.37 ± 8.87	56.12 ± 6.60

Values are means $\pm$ SD for tibia (n=15) and vertebral bodies (n=20). *Control mean is significantly ($p \leq 0.05$) greater than exercise mean.

Mechanical properties of tibia and L6 are presented in Table 2. There were no significant differences between control and exercised vertebra. Average tibial linear load (11%), maximum load (12%), energy to failure (17%), and energy to max load (17%), however, were significantly greater in the control than in the exercised group.

TABLE 2. *Mechanical Properties of Tibia and L6 Vertebra*

	Control	Exercise
Tibial Linear Load (N)	87.9 ± 12.3	79.1 ± 8.0 *
Tibial Max Load (N)	127.7 ± 15.6	115.7 ± 10.6 *
Tibial Energy to Max (N*s)	46.3 ± 8.9	39.5 ± 6.6 *
Tibial Energy to Fail (N*s)	106.2 ± 21.4	84.1 ± 27.5 *
Tibial Linear Stress (MPa)	276 ± 45	279 ± 36
Tibial Stiffness (kN/mm)	21.3 ± 4.0	19.3 ± 3.3
Tibial Elastic Modulus(GPa)	10.6 ± 2.2	11.1 ± 1.3
Vert-body Linear Load (N)	281.2 ± 76.4	264.7 ± 42.6
Vert-body Max Load (N)	395.5 ± 86.7	361.8 ± 56.5
Vert-body Linear Stress (MPa)	25.0 ± 5.9	24.3 ± 4.2
Vert-body Max Stress (MPa)	35.3 ± 7.5	33.2 ± 6.1
Vert-body Stiffness (kN/mm)	2.4 ± .64	2.3 ± .49
Vert-body Elastic Modulus(MPa)	11.2 ± 2.6	10.8 ± 2.5

Values are means ± SD for tibia (n=15) and vertebral bodies (n=20). *Control mean is significantly ($p \leq 0.05$) greater than exercise mean.

DISCUSSION

The exercise protocol resulted in significantly decreased tibial properties of length, periosteal circumference, linear load, maximum load, energy to maximum load, and energy to failure. Conversely, this exercise protocol did not significantly effect the differences in L6 geometry or mechanical properties, suggesting the tibia either recieved greater loading or cortical bone was more sensitive to the exercise-induced loads. Others have implied that trabecular bone is more sensitive to applied loads (Rambaut and Johnson, 1979); however, because the current protocol likely resulted in differential loading to the two bones, these findings may not be contradictory. While not significant, L6 geometrical and mechanical properties tended to decrease with exercise. This trend along with the significant differences seen in the tibia is in agreement with others who found decreased geometrical and mechanical properties of bone as a result of strenuous exercise in the immature animal (Matsuda *et al.*, 1986).

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THREE-DIMENSIONAL GEOMETRIC ANALYSIS OF HUMAN FEMURS FOR DESIGN OF TOTAL HIP ENDOPROSTHESES

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1. Objective of the Investigation

To obtain an accurate 3-dimensional surface model of the inner and outer shape of human femurs 81 specimens, representatively selected (age between 16 and 81), were embedded in plaster, transversely cut into slices of 3.55 mm thickness and all contours digitized resulting in a wire-frame model of the complete femur (CAD-System CATIATM). For comparison of the data and classification of the individual femurs, three out of 12 analyzed geometrical parameters were adequate to describe the inner surface required for implantation of cementless total hip endoprotheses. 26 different groups were filled with at least one specimen. 30 % of the femurs could be collected in three main groups were again the strongest group gathered 14 % of alle femurs. The bones were embedded in a box with a plaster in standardized position such that after they have been cutted into 3.55 mm thick slices each piece could be defined in its relative position and orientation in space. All of the slices were then digitized with their inner and outer contour.

The data set of each bone, consisting of at least 6.000 points in space with x, y, z-coordinates, was trasfered to the CAD-program (computer aided design program) CATIATM. The proximal/distal direction was defined parallel to the z-axis, so that in the xy-planes the points could be connected with a spline, giving two closed splines per plane, one for the inner-(called I-curve) and one outer-curve (called O-curve). The femur was modeled in distal direction up to the plane where the calculation of the cross-section-area, enclosed by the I-curve, gave a minimum. The result was a wire-frame model consisting of roughly 60 I-curves and 60 O-curves with a distance of 3.55 mm. Together with the measured ante-torsion=angle of each individual femur it was possible to define the corresponding frontal- and sagittal-plane which is important for cave modeling and later.

2. Design Classification

a) Sagittal View

The typical s-shaped profile of the proximal part of the femur when looking at the sagittal view was obtained by connecting the centers of gravity of the area of each I-Curve with a smoothed spline (called S-curve).

b) Frontal View

In the frontal view two characteristic curves (medial called M-curve; and lateral called L-curve) are obtained. Special care was taken of the design of the M-curve because the proximal part of this region is most important for supporting the implant. The M-curve was obtained with a smoothed spline running through the greatest elongation of the I-curves in medial direction. The curvature of the M-curve could be checked with the CAD-program in respect of curvature continuity.

The L-curve is a smoothed spline running through the greatest elongation of the I-curves in lateral direction. In proximal direction the

design criterion of the L-curve is given through the maximum allowable space for prostheses implantation.

In the distal part L-curve and M-curve together define a cone with a characteristic orientation of the cone-axis for each femur.

c) Transverse View

The shape of the cavity in the transverse section planes is designed with arcs (called A-curves) on the medial and lateral side. Taking into consideration the later mounting of the prostheses, the radius values of the arcs are not allowed to decrease in proximal direction.

3. Surface Model of the Cave

A surface model (free surface) is obtained through combining the curves from all three view-planes to a complex 3D-model. The above outlined design procedure gives two separate surfaces (medial, lateral) which can be regarded as a simplified surface model of the cave. The medial surface is a combination of S-curve and M-curve to form a guiding curve in space for the defined A-curves. This wire-frame model defines a free surface for the medial surface, and for the lateral surface.

A further spherical surface has to be described to define the shape of the caput femoris and thus to obtain the center of rotation of the hip-joint.

4. Classification of the different Caves

In order to define a set of various prostheses which fits optimal to the population of the different caves it was necessary to transform the computer models in space into a position relative to each other where a comparison of their geometry is possible.

The geometric-analysis showed that it is possible to collect the individuals into groups by using 3 characteristic parameters out of 12 analyzed. The predominating parameter is the mean value of the curvature of the M-curve within defined limits in the proximal part. Two further parameters are a measure of the distance of the cone-axis and the M-curve at two distinct locations.

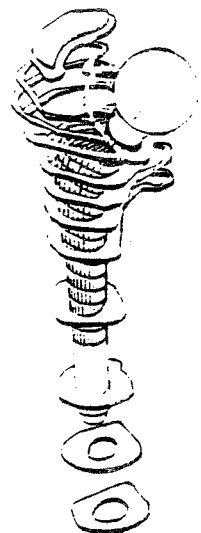
The distribution of these parameters has been evaluated. 26 groups are required in total. In the main group more than 10 % of the bones could be collected and the strongest three groups collected approximately 30 % of all bones.

5. Design of the Prostheses

The individual surface-models within each group were then analyzed to design a prostheses which

- gives maximum contact to the bone
- leaves at least 4 mm bone thickness
- is implantable
- does not require too much extra space for implantation or explantation

The CAD-system allows to design a free-surface model using these contour curves. The analysis for the contact-area bone/prostheses has shown that at least 70 % of the prostheses surface is serving as a real contact area.



KINETIC ANALYSIS OF CANINE GAIT BEFORE AND AFTER TOTAL HIP REPLACEMENT

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INTRODUCTION Animal models have frequently been used to analyze total hip replacement (THR) and evaluate the efficacy of various prosthetic designs (1-5). Such evaluations have traditionally focused upon quantitative assessment of implant fixation, bone remodeling, and alterations in strain. But, these studies have been more subjective in their evaluation of gait. Since restoration of normal gait is the goal of THR, a more quantitative assessment would be desirable. In addition, postoperative gait alterations after THR can profoundly affect the functional levels of strain. Therefore, joint loading becomes a critical factor to implant fixation and bone remodeling. For these reasons, a study was undertaken to quantify canine gait before and after cemented THR.

METHODS Five adult mixed breed dogs weighing 24-36 kg were selected for this study to receive cemented THR. Prior to surgery, they underwent testing to determine ground reaction forces (GRF) and kinematic data produced by their hind limbs during gait. Each dog was leash-walked across a Kistler force plate such that its right hind limb would be the only leg in contact with the force plate. Sagittal GRF were collected at 5000 Hz with a personal computer and kinematic data were collected at 100 fps using a Milliken high speed, motion picture camera. Then, the dog was walked in the opposite direction, collecting similar data for the left hind limb. This procedure was continued until a minimum of 3 repeatable experiments were collected on each hind limb. Unilateral cemented total hip replacement (THR) was then performed on all dogs. Post surgery, the dogs were allowed to bear weight as tolerated and to exercise freely in individual runs. The above procedure to obtain ground reaction forces was repeated at 1, 2, 3, and 4 month intervals postoperatively. Markers at the greater trochanter, lateral femoral condyle, lateral malleolus and the proximal phalanx of the fifth digit were digitized from projected images. Coordinate positions of the segment anatomical markers were scaled and filtered at 5 Hz. The force plate data were synchronized to the film, and a rigid link, segmental model was used to calculate intersegmental forces and moments at the hip, stifle, and tarsus. Segmental parameters were experimentally determined. This abstract reports intersegmental data for one dog.

RESULTS AND DISCUSSION Intersegmental vertical forces and moments are shown in Figures 1 and 2, respectively. They indicate an immediate postoperative drop in joint loading of the implanted limb. By the fourth postoperative month, both forces and moments have effectively returned to their normal magnitudes and patterns. Substantial compensation of the joint forces and moments was observed at the first postoperative month on the unoperated side with forces and moments elevated in all three joints. This compensation gradually subsided until effectively returning to normal by the fourth month. Relative changes in the kinematic data after THR were much smaller than the changes shown in the kinetic variables (Figures 1 and 2). Such results suggest that subjective clinical assessments based largely upon observed motion can be deceptive and that quantification assessment of kinetic variables is a more valid method to determine the normality of canine gait. These results are being further studied on the other four dogs.

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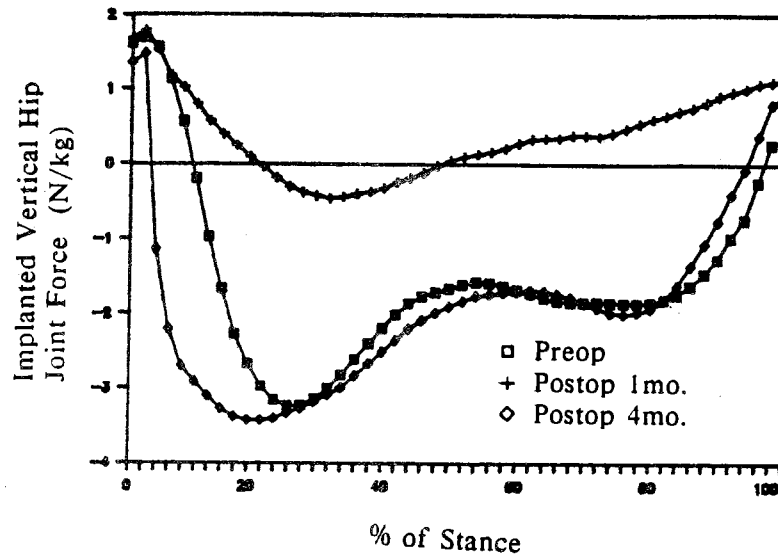


Figure 1 Each curve represents an average over six trials of the vertical resultant force at the implanted hip joint normalized per kg of body weight and plotted over the stance phase of walking. Averages are given for preoperative, 1 month and 4 month postoperative analyses.

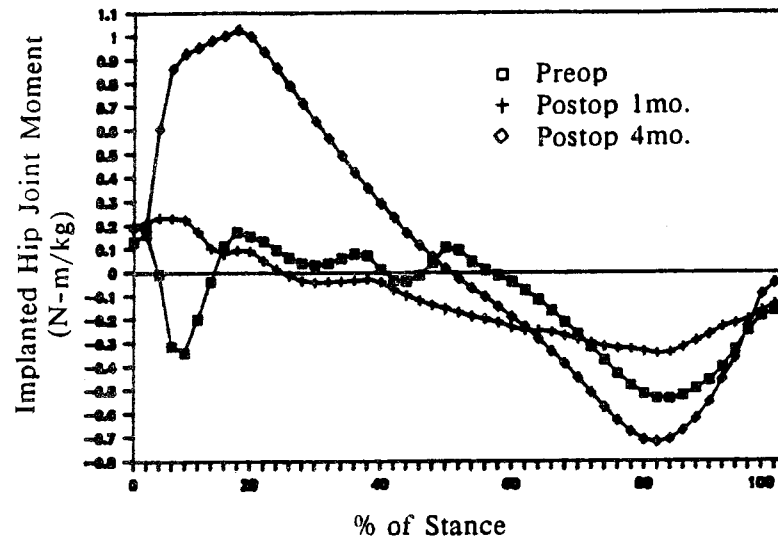


Figure 2 Each curve represents an average over six trials of the resultant hip joint moments in a sagittal plane normalized per kg of body weight and plotted over the stance phase of walking. Averages are given for preoperative, 1 month and 4 month postoperative analyses.

A THREE-DIMENSIONAL MODEL FOR LEG AND FOOT KINEMATICAL STUDIES

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INTRODUCTION

The actual possibilities of 3D, high accuracy data acquisition open a new perspective for the analysis of human motion. In fact a proper arrangement of markers allows to obtain all the three-dimensional information required. As a first step a simple model of the leg and foot system was developed to the purpose of a detailed analysis of absolute and relative motions during walking.

METHOD

Data acquisition system. The ELITE system (Ferrigno and Pedotti, 1985) was used to detect the position in space of a number of retroreflective markers. The adopted configuration was three-dimensional (two TV-camera) at 50 Hz. After data acquisition, some standard procedures were used for processing the rough data (Ferrigno and Gussoni, 1988).

Modelling. The leg, the foot body and the toes as a whole were schematized as rigid bodies. They were defined in space by the coordinates of the baricenter and by the orientation of the principal axes of inertia. The solids that ideally represent each of the body segments considered (fig.1A) are a truncated cone for the leg, tetrahedon for the foot body and a triangular surface for the toes as a whole. Definition in space was made by means of the Euler angles between the mobile systems of axes and a fixed reference frame.

Markers' arrangement. Two markers were arranged on the fibula head and lateral malleolus respectively and a third marker on a stick rigidly fixed on the anterior face of the tibia (fig.1B). The width of the leg at the two levels of the fibula head and of the lateral malleolus were measured on the subject. The foot position in space was defined by one marker on the heel and by two markers arranged on a stick perpendicular to the foot base, in correspondence with the head of the fifth metatarsal head. The distance between the first and the fifth metatarsal heads and between the first metatarsal head and the heel were also measured. The triangular surface representing the toes was considered hinged to a rotation axis connecting the first and the fifth metatarsal heads. Thus only one marker on the tip of the big toe was sufficient to define its position.

RESULTS AND DISCUSSION

In fig.1C a perspective view of one stride performed by a normal subject is reported. For the sake of simplicity only the long axis of the leg and the foot sole have been represented. A transverse section of the leg is schematized at the level of the fibula head, and the sagittal and frontal axes are reported. The projection of the foot sole on the horizontal plane is represented. Only one every two frames is considered. In fig.2 the time course of the angles of the leg in absolute reference are reported above. Flexion-extension angles of the

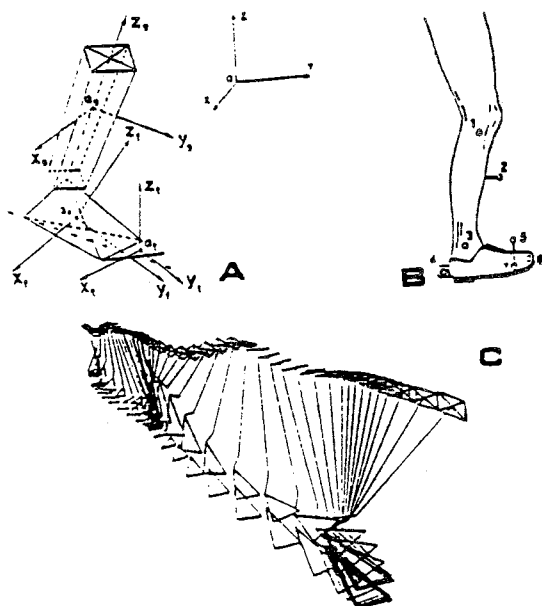


Fig.1A: Geometrical model
 Fig.1B: Markers arrangement
 Fig.1C: Model implementation

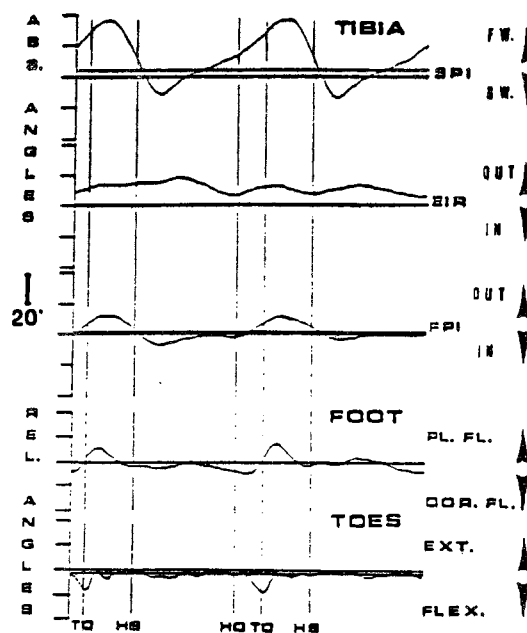


Fig.2: Absolute and relative angles

foot in relation to the leg and of the toes in relation to the foot are reported below.

The model proposed is designed to obtain all the three-dimensional information about leg and foot movements during locomotion; the approach seems particularly indicated for studying pathological gaits in which rotations around the longitudinal axes and large movements outside the sagittal plane occur. The choice of reference frame originating from the barycenter and oriented as the principal axes of inertia of each body segment has been done by considering the possibility of solving the inverse dynamics problem.

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APPLICATION OF COMPUTERIZED TOMOGRAPHY FOR OBTAINING INERTIA QUANTITIES OF THE HUMAN TRUNK

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Since computer assisted tomography was used for obtaining images of the slices of the human body, and was applicated to the medicine problems other applications of this method appeared. One of these is obtaining inertia quantities of the human body.

In 1976 Huang and Wu presented a method where they converted pixel by pixel CT numbers of each scan to mass densities. However dealing with live subjects one can use only scans of routine medical examinations. During these examinations patients before to be scanned are given some contrast fluid. This fluid causes different darkning of tissues of similar mass densities on the monitor's screen or on the photograph. Rodrigue and Gagnon /1981/ used data on density of three forearm tissues /bone, muscle, and fat/ to calculate inertia quantities of this segment with the help of computerized tomography. They calculated volume of each tissue from the consecutive scans of forearm and then they obtained mass and location of forearm's center of mass. Using this method one can obtain inertia quantities only along one /longitudinal/ axis.

In 1988 Erdmann and Gos measured mass density of about 50 human trunk tissues. Those data can be used for further investigations of the inertia quantities.

In this paper a method of obtaining inertia quantities of the trunk using mass densities of the trunk tissues is presented. The trunk is divided onto consecutive layers 8 mm thick - Figure 1. Then, the picture of every trunk layer /Figure 2/ is divided onto finite number of elements /columns and rows/ - Erdmann /1988/. Each element of known volume /0.2 cm³/ belonging to the particular tissue is given its density value /Figure 3/.

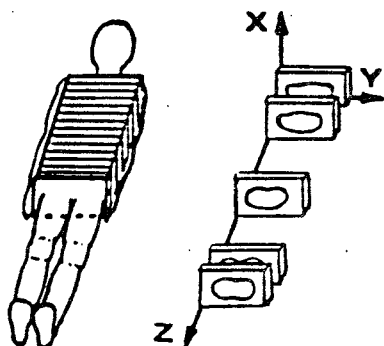


Fig. 1. Layers of a trunk.

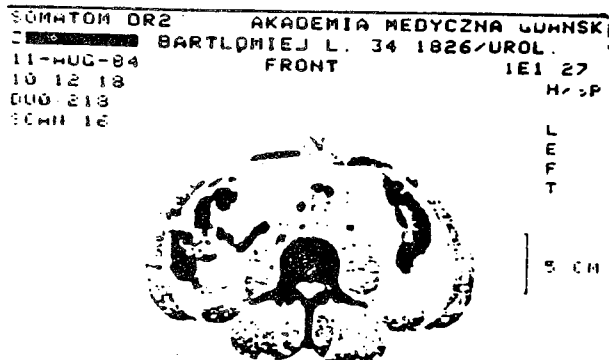


Fig. 2. CT image.

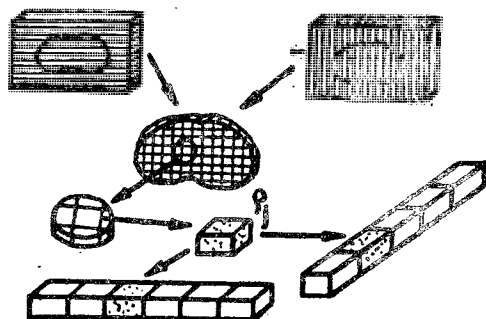


Fig. 3. Density ρ of j element.

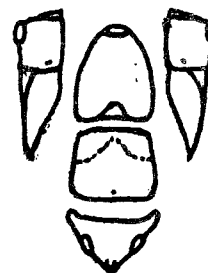


Fig. 4. Trunk segments.

The division of a trunk can be done according to Dempster /1955/ with additional division of the abdomino-pelvis region. In this way following segments are obtained: two shoulders, thorax, abdomen, and pelvis - Figure 4.

From every layer a proper part is taken which belongs to the proper trunk segment - Figure 5. Joining fragments from consecutive layers for particular segment, one can calculate mass, location of center of mass, and central radius of inertia for each segment.

With this new method obtaining of inertia quantities for a person other than this given in the analyzed picture is possible. If, e.g., the thickness of a fat tissue of an analysed person is in the picture 10 mm, and we want to know what will be the inertia values for a person with the thickness of a fat tissue, let say, 40 mm, we can draw in the picture additional area of a fat tissue and calculate for this changed condition new data - Figure 6.

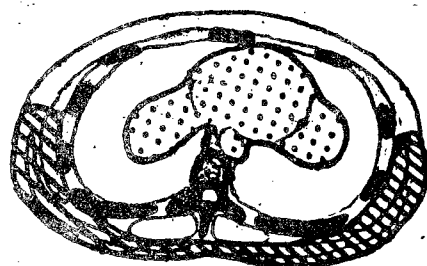


Fig. 5. Dashed area - shoulders, dotted - abdomen, the rest - thorax.



Fig. 6. Additional fat tissue drawn in.

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THE EFFECT OF PRETENSION AND MUSCLE LENGTH ON THE PROCESSION OF KNEE EXTENSION

Jozsef Tihanyi

Muscle tension development under isometric condotions always preceeds muscle shortening while energy is being stored in the serial elastic components (Hill, 1950). The amount of elastic energy is a function of the contractile force usually called pretension. The potential or elastic energy can be transformed into kinetic energy when a muscle is allowed to shorten. Additional energy can be generated by the contractile elements during muscle shortening depending on the number of the active cross-bridges. The human knee extensor muscles can exert their summed maximum force at a knee angle position of 120-135 degrees (Smidt, 1973, Tihanyi, 1981). The peak velocity of knee extension depends on the degree of angular displacement, but it is independent of the maximum force which can be generated at the respective knee angle position from where the extension is commenced when the leg is unloaded (Tihanyi, Apor, & Fekete, 1982). The present study aimed to investigate the particular role of the elastic and contractile components to the process of knee extension under various conditions.

METHODS

Eight power athletes (age: 23.1 ± 1.9 , height: 185.0 ± 5.7 , weight: 75.5 ± 4.9) joined in the experiment. A special dynamometer was used, (described elsewhere ; Tihanyi 1982) which made the quick release methods possible. Maximum voluntary contraction ($1.0 F_0$, Nm) was estimated at knee angle positions (KAP) of 90 and 135 degrees ($1.0 F_0$, KAP 90 and $1.0 F_0$, KAP 135). Muscle tension was maintained for 5 sec at $1.0 F_0$ (MT 1.0°) and $0.5 F_0$ (MT 0.5) or $1.0 F_0$ and $0.5 F_0$ was developed with the highest rate (DT 1.0 and DT 0.5) at knee angle positions of 90 and 135 degrees then the leg was released. For loading the muscles during concentric contraction 0.33, 0.73, 1.13, 1.53, 1.93, 2.33 kgm^2 inertial weights were applied equivalent to 4, 15, 21, 26, 31, 36 % of F_0 on the average. Peak angular velocity (V_p , $\text{Rad} \times \text{sec}^{-1}$) was estimated from the velocity-time curves and initial angular acceleration (" a_i ", $\text{Rad} \times (\text{sec}^2)^{-1}$) was calculated. Mean and standard deviation was calculated and the Student t-test was used to detect significant differences in the various criteria (at the $p < 0.05$ level).

RESULTS

Maximum isometric torque for KAP 90 was found to be 185.8 ± 25.5 Nm which is lower by 27.5 % than that of KAP 135 which was 256.2 ± 31.2 Nm. In the MT1.0 condition V_p for KAP 90 was significantly higher by comparision with KAP 135 when the two lowest loads were applied (Fig 1A). Also, V_p was significantly higher for KAP 90 than for KAP 135 at the four lowest loads (Fig 1B). In both cases the differences decreased gradually as a function of the increasing loads. Contrary, " a_i " for KAP 135 was higher in comparision with KAP 90 at each load regarding the respective conditions (Fig 2). Mean V_p values were almost identical for MT 1.0 and DT 0.5 in KAP 90 experimental situation (Fig 1A and 1B). In contrast, " a_i " values were always higher for MT 1.0 than for DT 0.5 in KAP 135 and the

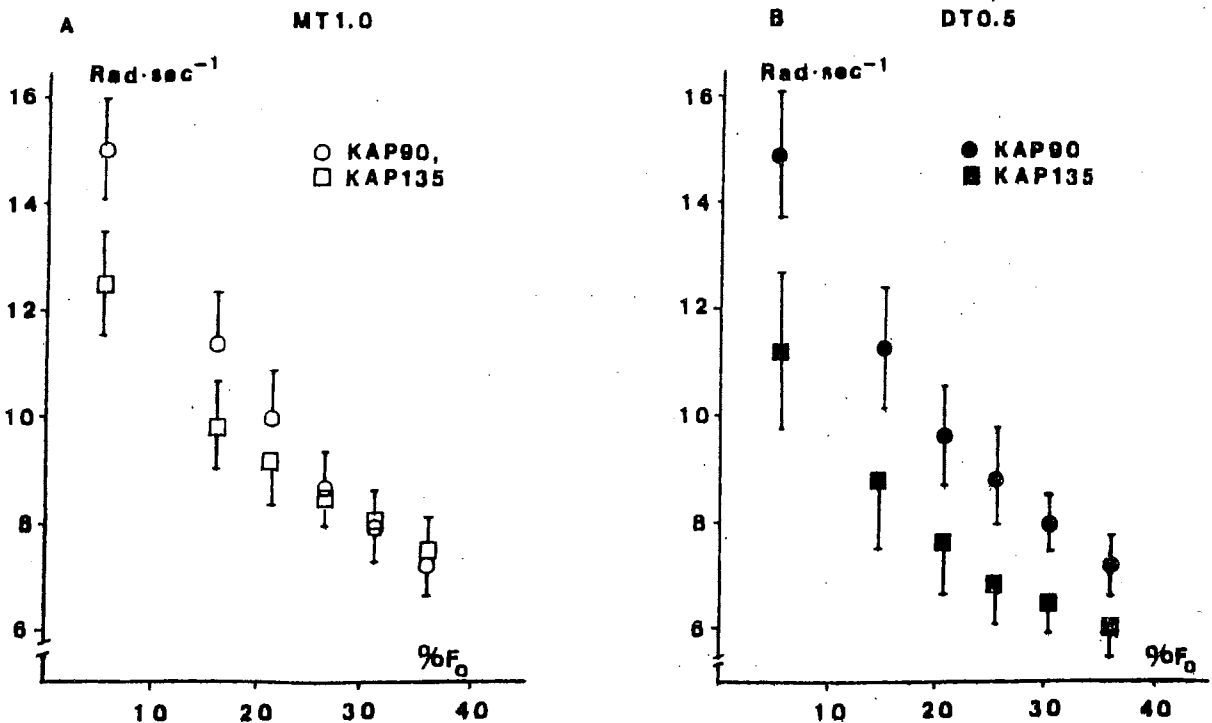
differences increased gradually as a function of the increasing loads which is significant at the highest two loads applied. No significant difference was found for "a₁" as MT 1.0 and DT 0.5 conditions were compared in both KAP 90 and KAP 135 situations, respectively. However, mean "a₁" values increased to a higher degree for DT 0.5 than for MT 1.0 as a function of the increasing load in both KAP 90 and KAP 135 (Fig 2). There was no significant differences in either V or "a₁" values regarding MT 1.0 and DT 1.0 conditions. However, V and "a₁" were significantly higher for DT 0.5 as compared to MT 0.5.

DISCUSSION

The findings of the present study suggest that the "a₁" is proportional to the pretension as Kaneko (1978) predicted. This means that the stored elastic energy is transformed into kinetic energy at the onset of muscle shortening. However, V₀ can also indicate the degree of pretension when a muscle begins to shorten from the same muscle length and the tension is developed in the same manner. Since same muscle tension can be developed by different motor unit recruitment patterns (DT 0.5, DT 1.0 and MT 1.0), it may modify the effect of elastic component and may increase the significance of the contractile components on the process of knee extension. The present findings indicate that the shorter the angular displacement available for knee extension, the greater is the role of the elastic component. In contrast, the contractile components play a more significant role when muscle length of knee extensors is longer than the optimum. Similarly, when the difference between the developed muscle tension and the load to be moved decreases, i.e. the loads applied are heavier and heavier, the role of the contractile components becomes dominant in contrast to elastic components since the shortening of the elastic components is strongly prevented. These statements are supported by the findings of Tihanyi, Apor, & Fekete (1982) that no significant correlation was found between percentage fiber distribution and V₀ applying loads below 0.25 F₀ as well as "a₁".

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DYNAMIC ANALYSIS OF ISOINERTIAL LIFTING TECHNIQUE

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The literature describing studies of lifting and manual materials handling provides abundant examples of predictive models that have been developed to determine maximal lifting capacities (Ayoub et al., 1980b; Kroemer, 1983; Jiang et al., 1986). Grieve (1984) urged researchers to provide data on velocity, force, and power output for lifting actions as these dynamic measures play an important role in determining the stress on the musculoskeletal system. Dynamic characteristics are important in both tasks and tests as predictive tests are expected to measure the salient muscular strength components necessary for task performance. However, technique and strength are variables which have been intertwined (Freivalds et al., 1984; Bush-Joseph et al., 1988).

One lifting device which has been widely used in the armed forces and some industries is an isoinertial lifting machine (ILM) (Kroemer, 1983; McDaniel, 1983; Stevenson et al., 1988). The ILM has proven to have modest to high correlations to manual materials handling tasks (Ayoub et al., 1987; Jiang et al., 1986; Stevenson et al., 1988) but in each case maximal mass lifted was the only correlate of test performance. The purpose of this study was to develop an analysis system to examine the technique used in ILM lifting and to explain test profiles by means of dynamic analysis.

Methods and Procedures

The ILM was instrumented with a displacement gauge attached to the armature. Data from this gauge were fed into an A/D Board and microcomputer at 100Hz. To capture accurate data at the initial phase, a 100ms buffer was used to store the first 3mm of armature movement along with 2s of maximal lift information. Once stored, the displacement data were processed with a second order Bitterworth filter at 5Hz. Velocity was calculated by the first forward differentiation method. Data were smoothed again prior to acceleration, force and power calculations.

Maximal dynamic profiles were captured on 33 women and 99 men executing a 1.82 m lift using the protocol of Ayoub (1986). The dynamic curves were quantified by 37 discrete points representative of time, displacement, velocity, force and power for each of the key points: total time, peak velocity, minimum velocity, peak acceleration/force, minimum acceleration/force, second peak acceleration/force, peak power, and minimum power. A principal component analysis was used to identify the underlying relationship of between variables. The selection of variables was based on a clean loading of variables onto separate "factors" based on Pearson product moment correlation coefficients, Kaiser's eigenvalue criterion test and the scree test.

Results and Discussion

The 1.82 isoinertial lift can be described by three phases: (1) a powerful pulling phase, which incorporates measures of maximum acceleration/force, velocity and power; (2) a wrist changeover phase where momentum is required for minimum force and acceleration values; and (3) a pushing phase, where the second maximal force and acceleration values were attained. Ninety-six percent of maximal ILM scores could be explained by five factors: (1) magnitude of force/power primarily from the pulling phase (38.5%); (2) timing of the lift (20.9%); (3) momentum during wrist changeover (15%); (4) timing and position of peak force in pulling phase (11.5%); and, (5) stroke of arm during the pushing phase (10.%). Although the force and power variables were significantly different between males and females, it were also a substantial number of significant differences in timing and displacement variables. Technique differences could cause an underestimation of female lifting capacity as timing and displacement points are more indicative of technique than pure strength. In conclusion, an instrumented isoinertial lifting devise can provide dynamic information which can be used to explain lifting technique and account for 96% of the variance in maximal isoinertial lift score.

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PHASE DIAGRAM APPLIED IN EVALUATING TECHNIQUE FACTORS IN RUNNING

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The phase diagram has long been utilized in mechanics to study cycling or repetitive phenomena. In biomechanics, Jaworek (1987) and Vermeulen and Nigg (1987) attempted to apply it in constructing a regulation model of the knee joint, and studying muscle activity, respectively, in running. The purpose of this study was to show the usefulness of the phase diagram, where angular velocity was given as a function of knee joint angle, in studying the effects of training, running velocity and (possibly) fatigue on running technique.

Material and Procedure

Four subjects participated in this study. Subject 1 was filmed twice: at the beginning of her athletic training (age 13) and four years later when, in Poland, she was ranked in the top five in the 100m hurdles and 200m, for her age group. Subject 2 was the Polish national champion in the women's 800m. Subjects 3 and 4 were male long distance runners of low caliber. All subjects were filmed with a Locam high speed camera set at 100fps, at running velocities of 8.1 and 7.3m/sec (Subject 1), 7m/sec (Subject 2), and 3.6m/sec (Subjects 3 and 4). Subjects 1 and 2 were filmed in practice. Subjects 3 and 4 were filmed at the 37th kilometer while competing in a marathon. The raw data extracted from the film was digitally filtered before being submitted to further analysis.

Interpretation and Conclusion

The effect of training on the running technique of Subject 1 is presented in Figure 1.

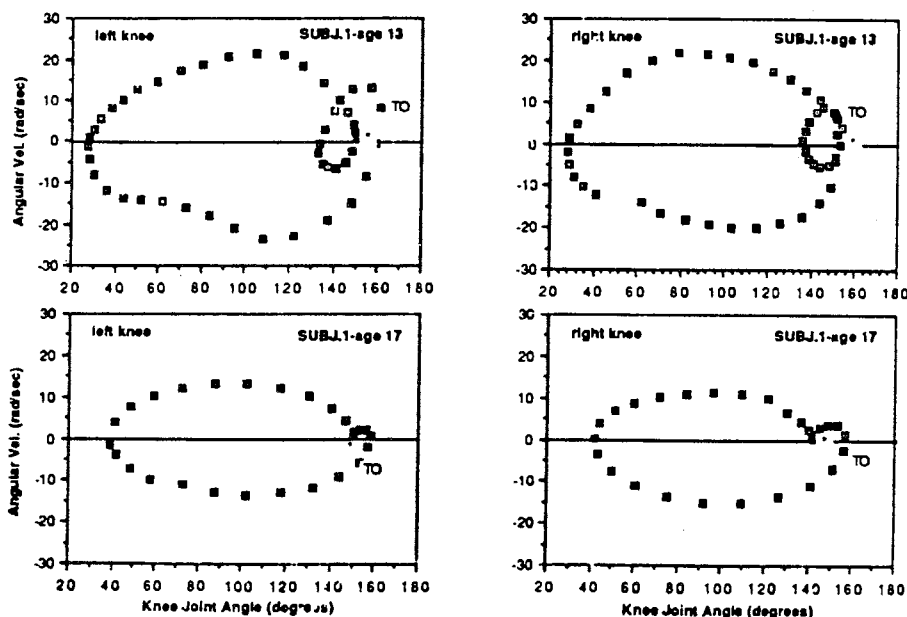


Figure 1. Phase diagram of knee motion during the swing phase (from take off-TO-to foot strike-") and support phase (from " to TO) for Subj.1. All loops are clockwise.

Asymmetry between right and left knee joint motion, high flexion and extension velocities, and (partially) erratic changes in each joint's angle and velocity characterized her running technique at the beginning of her career (SUBJ.1-age 13). All these characteristics have changed after 4 years of training. In addition, practice resulted in remarkable changes in the mechanics of the support phase. At age 13 she began flexing the knee after foot strike (from 150° to 133°), and took off at the moment of maximum knee extension (at 161°). At age 17 there was an evident arrest of knee joint motion at foot strike, with only extension occurring during the whole support phase. This pattern of support phase was possibly due to practicing the 100m hurdles. However, it is not recommended in running.

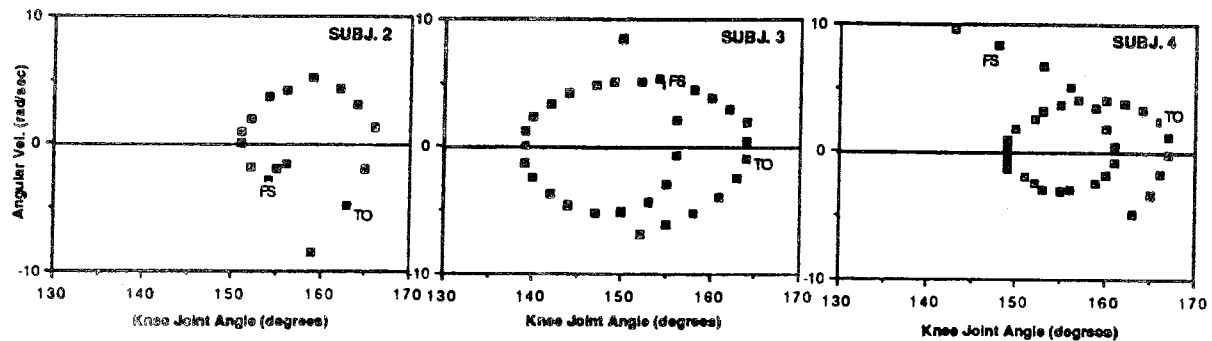


Figure 2. Phase diagram of knee angle during support phase in running from foot strike (FS) to take off (TO) for Subjects 2, 3, and 4. All loops are clockwise.

Figure 2 shows the support phases of the middle distance runner (SUBJ.2) and the two marathoners (SUBJs. 3 and 4). The most pronounced difference between the pattern of Subject 2 and the patterns of Subjects 3 and 4 was that the foot strike in the former was followed by further flexion before extension began and coincided with the highest (flexion) velocity. This pattern of foot strike is highly recommended in sprinting (Mann and Herman, 1985). Take off occurred after knee flexion was again initiated. In contrast, foot strike was followed by further knee extension in the marathoners. These differences in foot placement and knee joint motion may be due to differences in the runners' caliber, as well as differences in running velocity, and/or fatigue.

In conclusion, the phase diagram may be considered an easy and useful tool for detecting specific characteristics of different running techniques. From this point of view, the experienced coach may benefit from its usage.

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"ON A NEW MULTIFUNCTIONAL ISOKINETIC ERGOMETER"

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Undoubtedly, the ergometric evaluation of an athlete during continuous isokinetic work is a rather complex issue to deal with.

As a matter of fact, so far no ergometer able to work under real isokinetic conditions featuring a 360-degree and extended-through-time motion has ever been accomplished.

However, at the Institute of Sports Science a new ergometer meeting the above requirements has been designed (by Prof. Antonio dal Monte)(Fig.1).

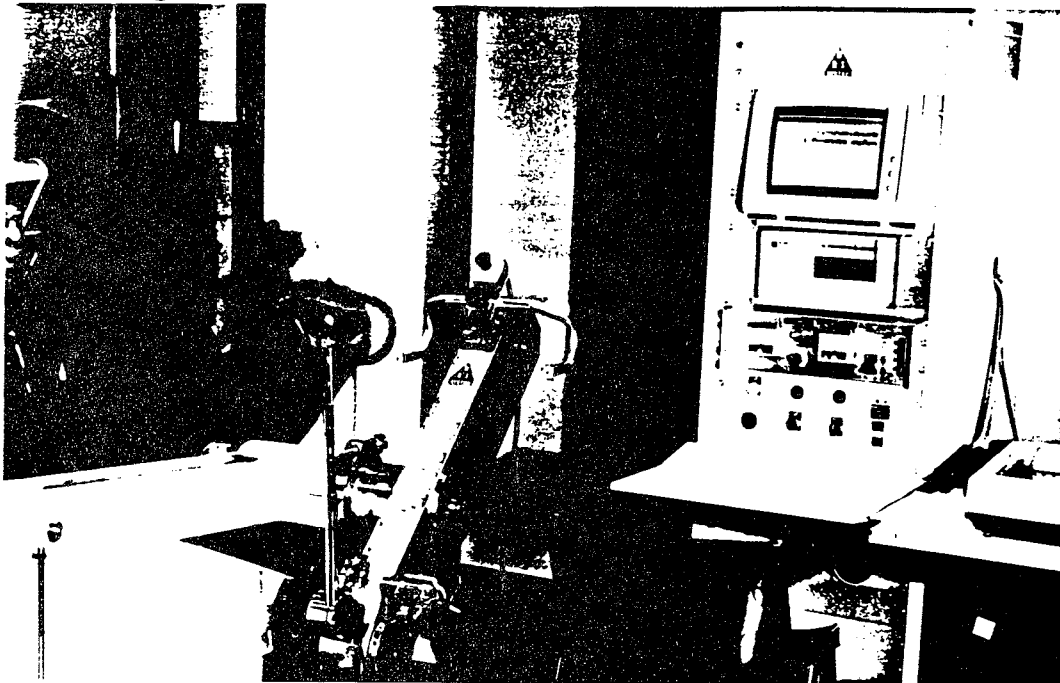


FIG.1

The new apparatus allows isokinetic work to be performed with the upper limbs, the lower and all four limbs simultaneously.

In addition, the position into space may be changed thus allowing the tests to be carried out in different positions (seated, prone, etc.). The apparatus has been designed in such a manner that the following tests may be carried out:

- 1) isokinetic strength-velocity curve;
 - 2) isokinetic power-velocity curve;
 - 3) constant-torque test (Wingate test);
 - 4) constant power test (i.e., specific ergometer tests for cycling).
- Tests were carried out on athletes practising cycling and handball to depict the Strength-Velocity and Power-Velocity Curves.

The apparatus measures the work, peak and average strength and power achieved during the various stages of the test.

Moreover, it allows for the interpolation, by means of software, of the values obtained for the depiction of the graphic curves (FIG.2-3).

Furthermore, both negative as well as positive work may also be measured (resistance to the drag opposed to the velocity proper to the muscle).

Variable-rotation-range and extra velocity work (that is a velocity higher than the one the subject may reach under voluntary conditions) may be performed.

The results achieved provide specific data on the investigated characteristics that enable us to have these new evaluation methods support those currently used in order to further enhance the opportunities for analysing the physical qualities of athletes. Moreover, owing to its very characteristics, the ergometer proves to be most suitable for the planning and monitoring of the athlete's post-traumatic rehabilitation and identification of deficits which, by using customary methods of analysis, would not be detected (deficits occurring only at specific execution velocities, etc.).

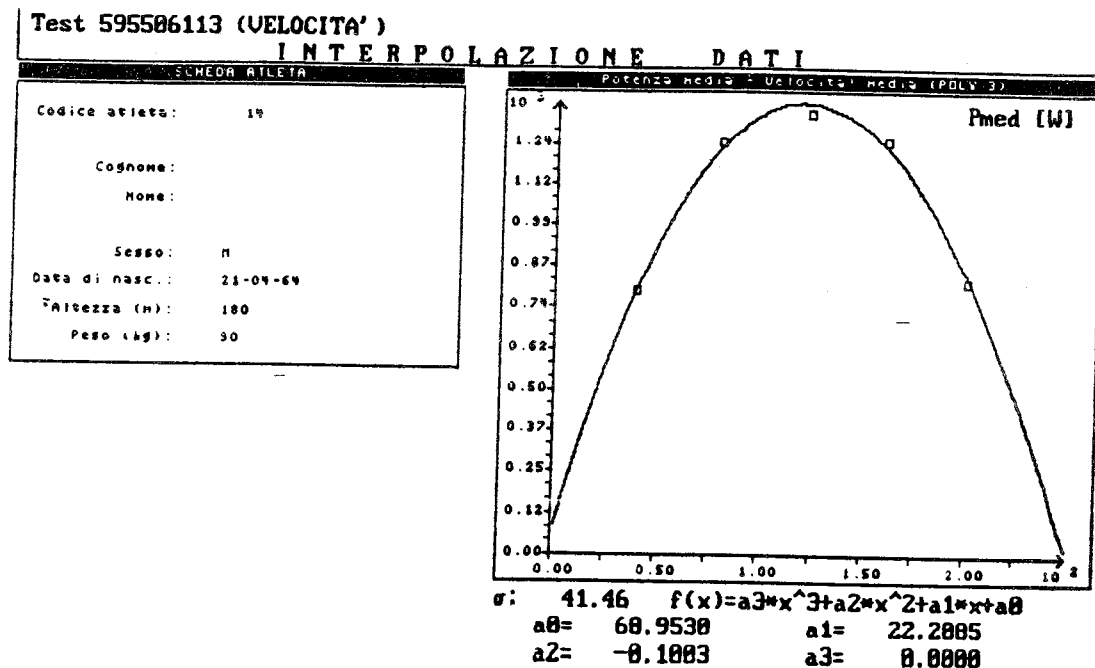


FIG.2

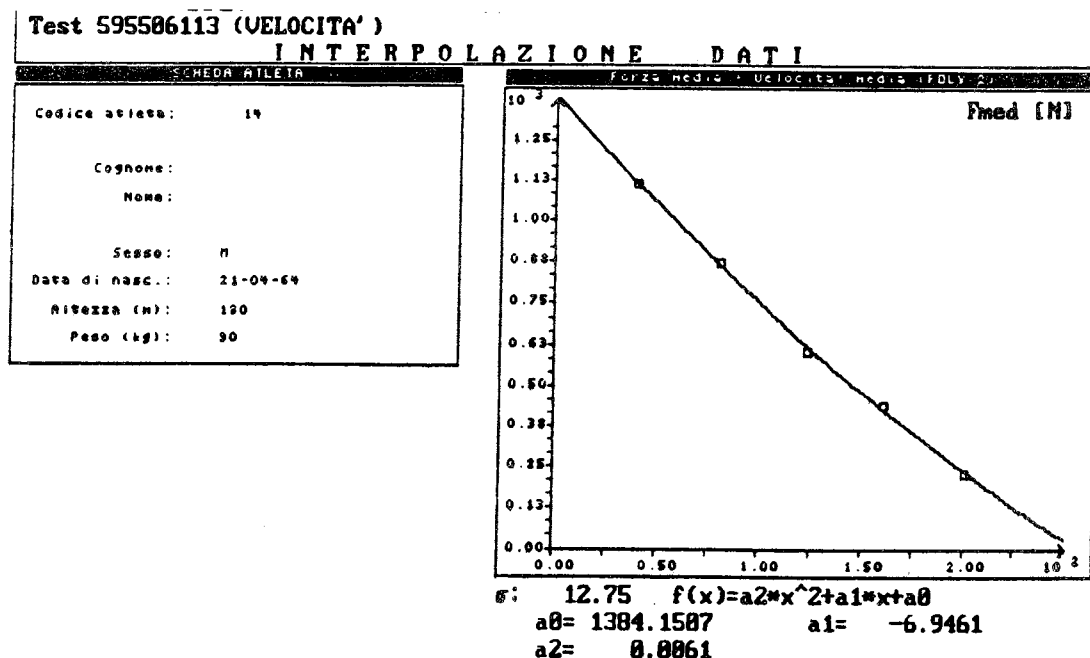


FIG.3

EFFECTS OF PRACTICE ON MYOELECTRIC AND KINEMATIC CORRELATES OF HUMAN MOVEMENT

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Studies of complex human movements have shown that performance improvement can extend over thousands or even millions of repetitions. However, it is not evident from such observations that extremely simple and everyday movements are subject to the same improvement through practice. Therefore, we had eight subjects, with no previous experience as experimental subjects, practice generating fast, accurate elbow flexion movements (54° distance to a 3° target) over seven separate experimental sessions (1400 trials). The instructions emphasized movement speed. Angle and acceleration were directly measured and velocity computed. EMG activity from the biceps and triceps were also recorded.

The results demonstrate increases in the peaks of selected kinematic variables (peak acceleration, peak deceleration (figure 1) and peak velocity (figure 2).

PEAK ACCELERATION AND DECELERATION

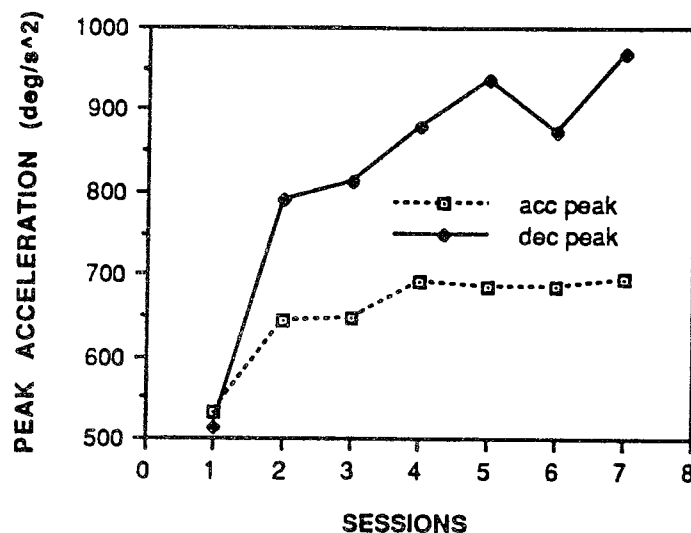


Figure1: Averaged peak acceleration and peak deceleration for seven experimental sessions (first 20 trials in block of 200). Data are for one subject.

The variability of peak velocity, however, decreased as is shown in Figure 2. These findings were most noticeable for the first 600 to 1000 trials by which time the performance of most subjects had stabilized. The data suggest that the most significant change occurred in the deceleration component (figure 1, compare peak deceleration to peak acceleration)

PEAK AND STANDARD DEVIATION OF VELOCITY

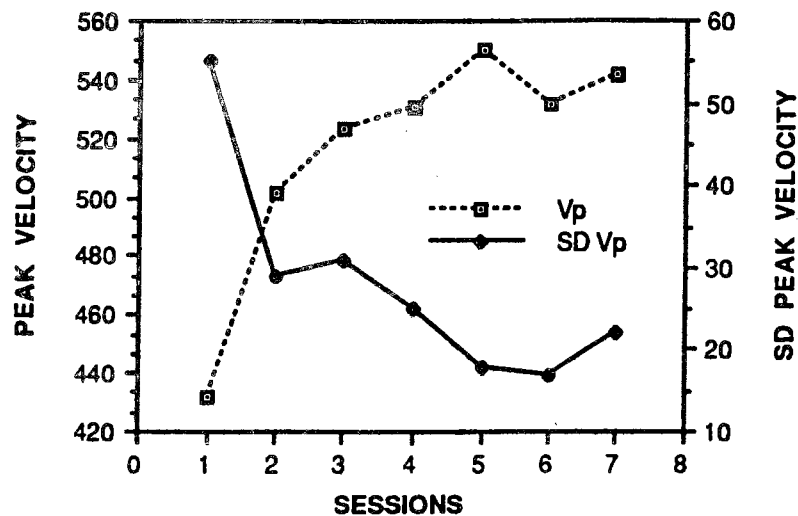


Figure 2: Averaged peak velocity (first 20 trials in block of 200) and its standard deviation for seven experimental sessions. The data are for one subject.

The EMG data suggest that these changes in kinematics can be attributed to changes in the neural commands. There was an increase in the initial slopes of both the first agonist and first antagonist bursts as well as a gradual evolution of the second agonist burst. These findings are very similar to the usual changes in myoelectric activity associated with increases in movement velocity (Mustard and Lee, 1987; Corcos, Gottlieb and Agarwal, accepted pending revision). We have suggested that changes in EMG slope can be interpreted as evidence for modulation of the height of a hypothetical "excitation" pulse (Gottlieb, Corcos and Agarwal, in press). However, there was also a prolongation of the first agonist EMG burst and a decrease in the amount of co-contraction. This implies that the change is not limited to a simple increase in pulse height. Temporal changes in the excitation pulse must also have occurred.

We conclude that even the simplest movements are associated with significant improvements in performance and concomitant changes in the underlying neural commands. This process is associated with both increases in height modulation as well as changes in the underlying temporal structure.

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REGULATION OF MUSCLE FORCE AND STIFFNESS DURING LONG JUMP TAKE-OFF.

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Good long jump performance is characterized by minimum lowering of the center of the gravity during the impact phase of the take-off (Bosco, Luhtanen, Komi, 1976). Thus the tolerance to high stress loading must be achieved by high muscular forces together with high muscle stiffness (Kyröläinen, Avela, Komi, Gollhofer, 1989). The purpose of the present study was to combine external and internal factors in the complex movement. Therefore the main focus was directed the relationship between external impact forces, directly measured tendon forces and muscle activation patterns in long jump take-off.

METHODS

A 33-year former triple jumper volunteered as a subject. He was fully informed of the risks of the experiment, which were approved by the ethical committee of the responsible hospital. The subject performed with maximal approach run of 30 meters two submaximal (5.20 ± 0.18 m) and two maximal jumps (6.53 ± 0.03 m) over a force platform (Kistler). Each jump was filmed with a Locam-51-0003 camera set to operate 100 frames $\times$ s^{-1} . Achilles tendon forces (ATF) were recorded by in-vivo technique (Komi et al, 1987). EMGs were recorded telemetrically (Medinik AB Model 1C-600-G) with surface electrodes (Beckman) from the mm. gastrocnemius (GA), tibialis anterior (TA) and vastus lateralis (VL). The signals of the EMGs, ground reaction forces and ATFs were simultaneously stored on a magnetic tape (Racal Store 14) for further analysis. The changes of the length of GA were determined by the model of Grieve (Pheasant, Cavanagh, 1978). The stiffness coefficients (the ratio between the changes of the ATF and the changes of the muscle segment length) were calculated. The two submaximal (medium) and two maximal (good) jumps were averaged with respect to the onset of the vertical ground reaction force. Peak angular velocities of the knee and ankle joints together with the stiffness coefficients and various EMG-parameters were calculated.

RESULTS

The maximum ATF was higher in the good jump (2044 N) as compared to the medium jump (1975 N). When the ATFs were normalized by the tendon crosssection area (31 mm^2) the respective values were 65.9 N $\times$ mm^{-2} and 63.7 N $\times$ mm^{-2} . Similar differences were also observed in the peak ground reaction forces. During the take-off the stiffness coefficient of the triceps surae muscle and achilles tendon was higher in the good jump than in the medium jump (figure 1). The respective coefficients in the steepest point of the curves were 10.7 for the good jump and 6.8 for the medium jump. The preactivities in the GA and VL were higher in the good jump than in the medium jump. During the impact phase of the take-off the peak angular velocities in these two jumps were 8.5 rad $\times$ s^{-1} vs 7.2 rad $\times$ s^{-1} in the knee joint and 10.8 rad $\times$ s^{-1} vs 8.4 rad $\times$ s^{-1} in the ankle joint, respectively. Correspondingly integrated EMG-activity of the GA was clearly higher in the good jump than in the medium jump. However, during the push-off phase of the take-off the respective values were vice versa.

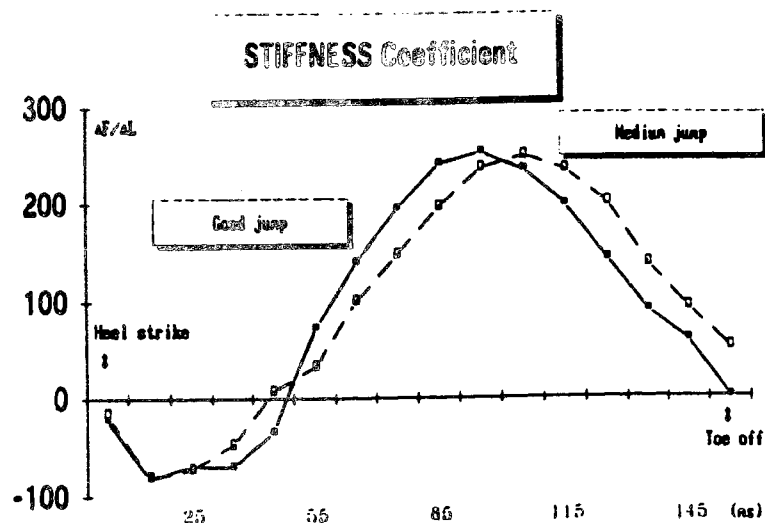


Figure 1. The curves of the stiffness coefficients of the m. triceps surae and the achilles tendon during the take-off contact in the long jump.

DISCUSSION

High preactivation in GA and VL muscles suggests that these muscles were well prepared to resist high impact forces. The fact that IEMG was also higher in the good jump during the eccentric phase supports the results of higher stiffness coefficients. High eccentric phase activation may also result from increased reflex activation, because the stretching velocity of the muscles was higher in the good jumps. Fast stretching accompanied with high muscle activation and improved stiffness prepare the muscle for better stretch-shortening cycle action resulting in effective recoil characteristics during the push-off phase. On the other hand, the finding of increased EMG-activity during the push-off phase suggests lowered recoil characteristics of the muscles in the medium jumps. The presented results support the concept that high tendomuscular stiffness is essential in the powerful long jump take-off. Quantitatively well coordinated activation patterns of the leg extensors during the preactivation and eccentric phase can be seen as the basis of adequate stiffness characteristics.

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THE RELATIONSHIP BETWEEN FORCE PRODUCED AGAINST OBJECTS OF DIFFERENT MOBILITY DURING STATIC AND DYNAMIC CONDITIONS

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Human upper extremity is actuated mainly by multiarticular muscles, whose contractions result not only in the desired motion, but also in additional movements in different joints. Stabilization of the redundant degrees of freedom is accomplished by stimulating mutually antagonistic muscles, whose moments of force are called the stabilizing functions. This process consists of imposing muscular constraints on temporarily uninvolved degrees of freedom within the musculo-skeletal system, and of reducing the excess mobility of external objects. It is known that the degrees of mobility influence the magnitude of applied force during static condition (Kornecki et al. 1983). The aim of this study, therefore, was to investigate the relationship between force produced against objects of different mobility during static and dynamic conditions.

Procedure

The first, static experiment was performed on a special apparatus with three replaceable elements, and was characterized by mobility of 0, 1z and 2yz degrees of freedom with respect to the immovable base (Fig.1). The employed technique measured the maximum force exerted by the upper extremity upon the external object of varying mobility: F_0 , F_{1z} and F_{2yz} . The second experiment, carried out under dynamic conditions, used a physical pendulum as the major part of the measuring apparatus (Fig.2). The mobility of the pendulum was 1y, 2xy and 3xyz degrees of freedom. Ten subjects were asked to apply maximal force to the pendulum to transmit as much kinetic energy as possible. The time histories of the forces, F_{1y} , F_{2xy} and F_{3xyz} , were obtained by calculating the second time derivative of the deflection angle considering the tangent component of the force of gravity. The moment of inertia of the pendulum was equal to 108.7 kgm².

Results and Conclusion

Analysis of correlation was performed in order to establish the relationship between the forces produced in different conditions against different mobilities. The correlation coefficients proved that there existed independently, a significant relationship between muscular forces produced against objects of different mobility during static as well as dynamic conditions (Tab.1). No relationship between force produced during static and dynamic condition was found. These results, however, are not definitive because of the nature of the apparatuses, where the precision of force application was as significant as its magnitude. Further investigation and analysis is in progress to determine the relationship between static and dynamic force when muscular stabilization is important.

Table 1. The matrix of correlation coefficients between parameters obtained during the first and the second experiment (n=10).

Parameter	F_0	Static F_{1z}	F_{2yz}	F_{1y}	Dynamic F_{2xy}	F_{3xyz}
F_0		0.642**	0.687**	0.091	0.155	0.199
F_{1z}			0.717**	-0.390	-0.321	-0.262
F_{2yz}				-0.100	-0.006	-0.011
F_{1y}					0.945**	0.978**
F_{2xy}						0.975**

** indicates the coefficients significant at $p=0.05$

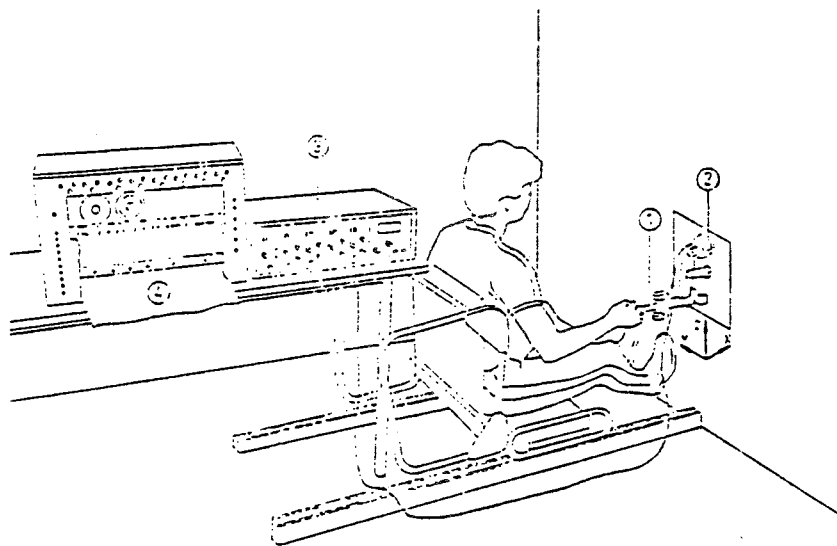


Figure 1. View of the measuring apparatus used in the first (static) experiment: 1) object of one degree of freedom in horizontal plane, 2) tensometric dynamometer, 3) amplifier, 4) recorder

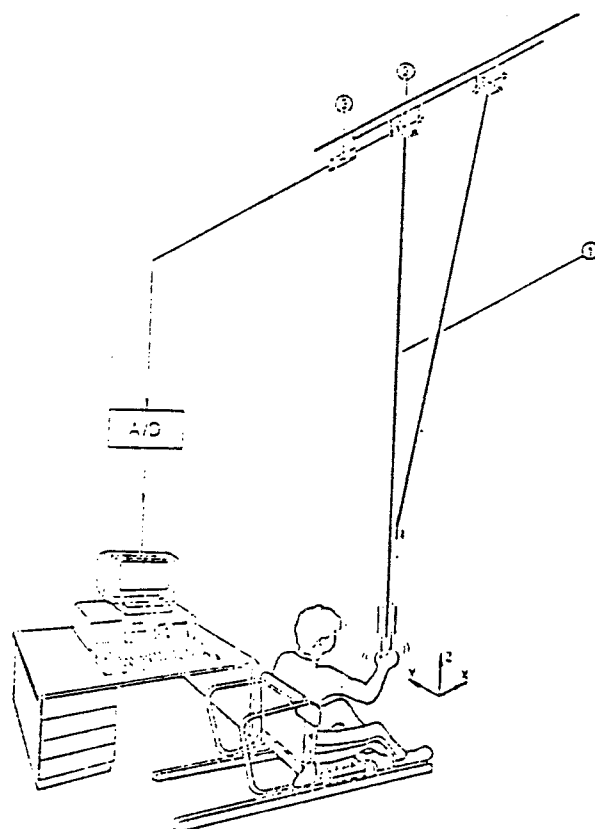


Figure 2. View of the measuring apparatus used in the second (dynamic) experiment: 1) pendulum of adjustable mobility, 2) Cardan joint, 3) potentiometer

Reference

Kornecki, S., Zawadzki, J. and Bober, T. (1981). Effect of forced stabilization on magnitude of revealed muscle force. In: A. Morecki et al. (eds). *Biomechanics VII-A* pp 107-113. Baltimore: University Park

INFLUENCE OF ECCENTRIC-CONCENTRIC MUSCULAR CONTRACTION UPON THE KIP-UP MANEUVER IN SKILLED AND UNSKILLED GYMNASTS

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The importance of the swing movement as an element of gymnastic technique can be explained in that it necessitates eccentric muscular contraction thus creating conditions for storage of potential elastic energy. Moreover, if the eccentric phase proceeds the concentric phase, the stored energy can be added to the muscles' metabolic energy. In this way the energetic efficiency of a given motor task can be improved. This effect is closely related to the sequence of eccentric and concentric contraction of various muscle groups.

The purpose of the present study was to determine how the cycle of eccentric-concentric muscle contraction affects the gymnastic swing movement in two gymnasts performing the kip-up maneuver on a horizontal bar.

Material and Procedure

Two gymnasts, a master (M) and a beginner (O), performed the kip-up maneuver on a horizontal bar. In order to calculate the actuating moments at the hip joint the inverse dynamic approach was employed with input data obtained from film analysis of the given movement (Siemienski et al., 1987). The contraction velocity of the following six muscles were calculated: rectus femoris, *Rf*, iliacus, *Il*, gluteus maximus, *Gmax*, semimembranosus, *Sm*, semitendinosus, *St* and biceps femoris caput longum, *Bfcl* and defined as the velocity at which the insertion of the given muscle approached its origin along its line of action. Shortening was defined as negative contraction velocity. This definition led to formulae in which the contraction velocity was dependent upon the angles and angular velocities at joints crossed by a given muscle. The time course of the actuating moment at the hip joint together with the time courses of the contraction velocities of the muscles enabled determination of the type of muscular contraction. The occurrence of the flexing moment indicated the contraction of flexors. If the contraction velocity was negative then the contraction is concentric. If the contraction velocity is positive, the contraction is eccentric.

Results and Conclusion

For the master gymnast (M) the following muscle coordination patterns were found: the flexors *Rf* and *Il* performed eccentric contraction, during the first phase of the exercise, then changed their action to concentric contraction, therefore flexing the lower limb in the hip joint. The flexion is braked by the eccentric contraction of *Gmax*, *Sm*, *St*, and *Bfcl* muscles. This brief braking action of the hip extension was followed by concentric contraction to extend the hip joint (Fig. 1). These data suggest that the technique of the master (M) was used to optimize the effect of swinging movement. The technique of the beginner (O) is different in that there was no eccentric-concentric contraction cycle of hip joint flexors. Such muscles' stretching-shortening work appeared only at the hip joint extensors, and the contraction velocity was almost equal to zero at that eccentric phase (Fig. 1). Therefore, this technique could not lead to effective utilization of the elastic energy of muscles during the swinging movement (Bober et.al. 1987).

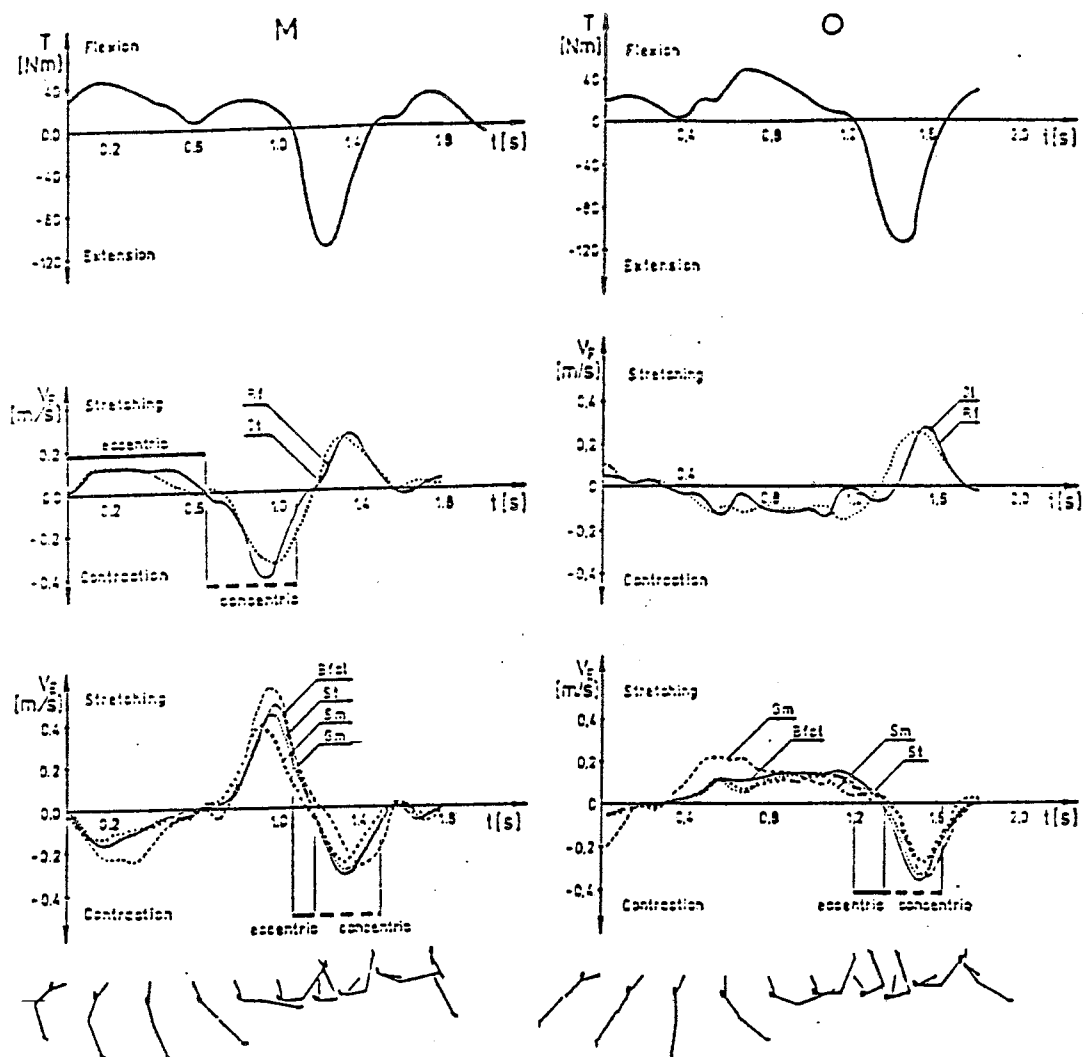


Figure 1. Actuating moment $T(t)$, contraction velocities of the flexors $V_F(t)$ and extensors $V_E(t)$ of the hip joint versus time t for the master class gymnast (M) and the beginner (O).

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INTERACTIVE EFFECTS OF NUTRITION, ENVIRONMENT, AND RAT-STRAIN ON CORTICAL AND VERTEBRAL BONE GEOMETRY AND BIOMECHANICS

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INTRODUCTION

Immature, rapidly-growing rats have been used to study the effects of spaceflight on bone biology, because the rapid growth and relatively short life-span of rats may accentuate the adaptations of bone during orbital flights of comparatively short durations, such as 1 to 3 wk. Different caging environments, diets, and rat strains, however, have been used in recent spaceflights with rodent experiments. As a consequence, the present experiment was designed to generate comparative data about the sensitivity of cortical and vertebral rat-bone adaptations of different rat strains to different cage environments and diets. The results of our comparative study showed significant main effects of these factors on the humerus and vertebral geometrical and mechanical properties, as well as significant interactive effects on the mechanical properties of the humerus.

MATERIALS AND METHODS

Cosmos biosatellite cages (65 x 20 x 16 cm) housed 10-rats per cage, while the USA Spacelab-3 (SL-3) cages (10.5 x 10.2 x 27 cm) housed only 1-rat per cage. Temperature for both caging arrangements was 22-24 °C, and the light-dark schedule was 16L:8D. The USSR paste diet (Cosmos 1514 and 1887) was the formula for vitamin supplementation of the paste diet. USA diet ingredients were obtained from Teklad Laboratories and Wayne Mills. Two different strains of rats have been used in Cosmos and SL-3 orbital flights: Czechoslovakian-Wistar (Czech) and Taconic Sprague-Dawley (Taconic) rats. All animals (48-day-old male albino, specific-pathogen-free) weighed approximately 200 g at the beginning of the test period (48 d). Animals were randomly assigned to one of eight experimental groups: (1) Cosmos cage (10-rats per cage)-USSR diet-10 Czech rats, (2) Cosmos cage (10-rats per cage)-USSR diet-10 Taconic rats, (3) Cosmos cage (10-rats per cage)-USA diet-10 Czech rats, (4) Cosmos cage (10-rats per cage)-USA diet-10 Taconic rats, (5) SL-3 cage (1-rat per cage)-USSR diet-5 Czech rats, (6) SL-3 cage (1-rat per cage)-USSR diet-5 Taconic rats, (7) SL-3 cage (1-rat per cage)-USA diet-5 Czech rats, (8) SL-3 cage (1-rat per cage)-USA diet-5 Taconic rats. The rats were maintained in their respective cages for 14 days, then killed by decapitation. Humeri and vertebrae were immediately dissected, moistened, and sealed in aluminum foil and plastic vials for freezing (-70 °C) until the day of testing. On the day of mechanical testing, humeri were equilibrated (37 °C) in a buffer solution (50 mM potassium phosphate, pH 7.4) for a minimum of 30 min prior to testing. Humeri were subjected to three-point bending loads until fracture with a servo-controlled electromechanical testing system (Instron 1122), using the same protocol as we used previously for humeri from SL-3 (3). While still wrapped, the T7 vertebrae were thawed to 7 °C for 16 h prior to the mechanical testing. On the day of testing, the body of each T7 was isolated from the intervertebral discs and vertebral pedicles, and using procedures that we have reported elsewhere (2), the vertebral bodies were compressed to a nominal 50% of specimen height at a fast strain rate (50%/s). Cross-sectional geometries were determined for both the humerus (3) and L6 (2).

RESULTS

After the 2-wk experiment, no significant differences were noted among body masses of the different groups when the factors of cage type or rat strain were compared. Similarly,

no differences were found for diet comparisons among the rats maintained in the 10-rat Cosmos cages. In the 1-rat SL-3 cages, however, Czech and Taconic rats on the USA diet were significantly heavier than the same strain of rats in the Cosmos cages. No significant differences were found among humeral lengths of the groups after the 2-wk experiment. The total (periosteal) and medullary cross-sectional areas, as well as periosteal and endosteal circumferences, however, were significantly greater for 1-rat cages (SL-3) than for 10-rat cages (Cosmos), but no differences existed between diets or between rat strains. Cage differences had significant effects on the mechanical properties of the humerus, with the 10-rat Cosmos caging producing generally stronger and stiffer humeri, regardless of diet or rat strain. The structural properties of maximum load and energy to maximum and failure loads were significantly different for the different diets; the USA diet produced greater maximum loads (16%), energy to maximum load (40%), and energy to failure load (41%). In contrast, for rat strain, only the flexural rigidity of the humerus was significantly different; the Czech rats had 22% stiffer humeri than the Taconic rats. Diet and rat strain had a statistically significant two-way interactive effect on linear tensile stress and energy to the proportional limit: the Czech rats favored the USA diet, and the Taconic rats favored the USSR diet. Rats housed in the 10-rat cages had significantly greater maximum stress and strain energy density at 50% strain than rats housed in the 1-rat cages. When rats were fed the USSR diet, the vertebral compressional stress at the proportional limit was significantly greater (28%) than if fed the USA diet. Czech T7 elastic modulus was significantly greater (29%) than the Taconic T7, and Czech T7 maximum stress was significantly greater (24%) and the strain energy density at 50% strain was significantly greater (25%) than the Taconic T7 values.

DISCUSSION

Both Taconic and Czech rats experienced a significant increase (>60%) in body mass during the ages of 48-62 d, but their rates of humeral longitudinal growth, during this period, were not affected by the different diets or caging conditions. Similar data for avians in different caging situations showed the rate of long bone growth to be relatively unaffected by the type of caging environment; whether chickens were cage or floor reared made no difference in bone growth rates (1). In contrast, after a 1-wk spaceflight, Shaw *et al.* (3) reported that humeral longitudinal growth was slowed, while in those same flight rats, tibial growth rate was not affected significantly. In general, the caging effect was the most potent influence on the biomechanical properties of both humeri and vertebrae. The group caging with 10-rats/cage produced consistently stronger and stiffer humeral and vertebral bones. This caging effect may have been a consequence of increased social interaction, competition, and physical activity among the rats in the USSR cages. Interestingly, the structural properties of the vertebral bone T7 were generally unaffected by the caging or diet differences, but the T7 material properties were significantly altered by the caging, diet, and rat-strain factors. These three factors, therefore, had a marked effect on the *quality*, but not the *quantity* of the vertebral bone. The comparative data from the present experiment indicate that caution is warranted when composite data are correlated from various international spaceflights.

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EFFECTS OF A HIGH-FAT/HIGH-SUCROSE DIET ON CORTICAL BONE REMODELING AND BIOMECHANICS

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INTRODUCTION

A high-fat/high-sucrose diet (HFS) may have deleterious effects on bone. A high sucrose diet has been reported to cause hyperinsulinemia (Grimditch *et al.*, 1988), which may induce hypercalciuria by inhibition of renal tubular resorption of calcium (Holl *et al.*, 1987). When fat is added to a high sucrose diet, insulin resistance is exaggerated and glucose intolerance is induced. In addition, even though still controversial, most studies show that a high fat diet may decrease calcium absorption and utilization, which also may affect bone properties. The present study examined the effect of a HFS diet on the mechanical, geometrical, and histological characteristics of cortical bone.

METHODS

Female Sprague-Dawley rats (8 wk of age, 150-170 g) were divided randomly into two groups: one group (n=9) fed a low-fat complex carbohydrate (control) diet and the other (n=9) fed a HFS diet for 10 wk. At the end of 10 wk, all animals were sacrificed by decapitization. The metabolic energy of HFS was 40% from fat (lard) and 40% from sucrose. The right tibia and second metatarsus (MT) were tested in three-point bending to failure. Loads were applied at middiaphysis (medial-to-lateral for tibia, ventral-to-dorsal for MT) with a crosshead speed of 25.4 cm/min. A 1-mm middiaphyseal cross section was cut from contralateral bones for a geometrical analysis that used a modified version of SLICE (Nagurka and Hayes, 1980); another 4 mm section was cut for histological studies using glycol methacrylate embedding and 2% toluidine blue stain. Five regions of the tibial cross section (anterior, medial, post-medial, posterior, and lateral) and four regions (dorsal, medial, ventral, and lateral) were examined under high power (400X) field.

RESULTS

Mechanically, in tibia, the maximum loads of the HFS were significantly lower (16%) than the control. The energy absorption was generally lower in the HFS, but only the energy at maximum load of HFS was significantly lower (35%) than the control. The linear tensile stress of HFS was significantly lower than the control (9%). In metatarsus, HFS had significantly lower (26%) elastic modulus and linear tensile stress (23%) than the control.

Middiaphyseal cross-sectional geometry showed that there was no difference in tibial geometry between HFS and control. But in the metatarsal, HFS had larger structures. Both the total cross-sectional area and cortical cross-sectional area of HFS were significantly (15%) larger than control. The ventral-dorsal diameter and lateral-medial diameter of HFS were significantly greater than the control (7% and 9%, respectively). The dorsal cortical thickness was significantly greater in HFS.

Histology of tibial cross section showed that the total number of osteocytes and osteoclasts was significantly less than control, by 13% and 45%. But the porosity and number of haversian-systems-like (HS-like) structures were not significantly different. The histology of HFS MT was markedly different from control. The HFS MT had significantly fewer HS-like

structures (37%), osteocytes (10%), and osteoclasts (46%). The percentage of inert area was larger (23%) in HFS MT. The dorsal and medial regions of MT were the most markedly different between the two groups.

Table 1. Bone mechanical properties

	C-tibia	HFS-tibia	C-MT	HFS-MT
Maximum load (N)	128 ± 16	107 ± 12*	44 ± 6	43 ± 6
Failure energy (Ns)	106 ± 19	69 ± 18*	14 ± 4	12 ± 3
Linear stress (MPa)	295 ± 18	270 ± 30*	443 ± 54	341 ± 37*
Elastic modulus(GPa)	11.2 ± 2.5	11.3 ± 1.8	10.0 ± 1.6	7.4 ± 1.2*

*significantly different from control ($p \leq 0.05$)

DISCUSSION

A HFS diet resulted in significantly different mechanical and geometrical properties of rat cortical bone. The histological data also showed significant differences in bone remodeling. The mechanisms for HFS effects on cortical bone may have been due to hyperinsulinemia and glucose intolerance and/or changes in the ability to absorb and utilize calcium. In addition, our data showed that more-distal bone (MT) was more sensitive to change in diet than more-proximal bone (tibia), consistent with the effects that diabetes has on distal and proximal bones.

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POSTURE-RELATED MODULATIONS OF LIMB DYNAMICS DURING SPONTANEOUS KICKING OF INFANTS

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INTRODUCTION

Spontaneous kicking movements in the first few months of life consist of coordinated flexions and extensions of the hip, knee, and ankle joints. Although the kicking motion appears to be stereotyped and reflex-like, in our study we show that kicking in even very young infants is modulated by a complex interplay between active and passive forces with a sensitivity to both changes in gravity and the movements of the limb's segments due to varying postures.

The trajectory of any limb movement is determined not only by the forces generated by active muscle contractions, but also by passive forces resulting from gravity and from the movements of other segments in the linked system of the human body. In skilled movements of adult humans, it has been shown that torques due to passive forces play a significant role in the control of movement trajectories and that active muscle actions counteract nonmuscular forces (Schneider *et al.*, 1986, 1987, in press). Is the ability to detect, adapt to, and control passive forces an early property of the central nervous system, or does it develop with experience? We began to answer that question by using the methods of inverse dynamics to analyze joint torques during spontaneous kicking movements (Schneider *et al.*, in press), and here we specifically examine how the control of passive forces may vary with a change in an infant's posture (supine, 45°, versus vertical).

METHODS

Infants were placed in three different postures (supine, 45°, versus vertical orientation of the torso), and movements were recorded using a WATSMART optoelectronic analysis system. The limb segments' masses, center of mass locations, and moments of inertia were determined with a mathematical model of the human body (Hatze, 1980; Schneider *et al.*, in press) which we modified for use with infants. The data from six infants (age = 0.23 ± 0.03 years, number of kicks = 42) were analyzed.

RESULTS

In all postures, throughout the motion, the muscle torque was modulated continually in response to the changing effects of gravity and the movements of the

thigh, lower leg, and foot. For example, the kicking motion was initiated by a large flexor muscle torque at the hip joint, and the muscle torque counteracted torque components due to gravity and the thigh and leg angular accelerations that tended to extend the lower extremity. As the kicking motion reversed from flexion to extension, again the muscle torque counterbalanced the torque components due to gravity and thigh and leg angular accelerations, thus, slowing the extension of the limb. Similar to the situation at the hip joint, at the knee and ankle joints the muscle torque component was counteracted by a gravitational torque component and motion-dependent torque components.

DISCUSSION

Our results revealed that underlying this organized but simple movement is a complex interplay of muscular actions in response to nonmuscular forces. As a result of the change in posture, the infant's leg was subjected to changing torque influences due to gravity and the motions of linked segments. Nonetheless, the torque profiles seen in our study indicate that the neuromuscular system adjusts to the differences in posture. The varying nonmuscular forces were detected and compensated, and thus, a kicking motion of high stability was produced regardless the infant's posture. These results suggest that mechanisms exist early in development by which the neuromuscular system responds to a changing environment, and that the neuromuscular system has self-organizing abilities to account for that changing environment.

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EFFECTS OF CONTINUOUS PASSIVE MOTION ON GEOMETRICAL AND MECHANICAL PROPERTIES OF LONG BONE

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INTRODUCTION

Continuous passive motion (CPM) has been used widely in clinical-patient care to prevent detrimental effects of immobilization when active motion must be restricted following injury. CPM effects on cartilage (7), tendon (3), ligament (4) and joint stiffness (2) have been documented, but its effects on bone have not been studied. Since trauma of the lower extremity may require internal fixation of bone as CPM is applied to the joint, the present study investigated the influences of ankle-joint CPM on geometry and mechanical properties of intramedullary-pinned rabbit tibia after deprivation of weightbearing forces.

MATERIAL AND METHOD

Ten female New Zealand white rabbits received CPM to one ankle and immobilization to the other, while the knee joints were fixed by intramedullary pins through femur and tibia (6). CPM delivered at a frequency of 300 cycles/day, through a range of motion of 90° to 150° plantar flexion. After treatment of 3 wk, rabbit were killed and tibias were harvested. Mechanical data were obtained through three-point-bending test to failure (Instron 1122) (5). Analog load-deflection data were digitized by IBM PC/AT and computerscope ISC-16 system, and structural and material bone properties were calculated. Geometrical properties were obtained by photographing and digitizing the 1-mm thick middiaphyseal transverse section of each tibia (Numonics 1220). Analysis of variance procedures (SAS, Cary, NC) and Ryan-Einot-Gabriel-Welsh multiple-range tests were used to reveal significant differences among main-effect means. A significance level of $p \leq 0.05$ was used for all statistical tests.

RESULTS

In geometrical properties, CPM-treated tibias had significantly greater middiaphyseal cortical cross-sectional area (>19%) and posterior cortical thickness (>37%) than immobilized (IMM) tibias. Other geometrical parameters, although not significantly different, showed a tendency of greater values in the CPM group. The material properties of CPM and IMM tibias were not different (Table 1). Structural properties, however, showed greater values to the CPM tibias; flexural rigidity (>9%), energy to failure (>12%), and nonlinear deformation (>19%) were statistically significant (Fig. 1).

DISCUSSION

In comparison to the IMM limb, CPM-treated tibias showed positive effects on the geometrical and mechanical properties of bone. During CPM treatment, the limbs were not weightbearing, and the principal forces applied to the tibias may have included local muscular forces produced from muscle contractions or passive stretching of musculotendinous units, and bending forces produced as the CPM apparatus moved the ankle into plantar flexion (150°). These forces would pull on the insertions to bone and also increase the bending and compressive strain on the middiaphysis. The posterior tibial cortex got significantly thicker during the CPM treatment. That result is consistent with previous work (1), showing that bone apposition occurs primarily on the concave compression surface of a bone. Our data showed that CPM can have significant and positive influences on the mechanical and geometrical properties of bone.

Table 1. Tibial Geometrical and Material Properties

	CPM (n=9)	IMM (n=10)
Geometric Properties:		
Medullary XSA (mm ²)	13.7 ± 2.4	14.0 ± 1.6
Cortical XSA (mm ²)	30.6 ± 2.8*	25.6 ± 1.3
Posterior Cortical Thickness (mm)	2.0 ± 0.6*	1.4 ± 0.1
2nd Moment of area (mm ⁴)	122.7 ± 20.7	106.4 ± 16.2
Material Properties:		
Tensile Yield Stress (MPa)	134.7 ± 24.9	136.1 ± 18.1
Elastic Modulus (GPa)	5.75 ± 1.12	5.58 ± 1.01

Values are means ± S.D., XSA= cross-sectional area

* CPM group is significantly greater than Immobilized group (p < 0.05)

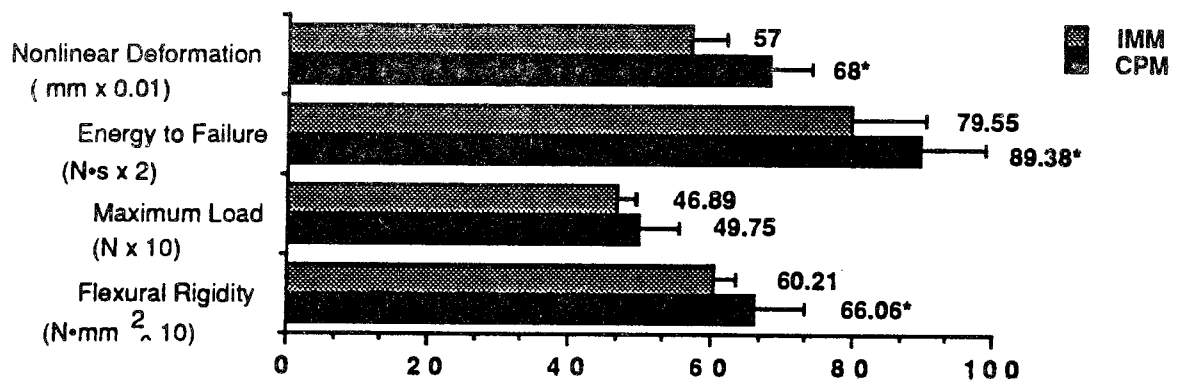


Figure 1. Tibial Structural Properties. (* CPM group mean is significantly greater than Immobilized group mean)

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SWING-PHASE DYNAMICS AND EMG DURING BACKWARD QUADRUPED WALKING

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INTRODUCTION

Grillner's (1981) unit-burst generator hypothesis suggests that backward (BWD) walking cats will recruit hip flexors with knee and ankle extensors in stance, and hip extensors with knee and ankle flexors in swing. Nevertheless, kinematics of BWD walking cats show that swing begins with flexion and ends with extension at the hip, knee, and ankle (not shown) joints (Fig. 1A). As in forward (FWD) walking, flexors (tibialis anterior, TA; iliopsoas, IP) are active early in swing, and extensors (lateral gastrocnemius, LG; vastus lateralis, VL; anterior biceps femoris, ABF) are not active until late swing (Fig. 1A; Smith *et al.*, 1988). Current results indicate that semitendinosus (ST) is active during late stance and throughout swing for BWD walking. In the present study, we use intersegmental dynamics in an attempt to explain swing-phase EMG at the hip and knee joints for BWD walking.

METHODS

Four cats were filmed with high-speed (100 frames/s) cinematography (Hoy & Zernicke, 1986) during BWD and FWD walking at 0.6 m/s on motorized treadmills. The three-segment hindlimb model and EMG recording and analysis techniques of Hoy *et al.* (1986) were used to calculate swing-phase hip and knee torques and to characterize muscle activities.

RESULTS

1. Knee (Fig. 1B). As the hip, knee, and ankle (not shown) joints flexed in early swing, torques related to the angular accelerations of the leg (LAA) and paw (PAA) segments tended to extend the knee. A flexor muscle torque opposed these interactive torques to produce a flexor net torque during the first half of swing, when ST was usually active (See LEGEND).

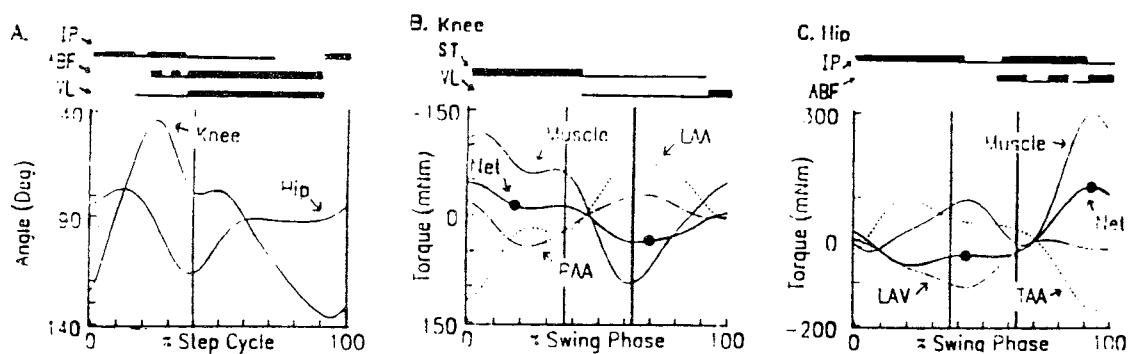


Fig. 1. Joint angles and EMG for one step (A) (paw off to paw off), and swing-phase torques and EMG at the knee (B) and hip (C) for the same step. Flexion is upward in A; Flexor torques are upward in B-C. The middle vertical line in A marks paw contact. Abbreviations and markings in B-C are explained in RESULTS. Data are from one cat, except ST EMG which is data from another cat.

After the hip reversed from flexion to extension (thin vertical line), torques related to LAA and PAA tended to flex the knee. An extensor muscle torque opposed these same interactive torques to produce an extensor net torque during the end of swing, when VL was active.

2. Hip (Fig. 1C). The hip was flexing when swing began, and an extensor net torque during early swing reversed the hip from flexion to extension. A flexor muscle torque (preceded by IP EMG) counterbalanced an extensor torque due to the angular velocity of the leg (LAV), so that these two components canceled each other in early swing. Extensor torques related to gravity and hip linear acceleration (not shown) were partly counterbalanced by torques related to the angular accelerations of all three limb segments (thigh (TAA), leg & paw not shown), so the net torque was extensor in early swing. These counterbalancing relationships changed after the knee reversed from flexion to extension (thick vertical line). While other torque components canceled each other, a flexor muscle torque opposed an extensor torque related to TAA, so the net torque was flexor in late swing, when IP and ABF (and ST) were active. This flexor net torque slowed hip extension and initiated hip flexion just prior to paw contact (Fig. 1A).

DISCUSSION

Knee-muscle torques opposed torques related to the angular accelerations of the leg and paw throughout BWD swing. Knee net and muscle torques were in phase: flexor with semitendinosus EMG in early swing, and extensor with vastus lateralis EMG in late swing. Two periods of flexor hip-muscle torque, each preceded by iliopsoas EMG, opposed torques related to the leg angular velocity in early swing, and the thigh angular acceleration in late swing. Hip net torque was extensor in early swing due to gravity and hip linear acceleration, and flexor in late swing due to the muscle torque. During slow FWD walking, in contrast, there are two periods of knee-muscle flexor torque in swing, while hip-muscle torque is flexor early and extensor late in swing (Smith and Zernicke, 1987). We propose that the general EMG pattern of flexor activity in swing and extensor activity in stance used for forward walking is retained for BWD walking, and that information from afferent feedback may be used to adapt details of the recruitment pattern to the different biomechanical demands for BWD vs FWD walking. Supported by NIH NS19864 and a UCLA Biomedical Research Support Grant.

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GRIP STRENGTH AND LEG DESIGN OF POWER GRIP HAND TOOLS

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In two legged hand tools designed for power grip, a maximization of the squeezing capacity of the hand is important in order to reduce muscle strain during work. The aim of this study was to compare the traditional grip with a "reversed" (see below), as well as to evaluate the significance of grip span of the individual fingers and the entire hand in relation to force contribution from each finger and total force between the jaws of the tool.

Material and Methods

Sixteen healthy right-handed volunteers participated, 8 women and 8 men (18-60 y, ordinary stature and body weight). An estimate of the subjects' hand size was calculated as the sum of palmar hand length, metacarpal breadth and mean finger length. A modified pair of slip joint pliers with force transducers were used to measure the force under each finger (finger force) as well as the resultant force (R.F.) between the jaws of the tool. The distance between the legs of the tool was altered in steps of 1 cm, ranging from 5 to 10 cm for each finger. Thus, each finger grip span corresponded to a specific hand grip span which was measured at the centre of the hand. The finger force and the R.F. were measured simultaneously. The traditional power grip was compared to a "reversed" grip with the little finger (V) closest to the tool axis of rotation. The subjects were standing in a standardized position and they were instructed to perform a maximal squeeze of the tool handles with their right hand. They did not receive any visual feed-back of their performance. All force measurements were repeated on a different day. The "optimal span" was defined as the span for which the peak force was obtained.

Results

Males obtained their maximal R.F. at hand grip spans about 6-7 cm, while the corresponding value for females was about 5-6 cm (fig.1). The curves could be expressed by 3-grade-polynomials (Pearson: $r > 0.91$, $p < 0.001$). In average, the traditional grip implied a higher R.F. and a more narrow optimal hand span compared to the reversed grip. For hand grip spans wider than 6 cm, the R.F. was reduced by 10.0 % (S.D.: 2.3) per cm increase in hand grip span.

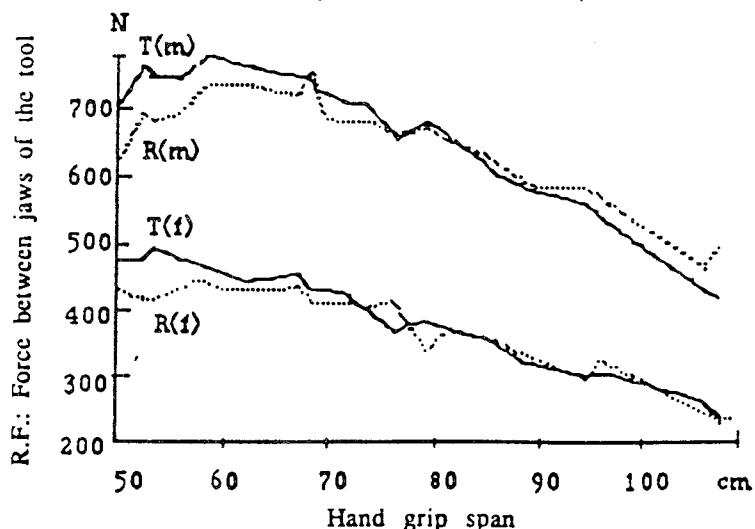


Fig. 1. T: traditional grip, R: reversed grip; m: male, f: female

Fingers (II+III) were stronger than fingers (IV+V) (fig.2). Fingers II-IV had their optimal grip span at around 5-7 cm, while it was 5 cm or lower for finger V.

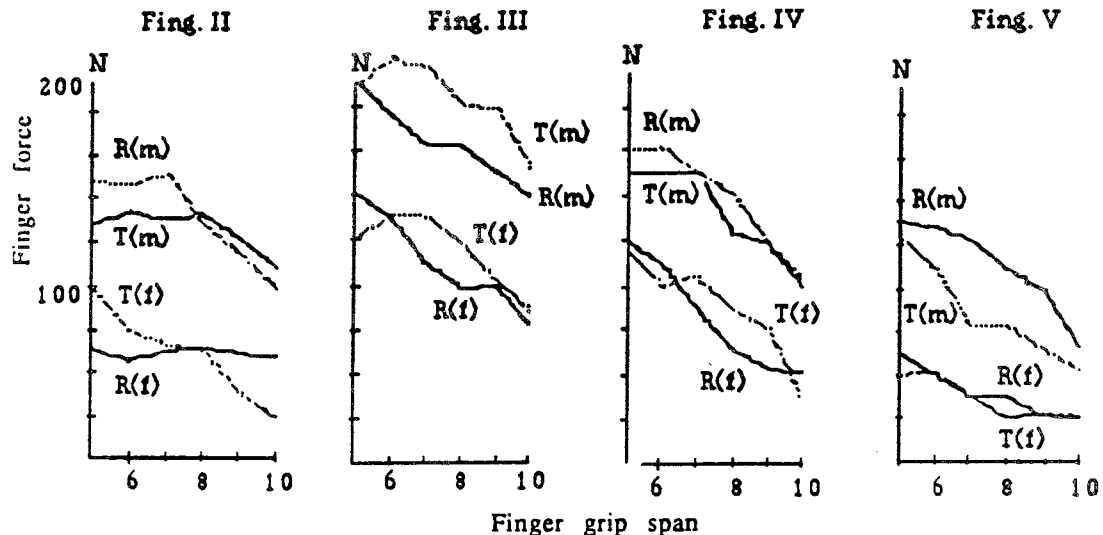


Fig.2.: T: traditional grip, R: reversed grip; m: male, f: female

The R.F. as well as the finger strength were significantly correlated to hand size (Pearson: $r=0.87$ and $r=0.97$ resp., $p<0.001$). The R.F. showed no sex difference after correction for differences in hand size.

Discussion and conclusions

The strength ranking order between the fingers (III>IV>II>V) agreed with previous studies (Hazelton 1975, Ohtsuki 1981). The presented optimal hand grip span of 5-7 cm seems to agree with a previous study by Petrofsky (1980), but it is obviously more narrow than that found by Greenberg & Chaffin (1976).

The idea behind introducing the reversed grip was to provide the stronger half of the hand (fingers II+III) with the longest set of lever arms in order to increase the R.F. However, the peak values of the R.F.s were obtained with the traditional grip. The reason for this might be that the subjects were unfamiliar with the new type of grip or that the reversed grip provided a more unfavorable interrelationship between the fingers.

1. The results of this study suggest that a power grip tool provides the highest output (the highest R.F.) when using:

i) the traditional grip ii) a hand grip span around 5-7 cm iii) a more narrow span for finger V than the rest of the fingers.

2. In addition, the results indicate that for hand grip spans wider than 6 cm, the R.F. is reduced by about 10% per cm increase in hand grip span.

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ORTHOPEDIC SHOES FOR CHILDREN WITH CEREBRAL PALSY
IN RESIDUAL STAGE AND WITH "PES EQUINO-VARUS CON-
GENITAL" WHICH IS FEEBLY MARKED OR WAS SURGERYLY
TREATED

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A biomechanical investigations were carried out which revealed three main functions inherent of the foot: an elastic spreading under load (spring function), balancing function and imparting of acceleration to the centre of masses during walking (propulsive function). It was also established that the determinative moment for the indications for the use of orthopedic techniques means is a degree of disturbance of these foot functions but not morphological changes corresponding with the every type of pathology. In this connection the use of one-type devices are possible in different deformities. So, the biomechanical analysis of the foot functions disturbances by cerebral palsy and pes equino-varus congenital enables us to design the orthopedic shoes which may be produced depersonally and then adapted to individual requirements. This is achieved by using adjustable straps and different material modules to provide individual foot correction and to create a dynamic loading area during walking.

The curvature of sole is calculated to minimize external torques which increase the pathology foot position (Table 1): curvature centers are coincided with rotation centers of knee and ankle (Fig.1). The rigidity of the high boot quarter can be adjusted by insertable flat springs (Fig.2) (Mendelewich, Pitkin, 1987).

Table 1. Prescription of new orthopedic shoes

Indication 1	Using variant 2
Drop foot. Normal position of the forefoot.	Shoe without adjustable straps (Fig.3.1) ore with only middle one for heel position fixating (Fig.3.2).
Drop foot with varus of the forefoot.	Middle strap combined with strap from upper-inner edge of the quarter up to the 4th metatarsal head (Fig.3.3).
Drop foot with valgus of the forefoot.	Middle strap and strap from the 1st metatarsal head up to upper-outer edge of the quarter (Fig.3.4)

1	2
Drop foot. Pathology position of the forefoot and hypertone of the m.gastrocnemius.	Middle strap and both cross straps with individual adjustment (Fig.3.5).

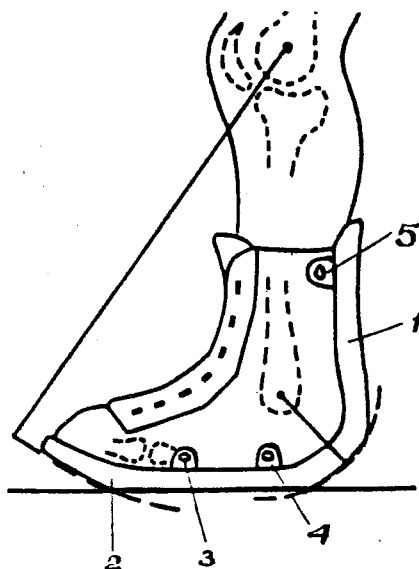


Fig.1. The sole curvatures.

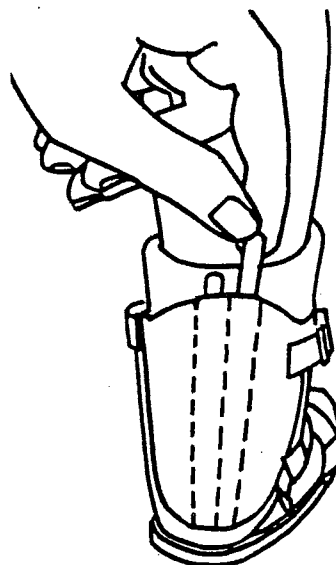


Fig.2. The spring adjustment of the quarter rigidity.

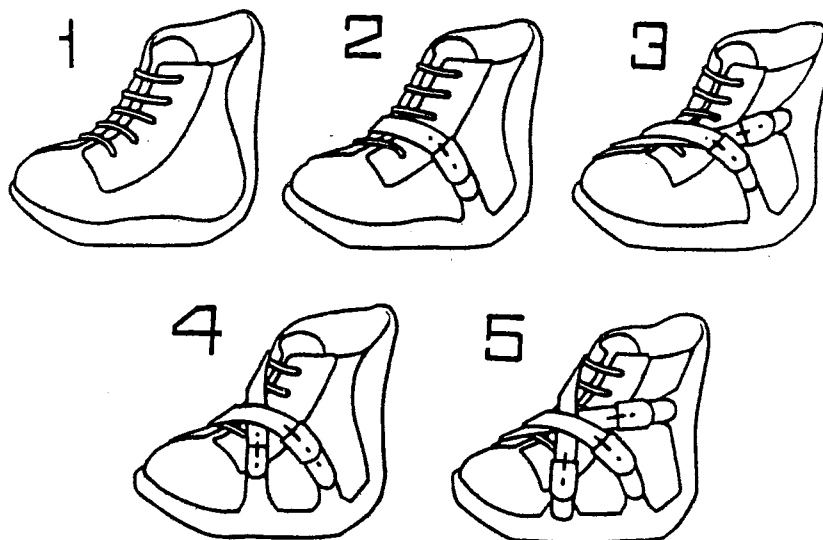


Fig.3. Using variants of the presented shoes.

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A KINEMATIC MODEL OF THE HUMAN HAND TO PREDICT ITS PREHENSILE CAPABILITIES

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INTRODUCTION

A biomechanical model is being developed to predict the prehensile posture of the human hand. This model is formulated from kinematic considerations and uses ellipsoids to approximate the geometry of the hand segments. The model predicts the hand posture needed for power grasp of ellipsoidal objects of various sizes. Hand posture is quantified via the joint angles. Contact points and vectors along the hand-object interface are calculated. The model will be used to evaluate tool designs so they may be used efficiently without adverse effects due to mechanical trauma.

METHODOLOGY

The kinematic skeleton of the hand is characterized by ideal joints connected by simple segments. The kinematic skeletal system is just a functional mathematical equivalent of the physiological skeletal system. For each segment, a local coordinate system is defined such that the proximal joint is the origin and the x-axis is defined as the vector from the proximal to the distal joint. The y-axis is projected dorsally perpendicular to the anterior-posterior plane for each segment, with the z-axis defined by the right-hand rule. For the root coordinate system, located at the wrist joint, the x-axis will be directed distally parallel to the third metacarpal. Finger-joint angulation is characterized by yaw (abduction-adduction), pitch (flexion-extension) and roll (axial rotation) angles. Movements can then be transferred proximally down the hand by homogeneous matrix multiplication. The neutral angular position of the thumb CMC joint is defined from the data of Cooney (1981), i.e. the reference axes of the trapezium are flexed 48° , abducted 38° and pronated 80° with respect to the third metacarpal.

The three-dimensional geometry of the surface of the hand is approximated by ellipsoids attached to each hand segment. These ellipsoids are used for determining contact points and therefore are called segment contact bodies. Objects are also described using ellipsoids, cylinders are established as ellipsoids with an infinite long axis that is truncated. Contact is determined mathematically based on the geometrical relationship between two ellipsoids that just touch in a single point. That is the outward normal vectors from the two ellipsoids are parallel but in opposite directions. The ellipsoids are expanded or contracted until a single point of contact is determined. If contraction was necessary, then the ellipsoids are in contact and soft tissue deformation may be estimated from the amount of contraction. This algorithm was developed by Fleck and Butler (1981).

The model functions by reducing the posture prediction problem to a sequence of one degree of freedom problems. Hand posture is estimated by varying a specified joint angle until a specified hand segment contacts the object and the soft tissue deformation criterion is met. Consider a circular cylinder held in a power grasp as a simple example. The initial posture has the fingers extended and the MCP joints have a neutral abduction-adduction angle. The thumb joints may either be abducted, to contribute power to the grasp, or adducted, for an element of precision. The exact thumb abduction-adduction angles are somewhat arbitrary. The CMC joint of the thumb also pronates considerably when abducted. The cylinder is then located in the crotch between the thumb and index finger by the contact algorithm. The model flexes the MCP joints of digits II to V, until the proximal phalanges contact the cylinder surface. The model proceeds distally, flexing the PIP joints until the middle phalanx contacts the object, then the DIP

joints are flexed until the distal phalange contacts the object. The joints of the thumb follow in a similar manner. The CMC joint moves in opposition until the metacarpal touches the object followed by flexion of the MCP and IP joints until the proximal and distal phalanges make contact, respectively.

This algorithm for power grasp of ellipsoidal objects is implemented on a microcomputer using Pascal. Coefficients for determining segment lengths, breadths and depths from hand length and breadth measurements are incorporated in the model, so that the prehensile capabilities of any size hand may be analyzed by the model. Ellipsoidal semi-axes lengths for describing the object and object orientation angles are needed as model input. Graphics procedures are included for visual display of the model. Hidden line routines are included, but these do increase processing time. Examples of the graphical display of the model are shown in Figures 1 and 2.

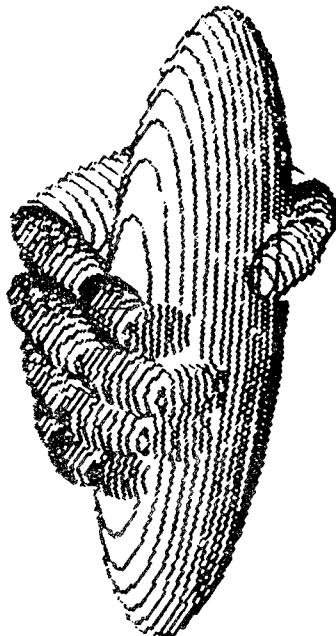


Figure 1. Graphical depiction of the hand posture predicted by the model for power grasp of a circular cylinder (Example 1).

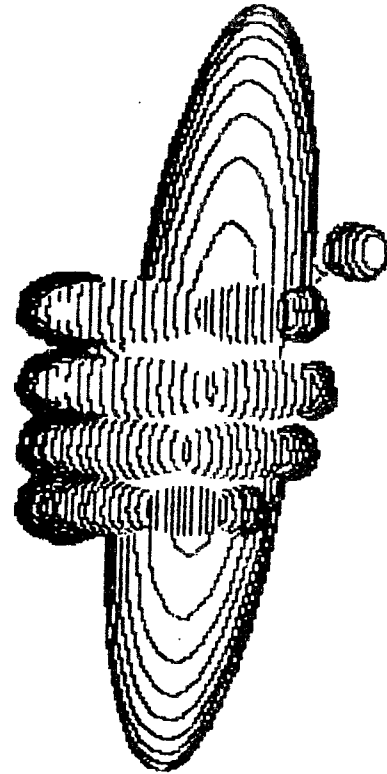


Figure 2. Graphical depiction of the hand posture predicted by the model for power grasp of a circular cylinder (Example 2).

SUMMARY

This biomechanical hand model provides an accurate description of power grip function. In this way, it will aid in the design of hand tools for both performance and safety. Algorithms for other more complex grasps are being examined. With further development, this model also will estimate the reaction forces on the hand-object interface using physiological tendon loading conditions. A pilot study using a planar kinetic model has been tested.

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REHABILITATION MECHANISMS OF CHILDREN WITH CEREBRAL PARALYSIS BY SIMULTANEOUS EFFECT OF ELECTRIC CURRENTS AND MAGNETIC FIELD ON CHILD NEUROMOTIVE SYSTEM

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In morbidity rate children's cerebral paralyzes (CCP) take second place after inborn deformities constituting 1.7 % of children's orthopedic diseases. To rehabilitate such patients numerous conservative, surgical and combined treatment methods have been developed and are widely used. However, their final effect does not guarantee patients appreciable result. Therefore, they are forced to use various prosthetic orthopedic appliances helping them to perform elementary motive functions and to attend to themselves.

As is seen from the comparative analysis of the well-known treatment methods their general shortcoming is in the symptomatic character of effects exerted on one of links (central or peripheral) of neuromotive system. These effects do not take into consideration either relation between central and peripheral links or possible response from the neuromotive system and the brain as a whole.

Our studies on the mechanism of simultaneous effect of electric currents (EC) and magnetic field (MF) on both links of the neuromotive system of patients with CCP have shown that these factors under certain parameter values admit controlling organism responses. As a result rehabilitation therapeutic effect is achieved by patients through proper organism efforts forced in the response development process to utilize all its reserves for forming functional reflex connections. As their formation under neuromotive system central link pathology is impossible without preliminarily mobilising brain reserves the effect of EC and MF is directed first at organism response formation providing conditions for functional activity of preserved or reversibly changed neurons and then at their optimal use.

To develop the necessary response and to regulate its formation process we obtained analytic dependences for setting EC and MF parameters permitting us to exert discrete individually balanced effects on the neuromotive system central link. Depending on the organism response character simultaneous balanced effects of EC from periphery to centres admit the formation of reflex connections and the regulation through this of the process of developing functional locomotions for patients with CCP. Setting EC and MF parameters is based on white blood formula studies according to our analytic expressions.

Except white blood formula studies showing organism response type patients' condition was assessed by clinic and instrumental methods. Thrice a week biochemical studies were carried out.

On the whole research results showed that during first

two weeks strengthened muscle hypertension and hyperreflexion obstructing performance of even those locomotions which were accessible for them. Exacerbation peak for such processes was observed as a rule by the end of the fortnight. Then gradually by the 3d midweek and discontinuously by the 4th week end substantial improvement in patient condition begins.

Among instrumental methods we particularly stress EEGraphic studies having characteristic dynamics. Thus, in the patient treatment process (1-2 weeks) slow irregular wave like forms in all branchings were replaced by quick rhythms, muscle potentials emergence and by convulsive components. By the 3d week end all branchings are characterized by emerging and later by predominating basic spontaneous rhythms typical for a given child age and by disappearance of convulsive components and motive artefacts.

Such regularity in the EEGraphic changes dynamics shows that during functional locomotion formation process by simultaneous effect of EC and MF on both links of the neuromotive system we observe first the destruction of initial components of synchronous and synphase activity and then the emergence of new complexes with heterochronously working brain cortex. Considering that in the base of rhythmic changes in the central nervous system taking place in the process of dynamic stereotype formation lies the formation of the conditioned reflex the diagram reflecting its dynamics is factually the motion portrait or image.

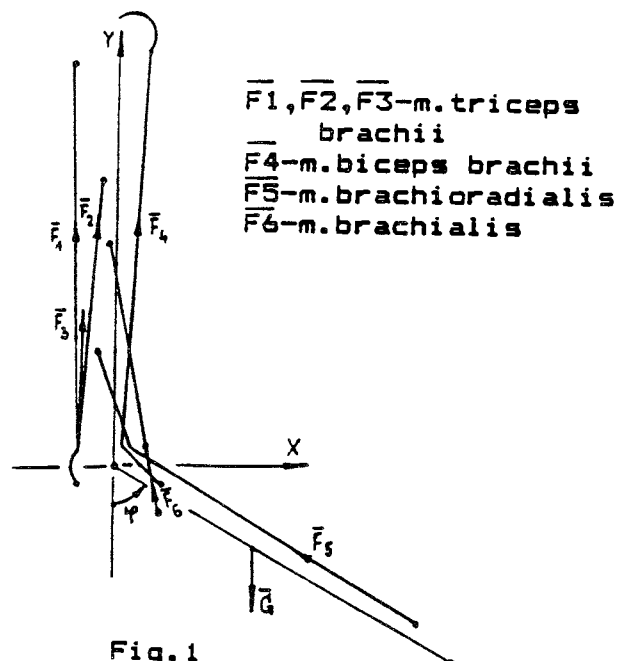
The method and devices for effecting EC and MF on central and peripheral links of the neuromotive system were thoroughly clinically checked and approved by the USSR Health Ministry.

SEGMENT-LINES MODELING OF HUMAN ELBOW JOINT DRIVE

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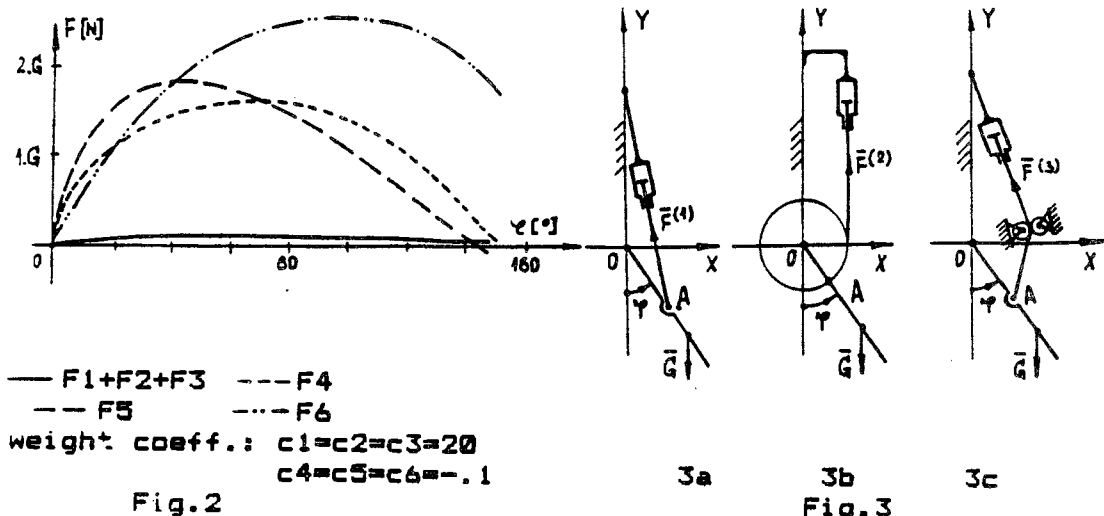
INTRODUCTION. Two basic problems are in the focus of attention when modeling the human joints - a good reflection of the real structures and functions and a suitable choice of the mathematical approach. The free body diagram usually contains straight-lines connecting with the muscle origins and insertions and presenting the directions of muscle forces [1]. As it is shown in [2] such approach should be treated with caution. The centroid lines describe well the muscle action lines, but this method is related with different difficulties [3]. For calculation of the muscle forces the equation of equilibrium or of motion are used. As the number of unknown forces exceeds the number of equations, the problem has to be solved using functional grouping, systematic elimination or optimization method. The linear programming method with various objective functions is often used with different success [4]. The aim of the paper is to describe more adequately the muscle action lines bearing a relation to the elbow joint during flexion-extension and using the Lagrange multipliers method to calculate the muscle forces. The proposed segment-lines modelling is compared with straight-lines modelling on the base of three different rotation drive gears.

TWO-DIMENSIONAL MODEL. The two-dimensional model of elbow



flexion-extension presented on Fig. 1 contains four segment-lines modeled muscles. Basic conditions: the moment of every muscle to keep permanent direction for the whole motion range; the moments of all synergistic muscles to have identical direction and opposite to the direction of antagonistic muscle moments. The Lagrange multipliers method with objective function $Z = \sum c_i |\vec{F}_i|^2$ is applied for muscle force calculation. The model is analyzed under quasistatical (Fig. 2) and dynamical conditions for different values of the weight coefficients. This method needs

simple software, the force solutions are of analytic type, the coefficients c_i are independent of the model geometrical parameters and the force values are guaranteed to be positive.



COMPARISON OF THREE MODELS OF ROTATION. A comparison is done between straight-lines modeling (Fig.3a) and segment-lines modeling (Fig.3b-m.triceps brachii, Fig.3c-m.biceps brachii). The geometrical and dynamical characteristics of the three models are identical with those of the model on Fig.1. The forces necessary for equilibrium supporting or for motion performing are calculated and compared with for estimation of the rotation models from power point of view. The motion law for the links OA (Fig.3) is $V^i = A^i \sin(\frac{\pi t}{T})$ $i=1,2,3$; $\gamma=0[^\circ]$ when $t=0[s]$; $\gamma=180[^\circ]$ when $t=T[s]$. Under quasistatic conditions the force $F^2(\gamma)$ is the most uniform and the force $F^1(\gamma)$ has the smallest maximum. Under dynamic conditions the force $F^1(\gamma)$ is the smallest for big $\ddot{\gamma}(t)$ and $\gamma > 0$. When $\ddot{\gamma}(t)$ is near zero - $F^3(\gamma) < F^1(\gamma)$. The model on Fig.3b can perform only a part of the given motion. The proposed approach allows to analyze the muscle cooperation during different motions and to use some biomechanical structures and functions for technical system design.

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A NEW APPROACH TO IDENTIFY KINEMATIC PECULIARITIES IN HUMAN MOTION

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1-INTRODUCTION. The procedure presented below, which is based on a new methodology recently developed by the authors [1] to study kinematic chains of rigid bodies, allows an easy identification of some kinematic peculiarities of the human body motion.

2-METHOD. Our method, beside the well-known homogeneous transformation matrices that we call *position matrices* M , introduces other 4x4 matrices: a) the *speed matrix* W containing both the speed and the angular speed of a body; b) the *acceleration matrix* H , containing both the acceleration and the angular acceleration; c) the *generalized speed ratio matrix* L , describing the Instantaneous Screw Axis (ISA) between two bodies. These matrices allow us to handle linear and angular quantities simultaneously, simplifying the usual kinematic relations. Moreover, since these matrices are cartesian representations of tensors, this approach allows the preliminary development and the symbolic handling of the kinematic relations out of any reference frame. The mentioned matrices are

$$W_{ij}(k) = \begin{bmatrix} 0 & -\omega_z & \omega_y & v_x \\ \omega_z & 0 & -\omega_x & v_y \\ -\omega_y & \omega_x & 0 & v_z \\ 0 & 0 & 0 & 0 \end{bmatrix} = \begin{bmatrix} \omega & V_0 \\ 0 & 0 & 0 & 0 \end{bmatrix} \quad \begin{aligned} U &= [u_x \ u_y \ u_z]^t \\ T &= [t_x \ t_y \ t_z]^t \\ B &= -U P + p U \end{aligned}$$

$$L_{ij}(k) = \begin{bmatrix} 0 & -u_z & u_y & b_x \\ u_z & 0 & -u_x & b_y \\ -u_y & u_x & 0 & b_z \\ 0 & 0 & 0 & 0 \end{bmatrix} = \begin{bmatrix} U & B \\ 0 & 0 & 0 & 0 \end{bmatrix} \quad L_{ij}(k) = \begin{cases} W_{ij}(k)/|\omega| & \text{if } \omega \neq 0 \\ W_{ij}(k)/|V_0| & \text{if } \omega = 0 \end{cases}$$

ω is the relative angular velocity of the bodies i and j , V is the relative velocity of two particular points [1] of the two bodies, U and P are the unit vector and a point of the ISA respectively; finally p is the pitch of the screw. All the quantities are expressed with respect to the frame (k) . The matrix H , which represents the relative acceleration of two bodies, is obtained by adding the square of W to its time derivative W' . Speed and acceleration matrices are related to the position matrix, as well as to its first and second time derivatives M' , M'' by the relations [1,2]

$$(1) \quad \begin{cases} W_{ij}(1) = M'_{ij} M_{ij}^{-1} \\ H_{ij}(1) = M''_{ij} M_{ij}^{-1} \end{cases} \quad H_{ij}(1) = W'_{ij}(1) + W_{ij}^2(1) = \begin{bmatrix} G & A_0 \\ 0 & 0 & 0 & 0 \end{bmatrix}$$

The following formulas show the relations linking the representations of the above mentioned quantities W , L and H into different frames $(r$ and $s)$

$$W_{ij}(r) = M_{rs} W_{ij}(s) M_{rs}^{-1} \quad L_{ij}(r) = M_{rs} L_{ij}(s) M_{rs}^{-1} \quad H_{ij}(r) = M_{rs} H_{ij}(s) M_{rs}^{-1}$$

3-APPLICATIONS. The following examples show how our method can be easily applied to practical biomechanical problems using (for instance) *space_lib*, a

software library, described in [5].

3.1-VELOCITY AND ACCELERATION ANALYSIS. Let us consider the kinematic analysis of an athlete executing a gymnastic exercise. Several experimental techniques are available to evaluate the successive positions (and hence the M matrices) of the body segments. The next step is generally the speed and the acceleration calculation. This operation, that consists in the evaluation of the matrices W and H , can be efficiently executed remembering the eq.s (1) and numerically calculating M' and M''

$$(2) \quad M'_{\langle t \rangle} \cong \frac{M_{\langle t+dt \rangle} - M_{\langle t-dt \rangle}}{2 dt} \quad M''_{\langle t \rangle} \cong \frac{M_{\langle t+dt \rangle} - 2 M_{\langle t \rangle} + M_{\langle t-dt \rangle}}{dt^2}$$

The evaluation of $W_{\langle t \rangle}$ and $H_{\langle t \rangle}$ through the use of the simple formulas (1) does not present numerical problems because the inversion of the matrices M can be done without the execution of any division.

3.2-IDENTIFICATION OF ISA. The ISA can be easily evaluated by the matrix L that is directly obtained from W . In fact the unit vector U is already contained in L while the pitch p and the point P can be easily evaluated as:

$$p = U^t B; P = U B. \quad (\text{In fact } U^t U = 0, U^t U = 1 \text{ and the pseudo-inverse of } U \text{ is } -U)$$

3.3-TESTING THE JOINT MODELS. As an example, we now analyze the problem to evaluate the error which is introduced when a given human articulation is considered as a revolute pair. This task can be performed analyzing (during the execution of a movement) the locus of the ISA between the two parts of the body which are connected by the articulation. If the considered parts of the body were connected by a revolute pair, the matrix L would be constant and the pitch would be null. Since the human articulations are not exactly revolute pairs, L will never be constant and the pitch of the ISA will never be null. An index of the variability of L is the norm ε of ΔL :

$$\varepsilon = \max(\|\Delta L\|) = \max(\|L_{\langle t \rangle} - \bar{L}\|)$$

where $L_{\langle t \rangle}$ is the instantaneous value of L and $\bar{L}$ is its average value. Moreover, the average value $\bar{p}$ of the pitch can be extracted from $\bar{L}$. Therefore the values of ε and $\bar{p}$ are proportional to the error that would be introduced with the substitution of the articulation by a revolute pair. Practically, a joint can be approximated with a revolute pair when both ε and $\bar{p}$ are smaller than two pre-established values.

Approximated spherical pairs can be tested with similar procedures.

3.4-IDENTIFICATION OF ACCELERATION CENTER. The acceleration center P^* of a rigid body can be easily evaluated, if it exists, by the following formula [3]

$$P^* = -G^{-1} A_0, \quad \text{where } G \text{ and } A_0 \text{ belong to the acceleration matrix } H.$$

4-DYNAMICS. The extension of this method to the dynamics analysis of human motion is presented in [4].

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MODEL OF HUMAN SPINE SYSTEM

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INTRODUCTION

The human spine is anatomically defined as a set of vertebrae, intervertebral disks and intervertebral joints. While dealing with spine mechanics, however, one has to take into account that the spine cooperates with other parts of skeleton and with muscular system as well as the nervous system which controls muscles. Thus constructed Human Spine System is too complicated even for modern analytic and numeric methods to consider all its parts simultaneously and with equal precision. The possible way to overcome the difficulty is two-stage modelling. In both stages the tool is Finite Element Method (FEM). In stage one a general but simplified model of the whole system is considered. It serves to determine those forces and dislocations, in principal nodes of the system, which are due to external loads. In stage two the chosen subsystem is considered in detail.

GENERAL FEM MODEL OF HUMAN SPINE SYSTEM

The general model consists of the following sets:

- 24 vertebrae; each vertebra consists of 34 rigid 3D finite elements;
- 23 intervertebral disks; nucleus pulposus of each disk is modelled with 18 incompressible elements and annulus fibrosus consists of 16 anisotropic 3D elastic elements;
- ligaments which are modelled with 400 anisotropic 3D elastic elements;
- chest (12 pairs of ribs and sternum) divided into 260 rigid elements connected by 34 flexible elements modelling cartilage;
- fragments of head, scapula and pelvis composed of 50 rigid 3D elements; the set includes only those fragments of skeletal system to which are attached muscles having their other attachments on the spine or chest;
- muscles modelled with 600 anisotropic, elastic 3D elements which contract (when stimulated) along muscle fibres. The contraction causes tension in muscles, and therefore, through attachment points (nodes) they act upon vertebrae, ribs, sternum etc.; there are 350 independently stimulated muscle zones (elements or groups of elements).

Moreover, limitations of the relative movement of ribs and vertebrae in intervertebral joints and rib joints are introduced in the model in the form of kinematic dependences.

The described physical model leads to a mathematical model composed of 12882 equations, generally non-linear, linearized for a given spine position.

Fig.1 presents the computer printout of the general model (side view). Configuration of vertebrae is like in straightened spine position. Fig.2 presents the result of one of the simulations in the form of a deformed spine. In that simulation the model has been loaded with the upper part of the human body weight equal 300 N. The load has been substituted with a concentrated forces system applied to finite element nods (~20000 nods). External load, modelling a weight of 400 N held in hands, has been applied in the form of concentrated forces to finite element nods around scapulas, symmetrically to sagittal plane.



Fig. 1



Fig. 2

No. Value

1 -	-.196E+02
2 -	-.188E+02
3 -	-.163E+02
4 -	-.146E+02
5 -	-.130E+02
6 -	-.113E+02
7 -	-.962E+01
8 -	-.796E+01
9 -	-.629E+01
10 -	-.463E+01
11 -	-.296E+01
12 -	-.129E+01
13 -	.374E+00
14 -	.294E+01
15 -	.371E+01
16 -	.537E+01
17 -	.704E+01

The muscular system tension equilibrates body and external loads, simultaneously keeping the straightened spine position (pelvis cannot change its location). The numeric characters in Fig.2 stand for dislocation values in sagittal plane (in mm - see the table on the right side of Fig.2). Calculations gave also the values of tensions in all elements of the system and forces acting between them in nods.

DISCUSSION

The presented general model of Human Spine System has proved its usefulness in many cases. Results obtained with its help serve for detailed statics analysis of the chosen spine subsystems (e.g.: Dietrich et al.). Our present works are conducted towards extending the model in order to allow for analysing some dynamic properties of the spine system (natural frequencies, vibration forms, etc.).

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BIOMECHANICAL ANALYSIS OF THE GLENO-HUMERAL JOINT
IN ERGONOMICAL ASPECTS

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INTRODUCTION

The application of the biomechanical model of the gleno-humeral joint (GHJ) ergonomical activities, including the variety of joint types found in population, has been discussed. The essence of the proposed procedure resolves itself into the choice of such a worker for the working process, whose acetabulum profile of the GHJ is known, to reduce the harmfulness of the joint loading to a minimum. The calculations proved that for certain types of occupations an optimum shape exists of the joint surface for which the GHJ reaches the maximum load capacity.

METHOD

The investigations have been restricted to a certain class of chosen working processes in a lamp producing factory. Ten typical kinds of activities were distinguished, performed at the intra-factory transport stand. Using the values of the geometrical parameters of the acetabulum as well as the data concerning the loading, the maximum load capacity for the respective acetabulum profiles during the realization of ten different types of activities has been calculated by means of an appropriate algorithm. In choosing the acetabulum from the point of view of the maximization of multi-criterion functionality the weight-correlation method (Pogorzelski, 1986) has been employed. It solves the problem by characterizing an optimum acetabulum with one number found out on the basis of partial utilities in the form:

$$q_0 = \sum_{i=1}^{10} w_i \cdot q_i \quad \text{where : } w_i - \text{weights, } q_i - \text{partial criteria}$$

The maximum values of the joint load capacity during performing the ten chosen types of activities, calculated for the 17 chosen acetabulum profiles have been regarded as partial criteria. Partial criteria are treated as different evaluation aspects of the given acetabulum. A problem of adjusting $q_1, \dots, q_{10}$ criteria arises by characterizing them with one number.

RESULTS

The analysis of the forces in muscles, ligaments and of contact force between humerus and scapula as well as the analysis of the boundary load capacity shows that their values are influenced by the geometry of the joint (Gielo-Perczak, 1988). In Fig.1 the value of contact force -RS (dash-dot-dot-line), in muscle 1-RM1 (continuous line), in ligament 1-W1 (dotted line), in ligament 2-W2 (dash-dot-dot-line) for this type of the acetabulum profile

are presented. The calculations of the loading (Fig.2) of the joint (Fig.1 arrow 5) show that certain arm positions under given load are impossible due to the damaging stresses.

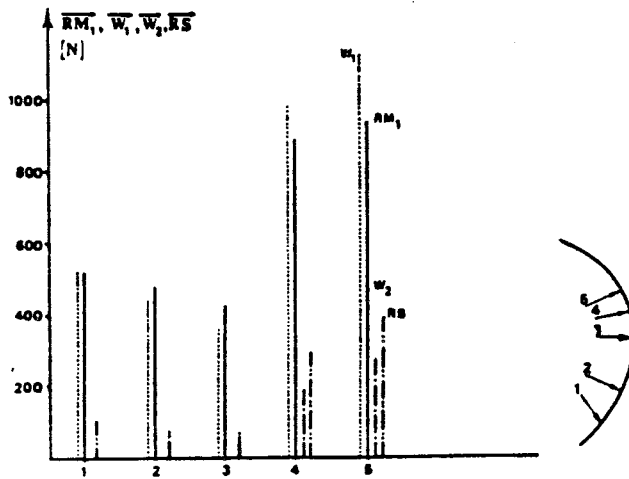


Fig.1

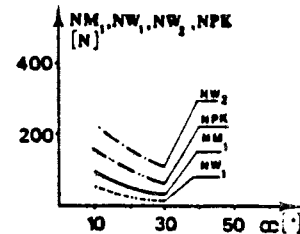


Fig.2

The ranking list of the profiles for the investigated work-stand has been obtained. The compromise criterion has been computed by means of WAKOR programme (software for weight-correlation method):

$$q_0 = 0.0018 \cdot q_1 + 0.0013 \cdot q_2 + 0.0018 \cdot q_3 + 0.0074 \cdot q_4 + 0.0053 \cdot q_5 + 0.0106 \cdot q_6 + 0.0117 \cdot q_7 + 0.0195 \cdot q_8 + 0.0034 \cdot q_9 + 0.0004 \cdot q_{10} + - 10.4633$$

Thus the participation of the individual criteria in the compromise criterion could have been fully observed. It implies the fact that not only the mutual meaning of the particular loadings but also biomechanical capabilities of the investigated joint have an influence on the final form of the criterion.

CONCLUSIONS

An essential property of the assumed model of the GHJ is the possibility of evaluating the influence of the geometrical parameters of the joint in the function of the arm position towards the trunk and the load conveyed by the arm. On the basis of the obtained results a range of practical applications is expected including in designing working stands, assigning the employers to their appropriate working stands and in examining the arduousness of the chosen activities for the human organism.

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THE EFFECT OF OVERSPEED TRAINING ON SPRINTING SPEED

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The study of the techniques of improving sprinting speed has been of interest to biomechanists, sprint coaches and athletes for many years. One technique which has received some support is overspeed training, or the production of sprinting speed by the athlete faster than can normally be produced, using some external aid such as a high speed treadmill, a towing technique, or downhill running (Dintiman, 1984; Mero and Komi, 1985). The purpose of this study was to determine the effects of overspeed training, combined overspeed and overload training, and traditional sprint training on the sprinting speed of male college athletes. Eighteen male varsity football players were divided into three groups of six subjects per group, and each group was subjected to a different training program. The control group participated in sprint training free of any external loading; a second group participated in sprint training accompanied by overspeed training using a Sprint-Master towing device; and the third group participated in overspeed tow training while wearing a ten-pound weight vest. The subjects were tested using both a pre- and post-test: while being timed while performing a maximum 40-meter sprint, and filmed from a sagittal view while performing a maximal sprint. The filming was conducted on an indoor track using a Photo-Sonics 1PL 16mm high speed movie camera set at a filming rate of 100 fps. The subjects were timed and filmed for two trials, both before and after a six-week training program.

The film of each athlete was digitized, and the digitized data was smoothed using a Fast Fourier transform technique with a cutoff frequency of 6 Hz. The resulting smoothed linear and angular displacement data was differentiated to produce velocity and acceleration values for each of the kinematic values of interest in the sprinting technique analysis. These parameters were calculated for each subject, and included: horizontal and vertical velocity of the center of gravity (CG); stride rate; stride length; stride time; support time; non-support time; horizontal foot position in relation to the CG; position of each of the leg and arm segments at takeoff, mid-stride, and touchdown; peak and average angular velocity of the limb segments during the stride; and acceleration of the CG during the takeoff phase of the stride (Williams, 1985; Mann and Herman, 1985).

An analysis of covariance (ANCOVA) was employed to determine whether there were significant differences among the pre/post test scores as a result of the three different training protocols. Of the twenty kinematic sprinting variables measured in this study, only five exhibited significant differences at the prescribed $p < .05$ level. The sprinting performance indicators which were significantly improved from pre- to post-test were: stride length (decreased), stride rate (increased), support time (decreased), and horizontal velocity (increased). The kinematic variable which exhibited the greatest improvement over the training program was the position of the thigh at maximum hip flexion, which increased significantly after the sprint training ($P < .01$).

The kinematic and performance variables that exhibited significant differences from pre- to post tests were treated with a least square means (LSM) test. The LSM calculations were used to determine which group differed significantly from the others in each of the variables measured. Only horizontal velocity displayed a significant LSM value at the $p < .05$ level between the weighted, overspeed group and the control group. There were differences in the means scores between subjects in each of the groups as a result of the different training techniques, but these differences were not statistically significant. The overspeed, overload group experienced the greatest alteration in the kinematic and performance variables in sprinting, suggesting that this training technique may be potentially superior to traditional sprint training and overspeed sprint training.

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JOURNAL OF BIOMECHANICS

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SOME MYOELECTRIC AND KINETIC CORRELATES OF GYMNASTIC SKILL

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Selected results of long term research activity in the area of sport gymnastics, encompassing altogether 25 gymnasts, are presented. The approach gradually introduced measurement of take-off force, lower extremities surface EMG signals and heuristic variables, during performance of somersaulting movements. Relations between measurement variables have been investigated via linear regression and correlation analyses.

In backward somersault from the standing position high skill level has been found associated with impulsive take-off force waveform, having the maximum value typically three times or more greater than body weight, duration ≤ 300 milliseconds and gradient 10 - 18 N/milliseconds. Good performances of this quasi-symmetric movement are characterized by bilateral symmetry of rectified and smoothed EMG signals of ankle extensors, with correlation coefficient of ~ 0.8 . In running forward somersault there is a higher correlation between force impulsiveness and skill of performance. Generally, all myoelectric and kinetic parameters display high degree of individuality.

Besides providing some insight in fundamental properties of neuro-musculo-skeletal system function, these findings are relevant to the practice of gymnastic coaching as well.

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SIMULATION OF 3D MOVEMENT REGISTRATION BASED
ON THE DIRECT LINEAR TRANSFORMATION METHOD

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In order to be able to determine the optimum experimental conditions for any particular movement research project, a computer program was developed to simulate a 3D movement registration based on the Direct Linear Transformation method. Considerations that played a part in the realization of this open-structured Modula-2 program, will be discussed.

Depending on the kind of motions that have to be studied, one or more of the following three recording techniques are available: 1) cinematographic film, 2) video, 3) Pulsed IR-LED detecting systems (Selspot, Watsmart etc.). Each of these recording techniques induces specific errors in the projections. These errors were classed and modelled so as to be able to incorporate them into the program. In addition to the possibility of choosing one of the three recording techniques, the simulation offers various calculation strategies to solve the overdetermined sets of equations that occur in the DLT-method. It appears that some of the fast procedures used in real-time experiments, such as the Subject Tracking Cameras method, are numerically unstable.

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LENGTH CHANGE OF THE HUMAN KNEE EXTENSORS COMPLEX DURING
PRELIMINARY COUNTER MOVEMENT IN VERTICAL JUMP

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Some investigators reported that a preliminary counter movement enhanced the performance of a vertical jump. It was suggested that one of the reasons for this enhancement was the utilization of the elastic energy which had been stored in the forcibly stretched actomyosin cross bridges in the knee extensors during the counter movement (Bosco et al., 1981). It has not been clarified, however, whether the cross bridges were forcibly stretched during the counter movement. The purpose of this study was to estimate the length change of the contractile component (CC) of the human knee extensors complex during the counter movement in a vertical jump.

METHODS Four male subjects, keeping their hands on their hips, performed vertical jumps with and without a preliminary counter movement on a force platform (KISTLER-9281B). The subjects were filmed (102 frames/s) with a 16 mm cinecamera (NAC-1P). Surface EMGs were recorded from the leg muscles. The torque around the knee joint during the counter movement was determined using a link model, ground reaction force measurements, and film analyses (Winter, 1979). The resultant force of the knee extensors complex during the counter movement was determined by dividing the torque by the moment arm of the quadriceps femoris muscle (Smidt, 1973). The length change (ΔL_t) of the tendon in the knee extensors complex was determined from the force-compliance relationship of the knee extensors complex (Shorten, 1985). It was assumed that the compliance of the tendon was $3/4$ of the compliance reported by Shorten and that the rest of the compliance was ascribed to the compliance due to the cross bridges. The length change (ΔL_m) of the knee extensors complex during the counter movement was determined by integrating the moment arm with respect to the knee angle. The length change (ΔL_c) of CC in the knee extensors complex was estimated using the following equation: $\Delta L_c = \Delta L_m - \Delta L_t$.

RESULTS and DISCUSSION Three subjects jumped higher in the vertical jump with the counter movement than in the jump without the counter movement ($n=5$, $p < 0.01$). Fig. 1. shows a typical example of the records observed in a vertical jump with the counter movement. It seems reasonable that the resultant force of the knee extensors was determined by merely converting the torque around the knee joint into the force since the activities of the biceps femoris muscle and the gastrocnemius muscle were not so high in that period. Fig. 2. shows a typical example of the time courses of the knee angle, the length change of the muscle complex, and the estimated

length changes of the tendon and CC in the same vertical jump. In all four subjects there was not appreciable stretch of CC during the latter half of the counter movement when the activity of the knee extensors was nearly maximum.

This result suggests that most of the actomyosin cross bridges in the knee extensors were not forcibly stretched during a preliminary counter movement in a vertical jump, as opposed to the suggestion by Bosco et al. (1981). The rate of the compliance of the tendon to the total compliance of the knee extensors complex (Shorten, 1985) seems to be underestimated during the counter movement, especially in the latter half of that movement, even though taking into account the dependence of the compliance due to the cross bridge on the muscular activity (Ford et al., 1977). Consequently, the actual shortening of CC during the counter movement might be underestimated in the present study, for example in Fig. 2.

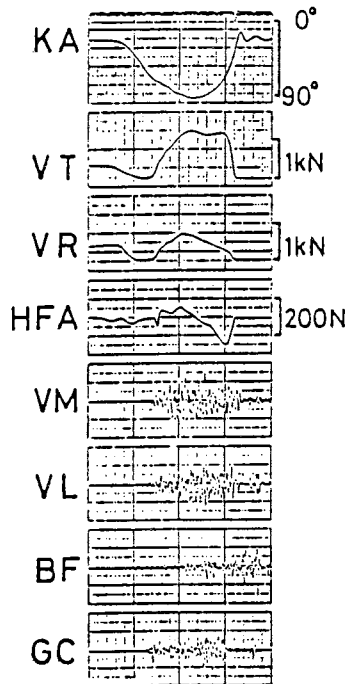


Fig. 1. An example of records observed in a vertical jump with a counter movement. KA=knee joint angle; ground reaction force, VT,VB=vertical components at the toe and 0.24m back of the toe, respectively, HFA=horizontal fore-aft component; EMG, VM=vastus medialis, VL=vastus lateralis, BF=biceps femoris, GC=gastrocnemius.

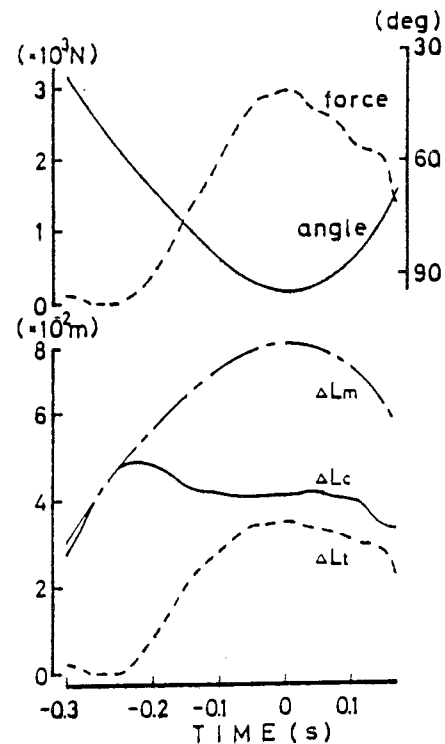


Fig. 2. An example of time courses of the knee joint angle, the force exerted by the knee extensors, and the length changes of the knee extensors complex. Time is expressed relative to the instant of the onset of knee extension ($t=0$).

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Joint Moments Caused By Muscle Action - Isometric Contractions

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The observed performance of the skeletal system reflects the moment that the muscles can exert about each joint. The joint moment, in turn, is influenced by the relative position of the adjoining segments, because of the action of bi-articular muscles. This property allows muscles to act in a semi-passive manner and conserve energy in certain movements. For example, the hamstring muscles can remain unaltered in length when hip flexion occurs with simultaneous knee flexion. This study investigates the moments caused by muscular action at the hip, knee and ankle joints during maximal isometric contractions and the effects of joint angle on the joint moment.

Mathematical Model

A four-component mathematical model has been used to describe the mechanical properties of skeletal muscle. The model incorporates

- a contractile element (as described by Woittiez et al, 1984),
- a parallel elastic element (as described by Woittiez et al, 1984),
- a series elastic element (as described by Bahler, 1967), and
- a tendon elastic element (as described by Benedict et al, 1968).

Muscle sizes and lengths have been obtained from Pierrynowski & Morrison (1985) and Wickiewicz et al (1983) and these were scaled to the particular subject based on anthropometric measurements. The maximum isometric muscle force (optimum force) was assumed to be directly proportional to the cross - sectional area of the muscle. This is not strictly correct but is the best available assumption in the absence of data on the individual subject's muscle physiology.

The active range of motion for each joint was used to obtain the maximum and minimum muscle lengths and these in turn were used to calculate the optimum fibre length (defined as the fibre length at the maximum isometric force). Hence, actual muscle force magnitudes are a function of the optimum muscle force, the optimum fibre length, the actual muscle length and the level of muscle activation.

The muscles contributing to the flexion or extension of a particular joint in the sagittal plane were activated maximally; the forces produced by the other muscles were set to zero. Twelve major muscles of the lower limbs were considered, namely the iliopsoas, the glutei, the hamstring group, the quadriceps group, the gastrocnemius, the soleus and the tibialis anterior.

Results

Experimental results were obtained using a Cybex isokinetic machine for isometric contractions at the hip, knee and ankle joints. Due to the fact that some of the muscles are bi-articular, the position of one joint will influence the moment produced at another joint. Additional experimental results for the knee and hip were obtained from Smidt (1973), Lindahl et al (1969), Lunnen et al (1981) and Waters et al (1974). Figure 1 shows the theoretical predictions of the mathematical

model compared to experimental results obtained from Lindahl et al (1969) for six subjects during isometric knee extension. At this stage our experimental results are not as encouraging and more experimental work needs to be undertaken.

Conclusions

Theoretical predictions from the mathematical model compare favourably with some published experimental results. At this stage, more experimental work is being undertaken and will be published at a later date.

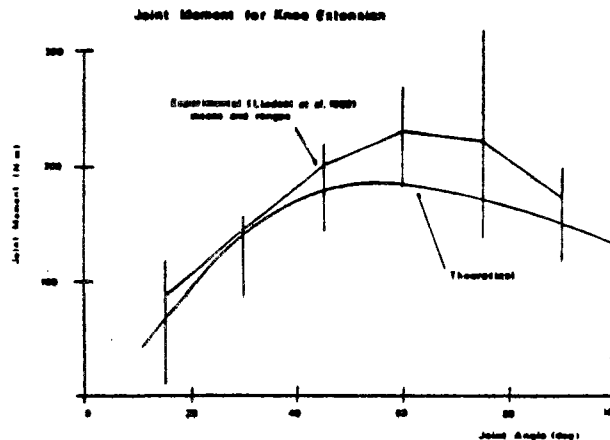


Figure 1

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EFFECT OF AGE ON THE TIME COURSE OF POSTURAL ADJUSTMENTS ACCOMPANYING STEPPING

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INTRODUCTION

One of the consequences of aging is an increase in reaction time and movement time for the performance of voluntary movement (Welford, 1980). Investigations of these changes, however, have been limited to simple arm and hand movements in which the functional consequences of this slowing are not serious (Gottsdanker, 1982; Spirduso and Clifford, 1978). In a task such as stepping to a new position to maintain balance or avoid collision with an obstacle, such slowing can increase the risk of falling. In addition, the time to complete such a movement is increased by the need to transfer weight to single-foot support prior to actually stepping with the contralateral limb. Since balance has been shown to decline with age (Bohannon et al., 1984; Woolacott et al., 1986), weight-transfer time might also increase with age. The purpose of this study was to examine the influence of age on the time course of postural adjustments accompanying stepping: reaction time, weight-transfer time and movement time.

METHOD

Eleven young (20-30 years) and nine elderly (60-75) subjects participated in the study. Elderly subjects were healthy and living independently. Subjects stood with the right foot only on a force plate and were asked to perform a choice reaction-time task involving stepping with the right foot forward, to the side or backward. A stimulus panel, displaying 3 lights oriented in the direction of each response, signalled response initiation. The direction of stepping was presented in a pseudo-random order across 30 trials. Each stepping response was divided into 3 phases: 1) reaction time was the interval from a visual signal onset to the initial rise in vertical reaction force; 2) weight-transfer time was the interval from the onset to offset of change in vertical reaction force; 3) movement time was the interval from the onset of single support to the completion of the step. The completion of the step was signalled by the closure of a microswitch when the foot contacted the floor. Each of the three measures was analyzed with a 2 x 3 (age x response direction) analysis of variance. A Neuman-Keuls test was used for post-hoc analysis.

RESULTS

Mean reaction time displayed a small, but significant, increase (10%, $p < .001$) with age and was uninfluenced by the direction of stepping for both young and elderly subjects. However, greater age effects were observed for weight transfer time and movement time. Mean weight transfer time increased by 72-122% ($p < .001$) and mean movement time increased by 54-106% ($p < .001$) for elderly compared to young subjects. In addition, the direction of stepping influenced weight transfer time and movement time for young but not elderly subjects. Among young subjects, weight transfer time was significantly longer when stepping forward or backward than to the side and movement time was significantly longer when stepping backward than forward or to the side. These results are displayed in Table I.

Table 1: Mean Performance Times (msec) of Young and Elderly Subjects

	Direction of Step		
	<u>Forward</u>	<u>Side</u>	<u>Backward</u>
<u>Reaction Time</u>			
Young	340	351	355
Elderly	383	386	385
% Change	12.6	10.0	8.5
<u>Weight-Transfer Time</u>			
Young	235	172	233
Elderly	407	282	401
% Change	73	122	72
<u>Movement Time</u>			
Young	183	187	241
Elderly	326	385	371
% Change	78	106	54

DISCUSSION

When stepping with the right left to a new position, reaction time, weight-transfer time and movement time were observed to increase with age. Reaction time showed the least increase with age; the per cent increase observed is comparable to that reported for a variety of simple arm and hand movements (Welford, 1980). Increases in reaction time have been related to increase central processing time to select and initiate an appropriate motor response (Welford, 1980). Of greater interest are the large increases in weight-transfer time and movement time observed for older subjects when stepping to a new position. For a similar 3-choice reaction time task involving extension movements of the arm, Spirduso and Clifford (1978) reported a 23% increase in movement time for older subjects. For the task of stepping in this study, weight transfer time and movement time increased by 72-122% and 54-106%, respectively. Weight-transfer time reflects the time to effect postural adjustments necessary for stepping to a new position; this has not been considered by previous investigations. The increased time for this phase of responding, as well as movement time, likely reflects poorer balance control by the elderly.

This study provides greater insight into the functional significance of the slowing of movement initiation and performance with age. Increases in reaction time, weight-transfer time and movement time can compromise the safety of the older adult and increase the risk of falling. In addition, this task may provide a useful tool for assessing the level of postural instability in the elderly.

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SOME OBSERVATIONS ON THE USE OF CENTER OF PRESSURE SIGNAL FOR ASSESSMENT OF BALANCE

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INTRODUCTION

The center of pressure is used by researchers and clinicians to assess balance. This signal determined from the ground reaction forces and moments is quantified both in the time and frequency domain. A major concern is the lack of information about the signal characteristics, such as stationarity, the right sampling duration, and range of normal values under different sensory and support conditions. This is evident in a lack of standardization in data collection and analysis seen in the literature. In this study, we examine the characteristics of this signal.

METHODS

Young healthy university students ($n=12$, average age = 20-30 years) participated in the study. Subjects were instructed to stand on the forceplate. Force plate data were collected (30 sec) online (sampling frequency = 64 Hz) for three sensory conditions (eyes open, eyes closed, head extended) and three support conditions (2-foot, 1-foot, Rhomberg stance). Following this, subjects were required to stand quietly for an hour. During this, data were sampled every five minutes for 30 sec.

DATA ANALYSES

The center of pressure signal was calculated over 30 sec for each condition and the standard deviation (SD) values around its mean determined. The signal was also analyzed in the frequency domain using a FFT algorithm. The power from 10 Hz to 15 Hz was averaged and a standard deviation calculated. This threshold ($\bar{X} \pm 3 \text{ S.D.}$) was used to determine the band where the power in the spectrum was above this level. For this band of frequencies, the mean power frequency (MPF) was calculated. The stationarity of the signal was analyzed with a standard run test.

RESULTS

Table 1 reports the range of centre of pressure values recorded across 12 subjects for 2-foot, 1-foot and Rhomberg stance performed with the eyes open. Manipulation of sensory conditions during 2-foot stance had very distinct influences on the mean values of SD and MPF as shown in Figure 1. Analysis of stationarity of SD and MPF for 2-foot stance over one hour revealed that the CP signal is stationary, i.e., no systematic shifts occur over time. Results for one subject are displayed in Figure 2.

Table 1: Mean/Range of Centre of Pressure: Standard Deviation (SD), Frequency Band Width (Band) and Mean Power Frequency (MPF)

	<u>2-foot</u>	<u>1-foot</u>	<u>Rhomberg</u>
SD (mm) - AP	5.87/2.56-11.5	12.7/6.96-30.3	6.67/2.78-14.8
Lat	1.75/0.97-2.96	5.68/3.90-7.72	5.51/3.90-8.08
Band (Hz) - AP	3.18/2.28-4.50	3.87/2.31-5.81	4.46/2.81-6.10
Lat	3.52/2.41-4.62	5.40/3.78-6.60	4.47/3.15-6.75
MPF (Hz) - AP	0.63/0.44-1.04	0.80/0.43-1.30	1.05/0.64-1.34
Lat	0.90/0.45-1.20	1.28/0.99-1.59	1.02/0.68-1.49

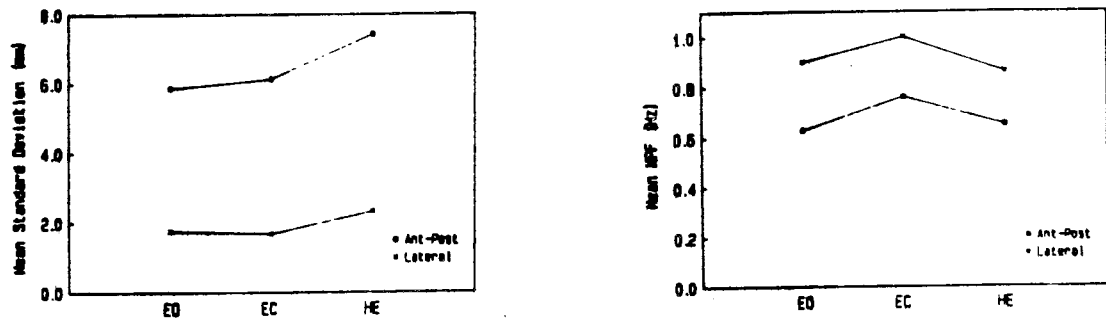


Figure 1: Centre of pressure mean SD and MPF across all subjects for three sensory conditions: eyes open (EO), eyes closed (EC) and head extended (HE).

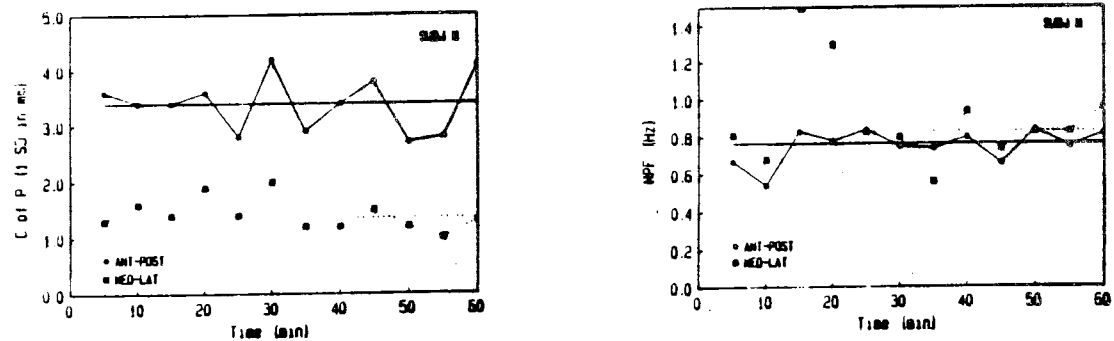


Figure 2: Stationarity of centre of pressure SD and MPF for one subject sampled at 5 minute intervals over 1 hour.

DISCUSSION

Two-foot, 1-foot and Rhomberg stance are commonly used tests of postural stability (Bohannon et al., 1984; Lucy and Hayes, 1985). This study presents normative values of centre of pressure SD, frequency band and MPF for a healthy, young adult population. In addition, it appears that manipulations of eye closure and head extension may help identify the relative contribution of different sensory inputs to the regulation of 2-foot stance. Eye closure, thereby removing visual input, was characterized by an increase in MPF. Head extension, which has been suggested to neutralize vestibular otolith input (Brandt et al., 1986), was characterized by an increase in SD. Finally, the presence of stationarity of CP signals further affirms the potential value of these measures for assessing postural stability.

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ACKNOWLEDGEMENTS

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TREADMILL ERGOMETER FOR POWER OUTPUT MEASUREMENT DURING SPRINT RUNNING

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INTRODUCTION

Exact evaluation of mechanical work during running on motorized treadmill is difficult, and only a few authors have made attempts in this regard (e.g. Asmussen & Bonde-Petersen, 1974). Problems are not easier when one wants to evaluate the power production in treadmill running. Evaluation of aerobic power from the measurement of oxygen consumption is only an indirect and incomplete measure of the total power output. Attempts have, however, been made to record maximal speed and power output in non-motorized treadmills (Lakomy, 1984). The present report focuses on the ergometer treadmill and makes an attempt to describe how power output can be measured during changing load-speed conditions of running.

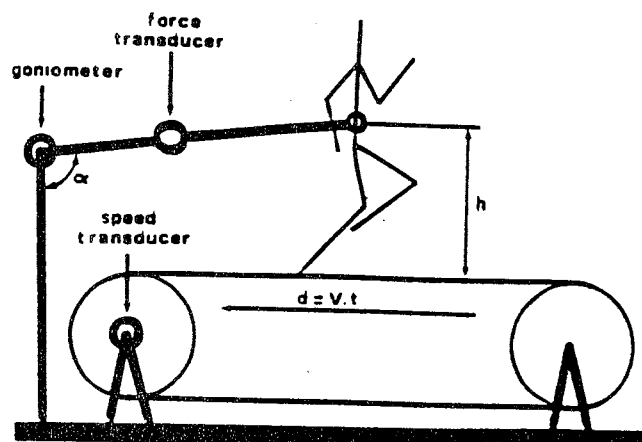


Figure 1. Schematic sideview of the ergometer

ERGOMETER

Figure 1 presents the sideview of the treadmill. The subject is fixed with a strong belt to an horizontal bar, which is attached at its other end to a joint containing a potentiometer for recording of vertical displacement during running. A strain gauge system installed in the middle length of the bar records the horizontal traction force of the runner.

During running the subject controls the speed of the treadmill by acceleration or deceleration, and these phases are sensed by the strain gauges on the horizontal bar. A constant torque motor installed in the rear wheel of the treadmill is not used to drive the treadmill belt, but to compensate the belt friction or to simulate different loads. Signals from the goniometer (vertical displacement), transducer (horizontal traction force) and from the treadmill (belt speed) are recorded by a microcomputer, which then calculates the necessary parameters as follow :

Potential Power (PP) :

$$PP = m \cdot g \cdot h \cdot t^{-1}$$

Kinetic Power (KP) :

$$KP = F \cdot d \cdot t^{-1}$$

Total Power (TP) :

$$TP = PP + KP$$

where

m = mass of the subject

g = gravitational constant

$$9.81 \text{ m} \times \text{s}^{-2}$$

h = vertical displacement of the runner

t = time

F = traction force

d = horizontal displacement of the treadmill belt

During sprint running there is usually no phase shift between PP and KP. For this reason adding of PP and KP to obtain TP is justified.

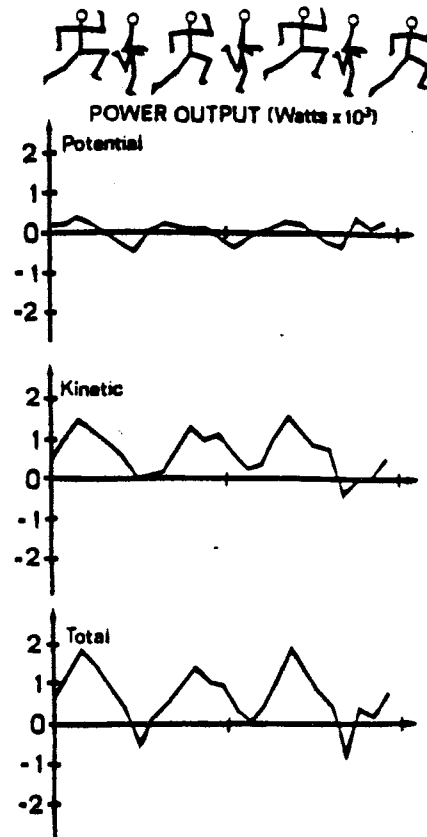


Figure 2. Power outputs

Figure 2 presents an example of the potential, kinetic and total power outputs during sprint running. In the example the peak total power reached the value of close to 2000 watts, the average power being approximately 650 watts for the complete stride cycle.

APPLICATIONS

Because the ergometer allows extreme variations in load and velocity, it can be used for respective analysis of load-speed relationships. The acceleration phase of sprint running is a good example of such a situation.

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INTERNAL SHOULDER FORCES DUE TO AN EXTERNAL LOAD
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INTRODUCTION

The shoulder complex consists of more than 20 muscles and three skeletal bones and it is hard to predict individual internal forces. Up till now no biomechanical shoulder model exists that is capable of predicting all of the musculoskeletal forces due to external load. The role of individual components in the shoulder has for example been studied using EMG. Observations on the effects of axillary and suprascapular nerve block have also been made (Howell et al. 1986).

The purpose of this study is to describe a tentative shoulder model, which can be used to predict internal musculoskeletal forces in structures attached to the humerus where each force is a function of static external load in static working position. The model is based on the works of Högfors et al. (1987, 1988).

METHODS

In our idealized mechanical model the shoulder bones are considered to be rigid bodies. The joints are modelled as ball joints. Muscles or muscle parts are modelled by stretched strings following the shortest path between attachments unless constrained by other structures. Forces acting on the rigid bodies are divided into muscle forces, joint contact forces and ligament forces.

In order to obtain the directions of the forces it is necessary to know the relations between the positions of the bones, the shoulder rhythm. Our shoulder rhythm is based on the work by Högfors et al. (loc. cit.) but extended to be used in the complete range of shoulder motion.

For our mechanical system consisting of three internal bones it is possible to obtain at most 18 equilibrium equations. Since the system is indeterminate there are an infinite number of solutions to this problem. To obtain a solution we have chosen to minimize the sum of squared muscle tensions, i.e. minimize $\sum (F_i/A_i)^2$. (F_i = internal muscle force, A_i = physiological cross-sectional area.) The choice of this specific objective function is based on a comparative study on objective functions made by Dul et al. (1986).

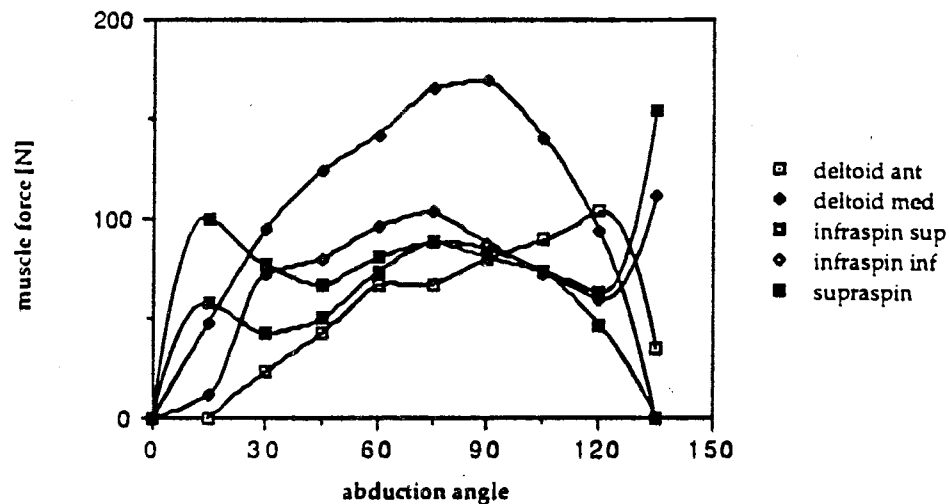
Furthermore we include in our model constraints on the muscle forces so that these cannot become negative or exceed a maximum value $F_{\max}$. The cross-sectional area and maximum force for the individual muscles are known from the work of Järvholm et al. (1988).

Computer calculations were done for static positions following an abduction pattern with straight arm in the plane of the scapula (i.e. a plane 45° forward from a frontal plane) and a forward flexion pattern in a sagittal plane with the arm flexed 90°. The interval estimated was 0° - 135° abduction and 0° - 135° forward flexion. The hand load was taken to be 1 kg. Anthropometric data were taken from the literature. This study is restricted to the musculoskeletal forces acting between the humerus and the other shoulder components.

RESULTS

For both abduction and forward flexion solutions were found describing the force levels and directions of the force producing components involved. Figure 1 is an example showing the major muscle forces in shoulder abduction. The contact force in the glenohumeral joint reached a typical value of 600 N in a 90° abduction position when the handload was 1 kg. The same handload in 90° forward flexion predicted a contact force of 550 N.

Fig 1: Major muscle forces in elevation in the scapular plane



DISCUSSION

During abduction the muscles predicted to be activated are those named in the literature as being the main abductors in elevation. In flexion the result is somewhat harder to analyse. Comparisons with EMG-measurements curves do not show much covariance. However, the relation between force-output and EMG-signal is unknown.

Previous reports on contact force by Poppen and Walker (1978) gave the contact force in this sort of abduction close to the body weight (0.8 times). In our model we used anthropometric data from a person with a body mass of 75 [kg]. The contact force being 600 [N] then gives the same result as in the study made by Poppen and Walker.

In Howell's study the results show that a paralysis of the axillaris nerve caused a 50% loss of ability to produce torque (in 90° abduction). A paralysis of the suprascapularis nerve was claimed to reduce the maximum torque also with 50%. In our biomechanical model we get the results of 45% reduction for axillary nerve block and 58% reduction for suprascapular nerve block. Neither did Howell include the infraspinatus muscle among possible abductors in the scapular plane. The results from our study indicates that the infraspinatus indeed act as an abductor in the plane of the scapula.

ACKNOWLEDGEMENTS

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A BIOMECHANICAL AND PHYSIOLOGICAL COMPARISON OF STANDARD AND REVERSE BICYCLE PEDALLING

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The purpose of the study was to compare the standard forward pedalling with the reverse pedalling mode.

Six male subjects with no background in competitive cycling were tested in the two pedalling modes at speeds of 50 and 75 rpm on a pedalling device connected to a bicycle ergometer. Three continuous work-loads of 32, 64 and 128 watt were applied at each pedalling mode for 4 min each. The subjects performed 10 training sessions on the reverse mode prior to the testing sessions.

The following parameters were measured:

1. Vertical and anterior-posterior forces applied to the pedal
2. Electrogoniometry of the hip, knee and ankle joints
3. Electrogoniometry of the crank and pedal
4. Electromiography of 5 muscles groups
5. Absolute oxygen consumption
6. Heart rate
7. Rating of perceived exertion (RPE Scale)

On the basis of the reaction forces on the pedal and the kinematic data (based on the electrogoniometric information), the moment histories at the 3 stated joints due to pedal forces and motion were calculated. Also moments at the crank and pedal axis were measured.

The results of this study suggest that the patterns of the joint moments and the EMG activity of the monitored muscles are quite different between the two pedalling modes.

The results of this study suggest that the patterns of the torques applied at the crank axis for both tested modes are similar in shape, but the the moments in the reverse mode are "moved" forward in the cycle by about 30 degrees as compared to the regular pedalling mode.

The largest differences in the torque patterns at the joints were found at the hip.

The electrical activity of the muscles tested are consistent with the torque results. The beginning of the electrical activity of the Rectus Femoris and Vastus Medialis is delayed in the reverse mode by about 30 degrees.

The EMG activity of the Biceps Femoris was found to be very different in both pedalling modes.

It was also found that the energy cost was about 6 percent higher in the reverse mode. A similar trend was shown by the results of the heart rate, and, as expected, with higher HR values at higher speeds. The RPE values were also consistent with the physiological findings.

BIOMECHANICAL DETERMINANTS OF THE RHYTHMIC STRUCTURE IN SPRINTERS START ACCELERATION

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The dynamic and cinematic parameters of running strides in start acceleration were investigated with a purpose to clear out the factors which determine the specificity of its rhythmic picture. The registration of the dynamic curves of ground contact reactions as well as the time intervals of each consecutive stride in 60-meter distance of running with maximal speed was made with the help of tensographical apparatus. The space characteristics of movement were analysed by means of speed cyclography (100 frames per second). The comparative analyses of biomechanical characteristics of sprinters' start acceleration (males - 10, females - 10) of international (8) and national (12) level was made.

The results confirm that the quality differences characterize the rhythm of start accelerations of sprinters of different qualifications. The general regularities of effective rhythm construction were determined as the system of well-organized coordination of biomechanic structure characteristics. The effective rhythm in sprinter's start acceleration is characterized by a graduate and quick approach of flight and support time correlations of running strides to the indices that are typical for the rhythm of distance running.

Two types of the rhythmic pattern in start acceleration were cleared out (table 1). The first type of the rhythmic pattern of the start is characterized by the graduate approach of the running strides frequency to its meanings near to the distance parameters. This approach is usually realized in the limits of the first 4-6 strides of start acceleration. The main peculiarity of the second type of the rhythmic pattern is the distance frequency realization just from the second stride of start acceleration. The losing of the rhythm in start acceleration is displayed in the violence of the progression of consecutive shortening in the meaning of ground contact time and simultaneous increase of the flight interval time in running stride cycle. Dynamograms of the vertical component of ground contact reactions in running strides have the distinct differences in start accelerations between the international and national level sprinters. The losing of the rhythm here is connected with the prolongation of amortisation phase time and irregular increasing of power in this phase especially. These dynamic processes in its turn are determined by the factors of cinematic character. These factors are: the selection of optimal position in starting blocks, the low

trajectory of foot motion of a free leg in the first stride, optimization of angles in knee and hip joints in every consecutive stride of start acceleration in accordance with the level of special physical performance of sprinter.

Table 1.

The time parameters (ms) of running strides in two types of sprinting start acceleration.

Type of Running		Numbers of strides											
rythmic stride	pattern component	2	3	4	5	6	7	8	9	10	11	12	
1	ground contact phase	160	150	135	120	110	100	105	100	100	95	90	
	flight phase	70	90	95	105	115	120	120	120	120	125	130	
	stride	230	240	230	225	225	220	225	220	220	220	225	
	ground contact phase	160	140	130	125	110	100	90	90	85	85	80	
2	flight phase	60	80	90	95	105	120	125	130	135	135	140	
	stride	220	220	220	220	215	220	215	220	220	220	220	

The results of the investigation demonstrate the obvious importance of the fast information about time and dynamic characteristics of running strides in the process of sprinters' technical performing. With the help of this information it becomes possible to identify the reasons of individual technical mistakes and to choose the adequate exercise for their improvement. At the same time it is naturally to suppose that the new technologies of fast information reseiving will be developed in accordance with the requirements of elite athletic performance. The theoretical aspect of the above mentioned results may be intresting for discussion in connection with the role of stride friquency stability as an important factor in the organisation of the optimal rhythmic structure of sprinting start acceleration.

Three-Dimensional Orientation of the Hips and Shoulders in Speed Skating Sprint Starts

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Introduction

The start in speed skating can represent 5 to 10 percent of the total distance to be covered in the 500 and 1000 meter races. Therefore an investigation of variables that might contribute to decreasing the time it takes for a skater to go from the start to maximum steady state skating velocity would be beneficial to the skater. Ingen Schenau and Bakker (1980) found that stroke velocity and work per stroke are variables that contribute to final velocity in speed skating. Angular relationships such as knee angle and hip angle and a skater's velocity have been explored (Ingen Schenau et al., 1983). In running, Hinrichs (1982) reported that the upper extremity was an active participant in balancing the angular momentum, and that the rotational frequency of the chest gives the hips and the legs the needed frequency to move rapidly through alternating strides. This paper will focus on the relationship of upper body angular orientation to starting performance. The nature of the relationship between hip and shoulder rotation angles and their contribution to forward rotation and balance in speed skating is unknown.

Methods

Data were collected during the Women's 1000m race at the XV Winter Olympic Games in Calgary, Alberta, Canada. Fourteen elite skaters were filmed using two synchronized, high-speed cine-cameras. The cameras were approximately 60m from the starting line, one was oriented along the y (longitudinal) axis of the track, and the other was oriented diagonally, between the y axis and the x (horizontal) axis. Nineteen anatomical landmarks on each skater, including the top of the head and centers of rotation of the neck, waist, right and left shoulders, elbows, wrists, hips, knees, ankles, heels and toes were manually identified and digitized.

Direct linear transformation (Abdel-Aziz and Karara, 1971) was used to reconstruct the 3-D coordinate data for each anatomical landmark. The 3-D data were smoothed using a quintic spline (Dierckx, 1975), fitted to the data based on a least squares criterion. The data were analyzed by means of 2-D projections and a 13 segment human body model consisting of head, chest, hips, upper arms, forearms, thighs, calves, and feet. Segmental and whole body center of gravity calculations were performed using the method of Clauser (1955).

Results and Discussion

The data presented in Figures 1, 2, 3, and 4 represent two different types of starting techniques with respect to both initial position and motion. In Figure 1, skater 1 is representative of skaters who maintained low whole body centers of gravity throughout their starts. Skater 2 is representative of skaters who showed a general upwards displacement of the center of gravity during the first 1.2 seconds of motion. Figure 2 shows that skater 1 maintained a relatively constant trunk angle during her entire start with a range of 18° to 35° , while skater 2 began with a trunk angle of 20° , and raised to 48° with the first step. Thus, the increasing trunk angle of skater 2 contributed to the upward motion of her center of gravity. Both "low" and "high" starting techniques had similar shoulder and hip rotations with respect to the z axis, or transverse plane (refer to Figures 3a and b). For both styles, the hips rotated slightly with each stroke, while the shoulders counter rotated with greater amplitude; perhaps to minimize total angular momentum about the vertical axis. The shoulder angle of skater 1 during the first stroke was similar to the shoulder angle during her third stroke. The shoulder angle of skater 2

during the first stroke was much greater than her shoulder angle during the third stroke. The shoulders and hips rotations of the "low" starter (Figure 4a) were relatively in phase and had similar amplitudes for each stroke, while the shoulder and hip rotations of the "high" starter (Figure 4b) were almost 180° out of phase. The shoulder amplitudes for skater 2 were much greater than her hip amplitudes. The latter finding is somewhat similar to the results Hinrichs found for running.

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Figure 1: Sagittal plane projection of the center of gravity

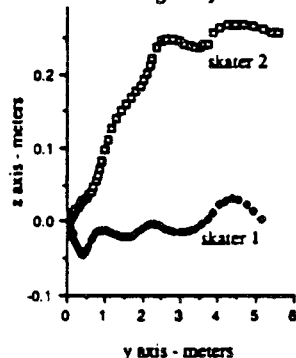


Figure 2: Upper trunk angle

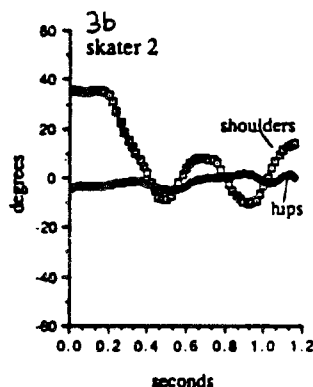
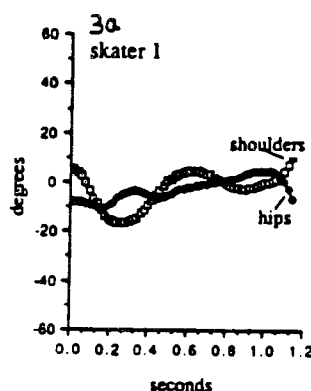
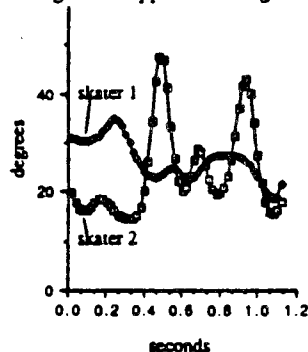


Figure 3. Rotation of the hips and shoulders about the z (vertical) axis, viewed in the transverse plane, a) skater 1, b) skater 2.

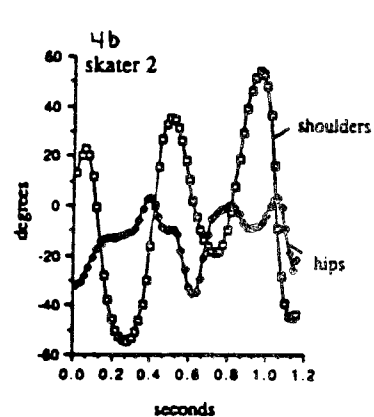
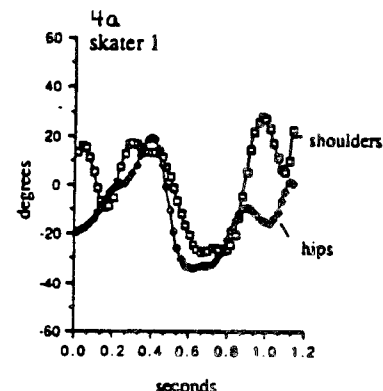


Figure 4. Rotation of the hips and shoulders about the y (longitudinal) axis, viewed in the frontal plane, a) skater 1, b) skater 2.

ARCH INDEX AND KINEMATIC LOWER EXTREMITY MEASURES

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INTRODUCTION

Little research has been completed using structural variability as a criteria for studying walking gait. General statements attributing different structural characteristics with variability in lower limb function have been made with respect to high arch (pes cavus) and low arch (pes planus) feet (Hlavac, 1977). A useful measure of the height of the plantar arch is arch index (AI). The purpose of this study was to evaluate selected rearfoot and knee gait parameters on individuals grouped according to AI as high, normal and low arch.

METHODS

Twenty-four subjects (11 males, 13 females) were recruited and assigned to one of three groups based upon a footprint evaluation of AI employing a procedure outlined by Cavanagh and Rodgers (1987). All subjects were healthy individuals who exhibited no lower extremity injuries and walked without pain or discomfort. A visual inspection of each subject's gait revealed no noticeable abnormalities or pathological gait patterns.

The experimental set-up consisted of a 15m walk-way in a large gym. A LoCam with a 15-150mm zoom lens was situated to capture a rear view of the lower extremity during the support phase in the criterion area. Data sampling was accomplished at 100 Hz. In addition, subjects were fitted with a CARS-UBC triaxial electrogoniometer to measure kinematic properties of the knee. The apparatus consisted of three precision potentiometers arranged to provide three degrees of freedom. The electrogoniometer was interfaced to a 16 channel A/D converter and ultimately to a microcomputer. Data were sampled for the electrogoniometer at 250 Hz. Walking speed was monitored by a photo-electric timing system (1.5 m/s + 5%).

Initial footprint data were collected prior to the actual experiment. A lower extremity evaluation was performed to measure static calcaneal inversion/eversion. Both static and dynamic footprints were evaluated to determine AI. The areas necessary to determine AI were calculated using a Numonics Digitizer in the area mode. Subjects were given sufficient time to practice in the experimental area. Calibration parameters for the electrogoniometer were collected on completion of the practice trials. Ten successful trials walking at the criterion speed were then collected for each subject.

The film data were digitized by a Numonics digitizer. The data were smoothed using a recursive low-pass digital filter. Rearfoot (RF) angles were then calculated. The goniometer data were evaluated to provide measures of knee motion. Ten trial means for all variables were then evaluated using a one-way ANOVA with a Tukey post-hoc procedure.

RESULTS

Means (sd) and statistical analyses for all variables are presented in Table 1. Significant differences were evident for the grouping variable AI in both the static and dynamic cases. None of the other variables measured, however, revealed statistically significant differences between the groups.

DISCUSSION

The results of this study indicate that AI can effectively group subjects based on the ratio of the area of mid-foot contact/total foot contact during stance. However, the static and dynamic measures evaluated on the groups did not differentiate between the groups. The indication is, therefore, that the correlation between AI and these other variables is low.

High arch feet have generally been associated with inflexibility in the sub-talar joint and thus poor shock absorption while low arch feet are thought to be hypermobile (Subotnick, 1980). The high arch foot, therefore, should be less pronated during stance while the converse should be true with the low arch foot. The results of this study do not support this contention, however.

TABLE 1 - Mean(sd) values and statistical analyses for all variables

<u>VARIABLES</u>	<u>HIGH</u>	<u>NORMAL</u>	<u>HIGH</u>	<u>F-ratio</u>
<u>Stride Parameters</u>				
Arch Index-Static	19.86(2.78)ab	23.84(2.50)ac	30.39(4.34)abc	25.57*
-Dynamic	19.53(2.05)ab	22.83(1.27)ac	28.49(4.41)abc	19.11
Angle of Gait	9.40(4.13)	8.29(3.60)	8.31(6.31)	<1
<u>Static Measures</u>				
RF Supination	19.4(3.8)	19.6(1.9)	19.9(5.8)	<1
RF Pronation	8.3(1.9)	7.4(3.0)	6.6(2.4)	<1
<u>Rearfoot</u>				
TD Angle	-1.40(2.44)	-1.26(2.13)	-2.29(2.92)	1.19
Max. RF Angle	-7.29(2.78)	-7.28(2.54)	-7.58(2.87)	1.88
% time to max RF angle	46.29(5.62)	46.11(2.38)	46.93(6.29)	<1
<u>Knee</u>				
Max. flexion	19.28(3.56)	17.05(4.44)	21.23(3.27)	2.44
Max. int. rotation	3.03(3.06)	4.19(2.79)	4.47(3.15)	<1
Max. ext. rotation	3.51(1.67)	3.03(1.69)	3.67(2.53)	<1

* p<.05

Angle : degrees

% time : percent of support

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A GENERALIZED THEORY OF A STRUCTURAL ANALYSIS OF THE
GAIT BIOMECHANICS

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An integrated theory of structural couplings of the gait biomechanics has been suggested for the first time. The theory makes possible to perform a differential evaluation of the walking stereotypes based on a total combination of their biomechanical and physiological characteristics taken as a whole and to analyse purposive orientation of separate phases of a step as well.

For this purpose formalization of the laws of the structural couplings has been developed based on an automodel approach in relation of biomechanics of the walking. The locomotions have been considered as an act directed to the translation of a human body in space and in time and defined by the three dominant factors: rationality of energetics, specificity of the environment and the need to take into consideration the limitations induced by the structural properties of a supporting-motor apparatus.

The specificity of the environment while walking in norm has been manifested mainly through a profile of the supporting surface and conditions a number of peculiarities in the formation of processes of fitting and pushing-off the foot, flexion-extension of the knee and hip-joints and also bending oscillations of the trunk and swinging motions of the arms. In pathology some part of elements of a rehabilitation device, for example, a prosthesis, may be included into the elements of the environment. A general pattern of the gait is manifested as a system, the structural elements of which have been fixed in the form of kynematic, dynamic, temporal, energetical and some other features. The results of such kind registration comprise a number of regularities: dependence of the angles of the limb swing in a step length; velocity of movement is associated with the length and the rate of steps; dependence of the knee flexion angles in the foot flexion-extension and the width of a step.

To get an information about internal structural laws of the gait stereotypes is rather difficult that is associated with a necessity to solve a semi-reversed task of dynamics, i.e. to use with it the data with such degree of "cross-talk interference" that the correctness of the received results needs some distinct improvement. Just these very difficulties don't allow to define to a great extent fine structural and individual peculiarities of gaits.

According to the suggested approach, a process of formation of a double-support step has been considered as a set of relative transformations of elements of a base model of a supporting-motor apparatus from one extreme instantaneous position to the other one which takes place on a general background of a swing motion of the whole system. In this

case it becomes possible to study a set of rather similar border-line tasks which a man has to solve while walking. Predicting a character of a swing motion and modelling different versions of the programs of solving border-line tasks some different variants of the gait patterns have been attained according to the generalized parameters of control functions in an automodel mode. To analyze the gait in norm the limitations have been taken into consideration associated with conditions of anthropomorphism, continuity of kinematic links and real values of the force and energetical indexes. While modelling pathological gaits some additional limitations have been included reflecting a passive character of mechanisms in the articulations and units of the prosthesis as well as some peculiar features of the stump contacts with the socket.

The choice of one or another program in combination with necessary limitations has made it possible to establish a number of the laws of structural couplings distinguishing one stereotype of the gait from another one, as well as the requirements to meet their realization. Collection of the stereotypes received has allowed to clear recommendations for rehabilitation means of care and establish the classification norms for application of lower-extremity prostheses of different designs.

CSSP CONTINUOUS SIMULATION SOFTWARE PACKAGE - AN APPLICATION IN
LOWER EXTREMITY IMPACT LOADS SIMULATION

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INTRODUCTION

Modeling of human structure focused on analysis of its loads, overloads and possible injuries resulting from extensive motion activity, gains significant attention in biomechanical research (Cavanagh, 1970, Frederick et al., 1981, Nigg et al., 1984, Voloshin and Wosk, 1981).

The various of experimental and analytical methods are used in studying that problem. The presented paper shows the application of a new, dedicated for biomechanical simulation, software package CSSP (Continuous Simulation Software Package) (Siwicki, 1987) in analyzing the dynamics of human lower extremity during impacts. The package can be implemented on any of IBM PC compatible computers and is particularly useful in simulation of continuous and non-continuous and non-linear processes like human locomotion, jumps and impacts of multi-link bodies described in the general form of the first order differential equations in form:

$$\dot{Y} = F(Y, U, t)$$

where: Y - vector of state variables, U - vector of control variables and t - time

The package is user - oriented and supplied with dedicated symbolic language for easy formulation of model equations, boundary conditions and constraints. The CSSP is also supplied with extended graphic package which enables a user to create own real-time or off-line time dependent or phase graphs.

THE SIMPLE MODEL OF IMPACT

As the example of modeling and simulation of impact effects in human lower extremity with the use of CSSP, the simple model of jumping from the given height has been shown (Fig.1). The model has been described by the following set of differential equations:

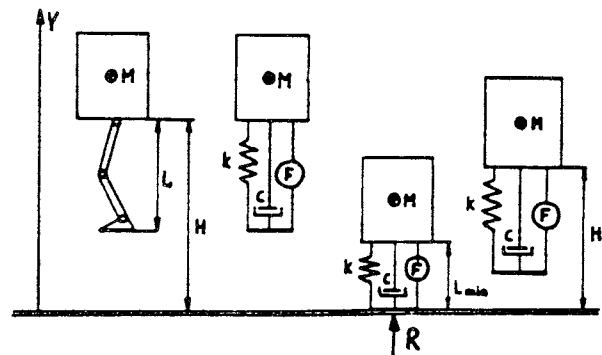


Fig.1

$$\begin{cases} m \ddot{y} + m g = 0 & y > H \\ m \ddot{y} + k \cdot \xi + c \cdot \dot{\xi} + F(\xi, t) + m \cdot g = 0 & y < H \end{cases}$$

where: m - mass of a body, k, c - stiffness and dumping coefficients of lower extremity passive elements, $\xi = L - y$ - relative shortening of lower extremity due to joint angles changes, $F(\xi, t)$ - force generated

by lower extremity muscular forces

In order to simulate and to study the best muscular "strategy" during the impact the following control characteristics have been also implemented:

$$F(\xi) = \begin{cases} 0 & y > L \\ (a_1 \cdot \xi^2 + a_2 \cdot \xi + a_3) (1 - \xi / \xi_{max}) & L > y > L_{min} \\ F_{max} & y < L_{min} \end{cases}$$

and

$$F(\xi, t) = F(\xi) \text{ when } t_i < t < t_f \text{ and } 0 \text{ otherwise}$$

where: a_1, a_2, a_3 - coefficients of polynomial approximation of muscular force limits measured under dynamic conditions, F_{max} - force generated by passive elements (ligaments) if maximum motion range in joints is going to be violated (L_{min}), t_i and t_f - time instants of muscular forces activation.

RESULTS

The vertical jump from the height of 1.1 m has been investigated as an example in order to find the best configuration of lower extremity (the angles in joints) just before impact and the best strategy of muscular action (t_i and t_f time instants) in order to minimize the loads (reaction forces) in lower extremity. Fig.2 show the transients of body mass center Y coordinate and velocity V_y during the landing while Fig.3 presents the simplified shape of ground reaction force $R(t)$.

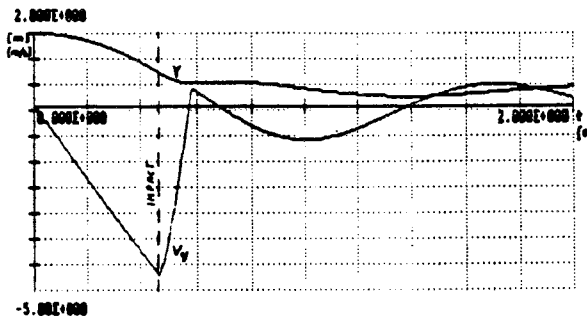


Fig.2

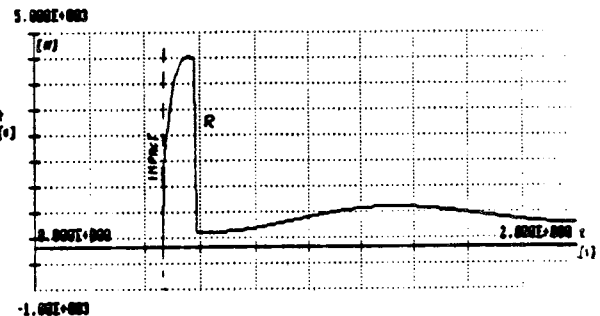


Fig.3

CONCLUSIONS

The software package CSSP was found very useful and easy to operate in studying various aspects of impacts of human body (i.e. the effects of body configuration during landing and muscles prestretching mechanism, the effects of ground and shoes dynamic properties, the best take-off strategies dumping transfer function in complex multi-link bodies, etc.). First experiments with the CSSP have also shown that it can be effectively used as an educational tool in biomechanics and biomedical engineering teaching.

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AN INTERACTIVE COMPUTER SIMULATION AND OPTIMIZATION OF MULTI-LINK MODEL OF SWIMMING MOTION

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INTRODUCTION

The ultimate objective of biomechanical research of swimming is to increase the understanding of human motion (in water medium) and to attempt an optimization of the movement in order to achieve the "best" sport result - minimum time of covering the given distance (Clarys et al., 1979, Di Pampero et al, 1974, Van der Vaart, 1987). The presented research puts the main stress on formulation of the advanced multi-link model of swimmer's dynamics and on development of interactive simulation package which can be used mainly in studying and identification of main limiting factors of individual swimming performance and in optimization of swimming motion technique.

MODEL

The crawl stroke model has been developed with the following assumptions taken into consideration:

- the model consists of four rigid elements i.e. hand, forearm+arm, trunk+head and lower extremity
 - the motion can be considered as planar however the 3-dimensional motion of upper extremities has to be included
 - the motion results from the action of two main muscles groups i.e. upper extremities (shoulder complex torques) and lower extremities (hip torques)
- The model and general forces acting on an athlete's body during the stroke are shown in Fig.1 (whole body) and in Fig 2 (upper extremity).

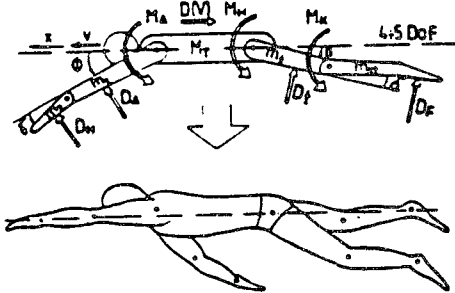


Fig.1

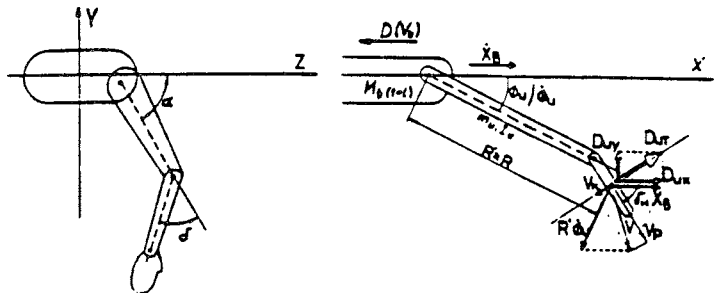


Fig.2

The model equations have been formulated basing on Lagrange's principle:

$$d/dt(\partial T/\partial \dot{q}) + T/\partial q + \partial V/\partial q = \bar{Q}$$

where: T - kinetic energy of the system, V - potential energy of the system, $\bar{Q}$ - generalized forces (water drag and muscular torques) and q - system variables

The muscular forces are subjected to the constraints (Komor, 1981) :

$$\bar{U}(t) < \bar{U}_{max}(q, \dot{q})$$

where : $\bar{U}_{max}(q, \dot{q})$ - polynomial approximation of maximum muscular torque limits measured under dynamic conditions

Considering the swimming performance optimization the following overall performance indexes have been also formulated:

[†] temporary address

T -> min - race time minimization or V_m -> max - mean velocity maximization

SIMULATION PROCEDURE

The SWIMMING interactive simulation package has been developed on IBM PC/XT/AT PS/2 personal computer which enables to investigate previously shown model and gives the wide possibility of:

- simulation of some real swimming patterns of a given athlete
- investigation of the influence of: selected anthropometric data, stroke initial conditions, muscular strength and control action on performance

The package is "user oriented" with extended graphics support and some additional tools which enable: easy modification of system parameters, stick figures graphics and up to 9 motion parameters transients graphics

SIMULATION RESULTS AND CONCLUSIONS

The example of swimming simulation real-time output is shown in fig. 3. The preliminary studies of the model have been focused on analysis of influence of anthropometric data and initial conditions on overall performance index. Considering on the example effects of changes of athlete's body mass it was found that reduction of standard mass of about 20% results in very small, 1% increase of mean swimming velocity, if other parameters remain the same (muscular strength, etc.).

However it was found that optimum variation of the hand angles results in increase of mean swimming velocity of about 3.7%. Finally an example of the influence of variation of muscular strength of shoulder complex has shown that 20% increase of athlete's shoulder strength abilities results in about 2.5% increase of mean swimming velocity.

The comparison done between simulation and experimental results shows that the model follows many major aspects of swimming motion with acceptable accuracy. The crucial point, if practical applications of computer simulation of swimming is considered, lies in precise evaluation of input data as: anthropometric data, athlete's strength data and drag coefficients.

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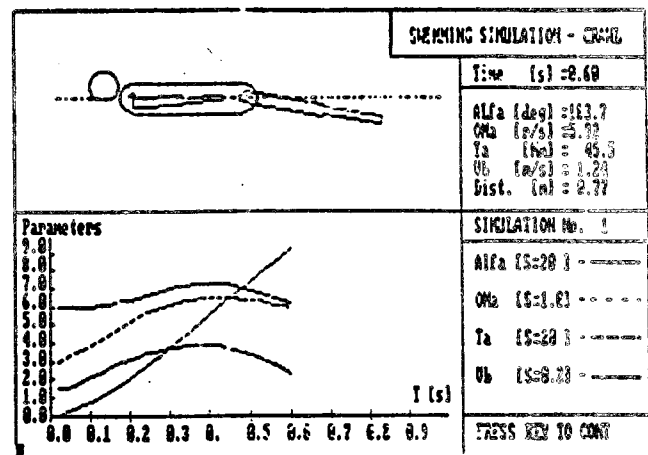


Fig.3

RELATIVE ROLE OF ANKLE AND METATARSAL-PHALANGEAL MUSCLES IN BALLET POINTE WORK

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INTRODUCTION

Almost half of the injuries in ballet today are to the foot and occur when ballerinas dance on the tips of their toes, that is on "full-pointe". Fatigue of muscles crossing the metatarsal-phalangeal joints is thought to be a causal factor. Tendinitis of flexor hallucis longus, acute intrinsic muscle spasm and repetitive muscle strain injuries at midtarsal are common in ballerinas and occur when manoeuvring to the full-pointe position (Ende and Wickstrom, 1982). Fatigue is also thought to be a major factor in fractures to the phalanges, metatarsals and sesamoid bones as acute fractures usually occur toward the end of the day when the dancers' feet are very tired (Sammarco and Miller, 1982).

The injury mechanisms involved in rising onto the tips of the toes suffer from a lack of biomechanical analysis. Only Galea (1983) has performed biomechanical analysis of pointe work; bone-on-bone forces at the m-p and ankle joints were found to be as high as twelve times body weight. This study was conducted to extend Galea's work and analyze the relative contribution of the ankle extensor group and the m-p flexor group to the energetics of the movement.

METHODS

An intermediate-advanced level dancer was the subject. Reflective markers were positioned over the hip, knee and ankle joint centres on the right limb. Four additional markers were placed on the lateral aspect of her right pointe shoe at the heel, sole, m-p joint and toe. The subject stood with her legs externally rotated, feet hip-width apart and with her right foot on the force plate. Each trial was recorded on video at a rate of 60 Hz while force plate data were collected on line at a sampling rate of 500 Hz. Digitized force plate data was coupled with digitized marker coordinates filtered at 6 Hz and used as input to an inverse dynamic model. The model, reported by Ishac and Winter (1988), featured the foot as two segments which permitted decomposition of the ground reaction forces into two components, one under each segment. The ankle and m-p joint powers as well as the foot (heel to m-p) and phalangeal (m-p to toe) segmental energies were calculated.

RESULTS

From the kinematic analysis it was found that there was a distinct pause when the subject extended the m-p joints maximally, rising as high as possible onto the ball of the foot before achieving full-pointe (Figure 1). A similar pause occurred during lowering.

Of the estimated 60 J of energy required to raise the centre of mass of the body 11 cm vertically upward, the model predicted from the integration of the moment-omega products that 26.2 J were generated from the m-p crossing muscles while 33.8 J came from the ankle plantarflexors. Thus, the percent of work done at the m-p joints was 44% versus 56% at the ankle joint in each of the two trials performed. Figure 2 summarizes the final integration of all energy changes from the beginning of the movement at flat foot to the top of the movement at full-pointe. Passive energy transfers across the m-p and ankle joints were calculated to be 25.9 and 60.2 J out of the respective segments. The change in potential energy stored in each segment was calculated based on the height of rise of the centre of mass of each segment.

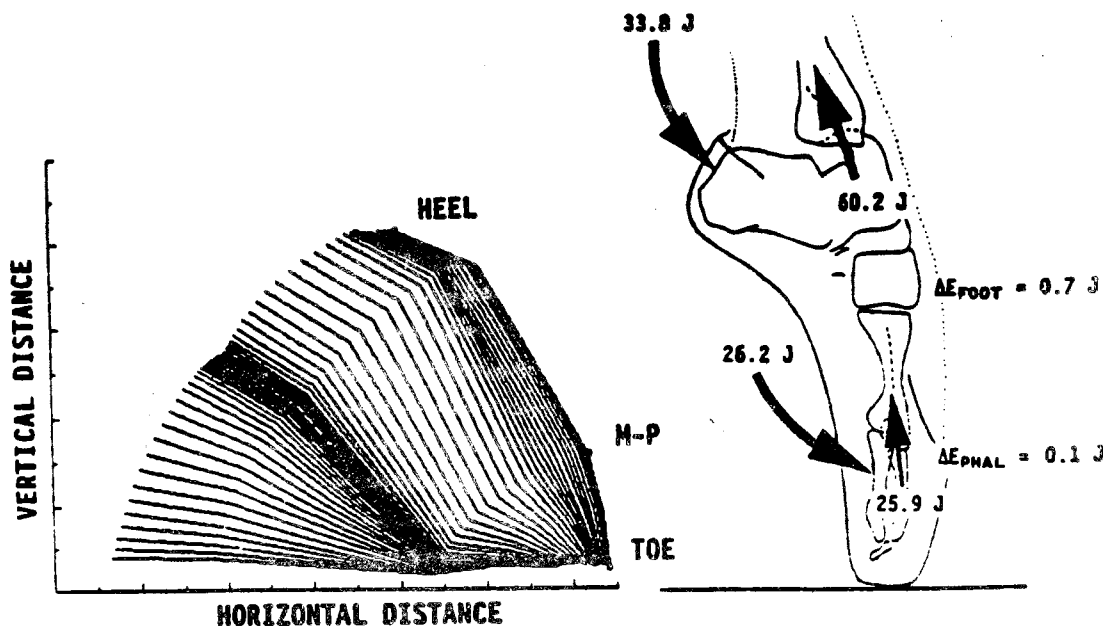


Figure 1

Figure 2

DISCUSSION

The ratio of physiological cross-sectional areas of the relevant muscles at the ankle joint versus the m-p joints is about 10:1 (Bo et al., 1980). Since the m-p muscles must generate almost half of the total energy required it is no wonder that injuries at the m-p joints are so prevalent.

The energetics of this movement explain the high incidence of fatigue-related injuries of the muscles crossing the m-p joints. The only way to reduce these injuries is to reduce the muscles' activity, which would also lead to lower bone-on-bone forces (Galea, 1983). The osteoarthritic changes of the feet linked with high forces may also be lessened. However, there are no alternative muscles available to assist the phalangeal segment in lifting the entire mass of the body vertically upward. Perhaps, redesign of pointe shoes is a solution to help solve this problem.

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ACKNOWLEDGEMENT

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TECHNIQUE FOR INTERPRETATION OF EMG FOR CONCENTRIC AND ECCENTRIC CONTRACTIONS IN GAIT

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INTRODUCTION

Knowledge of muscle activity during gait is commonly ascertained from processed or raw EMG profiles. The tension in a muscle has been interpreted from processed versions of the EMG, such as the linear envelope (LE) (Olney and Winter, 1985). However, simple LE processing does not account for force changes due to the rate of lengthening or shortening of the muscle (Lippold, 1954). The muscle velocity profile also indicates whether the muscle is acting as a generator or absorber of energy. This paper presents a computer graphics technique to provide information on both the muscle fibre velocity and muscle activation profiles which enhances the interpretation of muscle activity in both normal and pathological gait.

METHODS

Several representative muscles of the lower limb were analyzed including the tibialis anterior, soleus, lateral and medial gastrocnemius, vastus lateralis, rectus femoris and semitendinosus. Activation profiles were the LE with a 3 Hz critically damped filter, chosen specifically to model the twitch response of most muscles of the lower limb (Winter and Yack, 1987). The EMG profiles in this study are the normalized average values from the large data base at the University of Waterloo Gait Laboratory (Winter, 1987).

A model was constructed to measure the muscle-tendon lengths using straight lines between the muscle origin, insertion and constraint locations for different joint angle configurations. Third order polynomials fitted the ankle, knee and hip joint angles to the muscle-tendon length for each muscle. The muscle fibre velocity was determined by combining changes in the muscle-tendon lengths with the assumption that the muscle is incompressible. When the muscle length is above or below resting fibre length, the angle of pennation of the muscle increases or decreases, respectively and follows experimentally determined variations in fibre length/pennation data (Huijing and Woitiez, 1984). This results in a lower fibre velocity than estimated with a fixed angle of pennation.

Muscle fibre velocities below .25 resting lengths/second (lo/s) were considered to be a near-isometric state, while shortening and lengthening velocities greater than .25 lo/s were classified as concentric and eccentric contractions, respectively. Fast and slow contraction velocities were divided at 1.5 lo/s, thus improving the velocity data output resolution. The fibre velocities in this paper were calculated from averaged kinematic data from the University of Waterloo Gait Laboratory (Winter, 1987).

The graphics program plots the normalized EMG profile for the gait stride and colours below the waveform based on contractile state: dark green for fast concentric, light green for slow concentric, yellow for isometric, light red for slow eccentric and dark red for fast eccentric. However, the results are displayed in this paper using slanted (/ / /) for concentric, void for isometric and black slanted (\\ \\) for eccentric.

RESULTS

The computer plot of the soleus muscle is shown in Figure 1 where stance time is from 0 to 60% of the total stride. Peak activity occurs concurrently with concentric contraction during push-off and shows the soleus to be used primarily as a generator of energy during gait. The

EMG profile displays that there is also significant activity to control forward rotation of the leg over the foot. Combined with the lengthening of the muscle at this time, the soleus is also seen to absorb a significant amount of energy.

The rectus femoris muscle EMG and velocity plot are presented in Figure 2. The major burst of activity begins before HC and continues through weight acceptance. Simultaneous muscle lengthening demonstrates that the rectus femoris is primarily used as an absorber of energy during gait. The second wave of activity occurs during push-off and is eccentric control of knee collapse. This activity continues after toe off to arrest the backward leg movement, and then accelerate it forward during mid and late swing. The rectus femoris, unlike the vastus lateralis, is not seen to be concentric during mid stance as the knee extends and this is because it simultaneously lengthens at the hip at the same time.

DISCUSSION AND CONCLUSION

The calculation of the fibre velocity can be effectively determined using joint angle data and assuming that the angle of pennation varies with fixed muscle volume. The display of the velocities augments the EMG profile by providing simultaneous information on the activity level of the muscle, as well as whether the muscle is shortening or lengthening. This provides insight as to the role played by each muscle as a generator or absorber of energy which is useful in a clinical environment as a quick and effective analysis tool of EMG and muscle velocity data.

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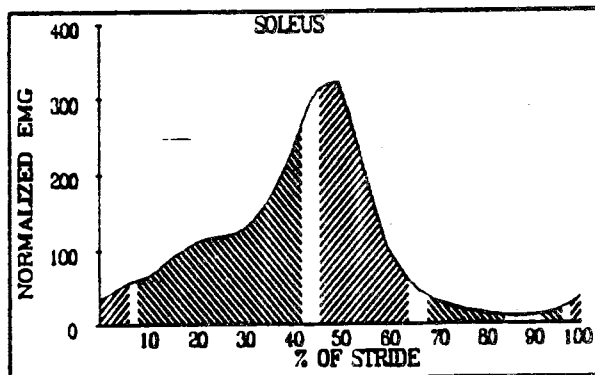


Figure 1

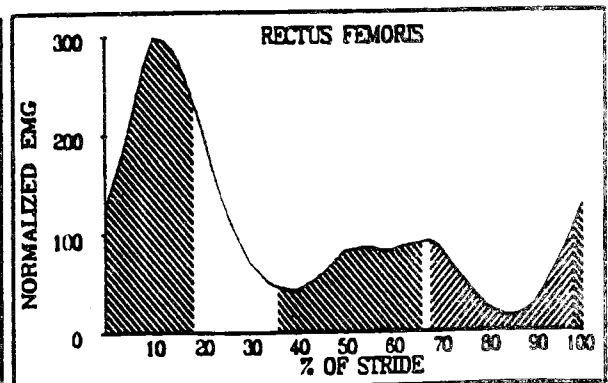


Figure 2

SEGMENT-LINES MODELING OF HUMAN ELBOW JOINT DRIVE

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INTRODUCTION. Two basic problems are in the focus of attention when modeling the human joints - a good reflection of the real structures and functions and a suitable choice of the mathematical approach. The free body diagram usually contains straight-lines connecting with the muscle origins and insertions and presenting the directions of muscle forces [1]. As it is shown in [2] such approach should be treated with caution. The centroid lines describe well the muscle action lines, but this method is related with different difficulties [3]. For calculation of the muscle forces the equation of equilibrium or of motion are used. As the number of unknown forces exceeds the number of equations, the problem has to be solved using functional grouping, systematic elimination or optimization method. The linear programming method with various objective functions is often used with different success [4]. The aim of the paper is to describe more adequately the muscle action lines bearing a relation to the elbow joint during flexion-extension and using the Lagrange multipliers method to calculate the muscle forces. The proposed segment-lines modelling is compared with straight-lines modelling on the base of three different rotation drive gears.

TWO-DIMENSIONAL MODEL. The two-dimensional model of elbow

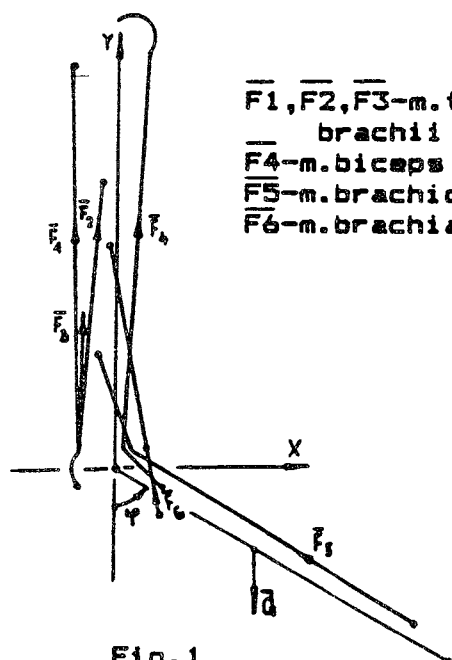


Fig. 1

simple software, the force solutions are of analytic type, the coefficients c_i are independent of the model geometrical parameters and the force values are guaranteed to be positive.

flexion-extension presented on Fig. 1 contains four segment-lines modeled muscles. Basic conditions: the moment of every muscle to keep permanent direction for the whole motion range; the moments of all synergistic muscles to have identical direction and opposite to the direction of antagonistic muscle moments. The Lagrange multipliers method with objective function $Z = \sum c_i |F_i|^2$ is applied for muscle force calculation. The model is analyzed under quasistatical (Fig. 2) and dynamical conditions for different values of the weight coefficients. This method needs

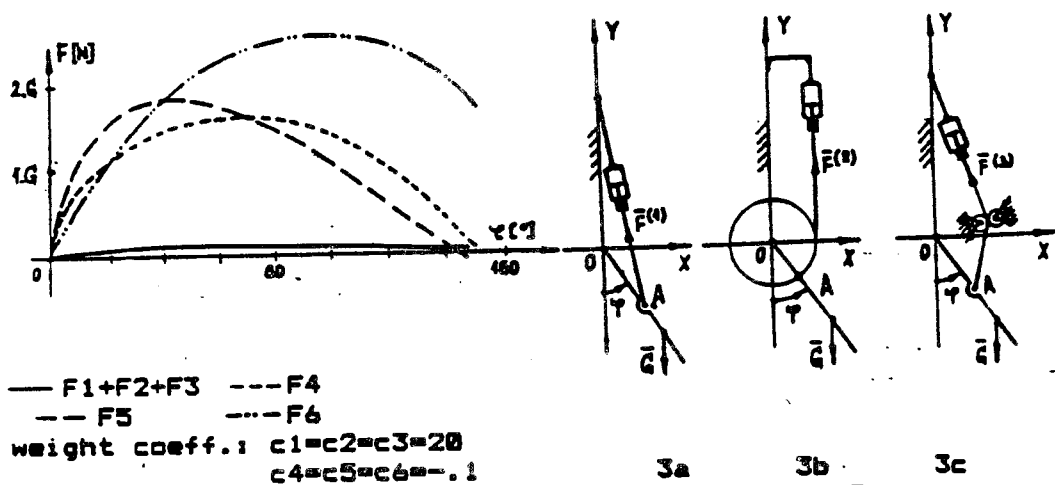


Fig.2

Fig.3

COMPARISON OF THREE MODELS OF ROTATION. A comparison is done between straight-lines modeling (Fig.3a) and segment-lines modeling (Fig.3b-m.triceps brachii, Fig.3c-m.biceps brachii). The geometrical and dynamical characteristics of the three models are identical with those of the model on Fig.1. The forces necessary for equilibrium supporting or for motion performing are calculated and compared with for estimation of the rotation models from power point of view. The motion law for the links OA (Fig.3) is $V^i = A^i \sin(\frac{\pi t}{T})$ $i=1,2,3$; $\gamma=0[^\circ]$ when $t=0[s]$; $\gamma=180[^\circ]$ when $t=T[s]$. Under quasistatic conditions the force $F^2(\gamma)$ is the most uniform and the force $F^1(\gamma)$ has the smallest maximum. Under dynamic conditions the force $F^1(\gamma)$ is the smallest for big $\dot{\gamma}(t)$ and $\gamma>0$. When $\dot{\gamma}(t)$ is near zero - $F^2(\gamma) \ll F^1(\gamma)$. The model on Fig.3b can perform only a part of the given motion. The proposed approach allows to analyze the muscle cooperation during different motions and to use some biomechanical structures and functions for technical system design.

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THE USE OF STORED ELASTIC ENERGY: DIFFERENCES BETWEEN
VERTICAL JUMP AND JUMP WITH THROW OF THE BALL FOR FEMALE

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Introduction

This study was designed to further examine the performance in vertical jumps and jump with throw of the ball. One way to measure the elastic potential of the leg extensor muscle is to use maximal vertical jumps on the force-platform with or without counter movement.

It was also thought, that this technique might be of use to investigate the possible differences in the storage capacity and utilization of elastic energy between countermovement jump and jump with throw of the ball. The possibility that muscle stores and utilizes elastic energy depends on the mechanics of exercise.

The purpose of this study was to examine the possible differences in the storage capacity and the utilization of elastic energy between countermovement jump and jump with throw of the ball in place on a force-platform for females.

Methods

The subject of this study were 12 basketball players from the females national team. All the subjects tested were familiar with the test procedure used. This consisted of performing maximal vertical jump from static, squatting position (SJ), countermovement jump (CMJ) and jump with the ball throw (JT) on the Kistler force-platform with an amplifier and microcomputer Neptune 184 with an A/D converter. The vertical force-time curve produced by each jump was the basis for computerized calculation of various parameters such as maximal vertical acceleration during take off phase (a), vertical take off velocity (v), height of rise of the body mass center (h), depth of the body mass center lowering (d), duration of the take off phase (t), maximal external power (P_{max}), average power ($\bar{P}$), potential energy of the body mass center during eccentric phase (E_e), potential energy of the body mass center during take off phase (E_c) (concentric), effective integration of legs (EIL), and use of stored elastic energy (USEE). The USEE was obtained from the formula [Asmussen & Bonde-Petersen, 1974].

Statistical methods were employed to calculate the mean values (mean), standard deviation (SD), and linear correlation coefficient (r). Differences between mean values were tested for significance using the dependent t-test.

Results

The values of measured parameters (mean $\pm$ SD) for static jump (SJ), countermovement jump (CMJ), and jump with throw of the ball (JT) are given in Table 1. With the exception of maximal external power, no statistically

significant differences were found between parameters of CMJ and JT. Without exception CMJ and JT produced a much higher maximal vertical acceleration during take off phase (a) and average power ($\bar{P}$) than SJ. There was a difference in USEE between CMJ and JT, but the difference between the jumps was not statistically significant ($p < .05$). Correlation analysis revealed that several mechanical parameters were correlated with the USEE. For example the value of USEE of JT was correlated positively with h of JT ($r = .577$, $p < .05$), and with EIL of JT ($r = .955$, $p < .01$). Similar correlation coefficient is observed between USEE of CMJ and EIL of CMJ ($r = .944$, $p < .01$).

Conclusion

Even though improvement of performance after prestretching has been attributed to a combination of the use of elastic energy and stretch reflex potentiation, one cannot overlook the role of the neural-muscular co-ordination during the exercises.

It appears that there are not statistically significant differences in use of stored elastic energy between countermovement jump and jump with throw of the ball.

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Table 1. General variables (mean $\pm$ SD).

Parameters	SJ	CMJ	TJ
a [m/s ²]	1.40 ± 0.26	1.73 ^{**} ± 0.22	1.65 ^{**} ± 0.36
h [m]	0.363 ± 0.048	0.370 ± 0.050	0.366 ± 0.054
d [m]		0.197 ± 0.052	0.178 ± 0.050
t [s]	0.261 ± 0.057	0.190 ^{**} ± 0.040	0.197 ^{**} ± 0.040
Pmax [W]	1907.5 ± 313.1	2212.5 ^{**} ± 245.0	2006.7 ^{**} ± 284.0
$\bar{P}$ [W]	903.3 ± 210.0	1248.4 ^{**} ± 182.6	1179.0 ^{**} ± 178.7
USEE [%]		2.7 ± 26.6	0.1 ± 34.0

Notation: a - maximal vertical acceleration during take off phase, h - height of rise of the body mass center, d - depth of the body mass center lowering, t - duration of the take off phase, Pmax - maximal external power, $\bar{P}$ - average power, USEE - use of stored elastic energy, x - the value during CMJ or JT differs from the value during SJ ($p < .05$, ^{**} - $p < .01$), ^{**} - the value during CMJ differs from the value during JT ($p < .01$).

A NEW CONCEPT ABOUT THE SUPPORTING ORGAN OF THE MUSCULOSKELETAL
SYSTEM

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A principally new approach in biomechanics in the hip joint is proposed. It is assumed that: i - bones are not capable to bear alone loads, producing great tensile stresses in them; ii - some muscles, apart from their well known functions, play also an important role, promoting in turn the loadbearing capacity of the bones; iii - the cortical tissue takes up mainly the compressive efforts; iv - the cancellous structure represents a specific intermediate tissue, preserving the great tensile strenght of the soft tissue from which it originates, and acquiring the compressive strenght of the cortex it gives origine to.

Proceeding from these assumptions it is postulated that the supporting organ of the musculoskeletal system constitutes an "adaptive supporting complex", including apart from bones, the respective "supporting" muscles also. This new fundamental postulate is warked out by biomechanical re-interpretation of well known data from anatomy of the human hip area.

As it is well known the abductors - gluteus medius and minimus - are powerful muscles, promoting the balance during unilateral support. The attachment site of these muscles is on the lateral surface of ala ossis ilii. The latter represents an externally convex shell, exceptionally thin at fossa iliaca level, not capable to endure great tensile efforts, produced in it by the action of the abductors. In this respect the bone is assisted by the iliacus muscle, whose attachment site is on the inner aspect of fossa iliaca. Apart from being a hip joint flexor, by the presumed function the iliacus becomes a "supporting" muscle of the pelvis bone. The latter statement is corroborated by the synchronous EMG-action of gluteal an iliacus muscle, documented by Patriarco et al (1981). Moreover, in its distal end the iliacus muscle rests tightly on the femoral head outside the acetabulum and after gliding beneath the femoral neck, terminates at the posteromedial surface underlying the lesser trochanter. Its curvilinear course supports anteriorly and from below the femoral head and the neck in anteversion, thereby cousing their backward

bending. The latter phenomenon was established by Brown et al (1982), but no explanation was offered by them. Obviously the iliacus muscle is a "supporting" muscle of the femoral neck too which, considering its cancelous structure, could hardly be in a position to endure alone the loading on the femoral head, measured by classical methods.

The external and internal obturators are likewise considered as "supporting" muscles of the femur. In the human hip area these are the only muscles, whose origin lies beneath their insertion. The external obturator originates from rami superior and inferior ossis pubis and ramus ossis ischii, and terminates at a higher point on the posterior surface of the femoral neck within fossa trochanterica. The curvilinear course of its superior fibres secure neck support from below, while the fibres coming from ramus ischii serve as a hanger on which the minor pelvis is suspended. The internal obturator has an analogous function. It is evident that the loading force, originating from ala ossis ilii is distributed within the cancellous tissue overlying the acetabulum into three parts - one part accounts for the direct loading of the femoral head, but the remaining two are transmitted through os pubis and os ischii to the obturators, which in turn transfer them into their own attachment site. It's more, fossa trochanterica lies along the longitudinal axis of the femur and that part of the hip joint force, transferred at this point exerts no bending effect on the neck and diaphysis. The above statement is confirmed by the fact, that the axis of the neck anteversion coincided with the longitudinal axis of rami ossis pubis.

As clear shown by the results obtained, the femoral neck should be no longer considered as a free cantilever.

A SYSTEM FOR MEASUREMENT OF MECHANICAL POWER IN STRETCH - SHORTENING TYPE EXERCISES

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The measurement of the concentric power and work during plyometric type exercises is of importance for evaluating the elastic energy contribution in stretch-shortening type exercises. It is wellknown that the power $P(t)$ can be determined from the force platform data by means of the relation : $P(t) = R(t) \cdot V(t)$ where R and V are the ground reaction force and the velocity of the center of gravity. Meanwhile this method needs often additional kinematic data specifying the initial velocity of the center of gravity.

The aim of the present work was to realize a data acquisition and processing system devoted to a rapid and automatic measurement of the mechanical power in jumping, for field and Laboratory conditions. Actually, the datamation can be made either on the field, the results being displayed on a LED System, or in the Laboratory on a terminal or a printer as shown below.

MUSCULAR MECHANICAL POWER MEASURE

Subject 02

Time in the air (s)	on the ground (s)
0.49	0.40
0.47	0.35
0.40	0.35
0.46	0.39
0.46	0.42
0.40	0.42
0.39	0.51
Mini on the ground : 0.35 s	
Maxi : 0.51 s	
Mini in the air : 0.40 s	
Maxi : 0.49 s	
Accumulated times	
3.07 s	2.84 s
Greatest hight reached : 294 mm	
Average power : 21.91 W/Kg	

The theoretical principles of the calculation of the average concentric power $\bar{P}^+$ per mass unit in a serie of vertical rebound jumps have been given in details by Bosco. (Bosco, 1983).

$$\bar{P}^+ = \frac{1}{4} g^2 \frac{T_f(T_f + T_C)}{n T_C}$$

where T_f and T_C are respectively the total flight and contact times for n jumps.

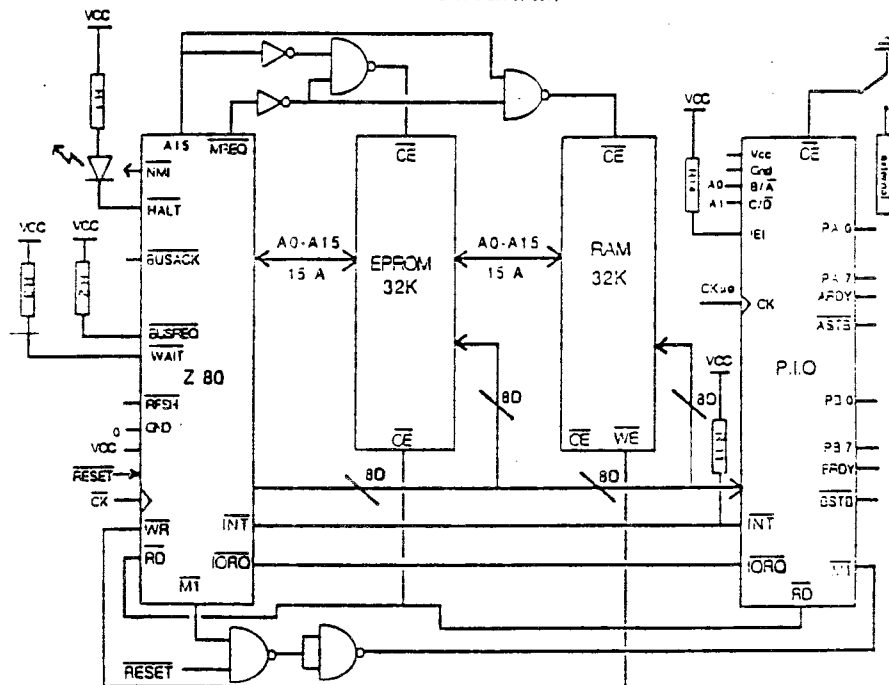
Technical description and block diagram

Data acquisition, output and commands are transmitted on the line of the 2 ports of the PIO. The port A is programmed in input and interruption and permits :

- to generate a digital timer ($\pm 1/100s$) by counting the interruptions of a 3.268 MHz quartz oscillator associated with a 14 stages divisor (4060)
- to sample the electric state of the switch sensor on the bit 0
- to validate the data acquisition on the bit 4, to start the data processing and to print the results on a parallel printer (centronics type), or a serie type (RS 232), or on a 6 digits 7 segment LED display. The subjects are numbered from 1 to N, and the results output of a given subject is selected by an encoding wheel. The output peripheral is automatically selected on the bits 6 and 7 by means of the peripheral plug. Data are transferred on the 8 bits of the PIO B programmed in interruption for the centronics printer. For the RS 232 printer, data are transferred on the bit 6 at a level from -5V to +5V and a baud rate of 300. For the LED display a synchronous type serial type is used by the bit 0 for the clock and 1 for the data. In addition a "Minitel" peripheral type connected before the driver can be used at 1200 bauds. The main parts of the system are shown on the block diagram.

This device allows a rapid and automatic test of the mechanical power during a serie of jumps ; it can be used on the field without additional equipment. As the main collected data are the flight and ground contact times, this system is also suitable for collecting all these times during distance running, or hurdling, using electrical switches inside the shoes.

BLOCK DIAGRAM



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ABSTRACT

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TITLE EFFECT OF ARM SWING AND VARYING KNEE ANGLES ON PERFORMANCE IN STANDING JUMPS.
AUTHOR(S) Dr. Tejinder Singh Brar
INSTITUTION Lakshmibai National College of Physical Education, GWALIOR -474 002 (MP) INDIA

Purpose : The purpose of the study was to investigate the effects of arm swing and varying knee angles during the take off, on performance in standing broad jump and vertical jump.

Method : One hundred randomly selected college students between seventeen and twenty two years of age served as the subjects. The performance of the subjects in standing broad jump and vertical jump was recorded with and without arm swing, maintaining three different knee angles i.e. sixty five, ninety and hundred fifteen degrees. The angles were measured with the help of a goniometer and use of a dip regulator ensured proper angle while taking off.

Results : The data was analysed by two way analysis of variance at .05 level of confidence. The results of the study indicated that performance with arm swing was significantly better than without arm swing in standing broad jump and vertical jump ($F=612.5$ and 436.2 respectively). Significant differences were found in standing broad jump performance as a result of varying knee angles ($F=6.41$).

Conclusions : It was concluded that arm swing contributed significantly to performance in standing broad jump and vertical jump. Best performance in standing broad jump can be achieved by keeping knee at ninety degree angle while taking off.

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KINETICS IN STANDING LONG JUMP OF 3 THROUGH 59 YEAR OLD
MALES AND FEMALES

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The purpose of the present study was to investigate the changes of the impulse and average force generated during the propulsive phase in standing long jump of 3 through 59 year old males and females.

Methods

213 males and 174 females performed standing long jump on the force plate (KISTLER, 2981-B11). All jump motions were filmed from a side view with using a 16mm cine-camera (Photo-sonics, 16-1PL) operating at a sampling rate of 35-50 frames per second. The jump performances were obtained from 16mm films. The following variables were calculated from the horizontal and vertical ground reaction force-time curves.

I[h]: Horizontal impulse of the propulsive phase

I[v]: Vertical impulse of the propulsive phase

F[h]: Horizontal average force of the propulsive phase

F[v]: Vertical average force of the propulsive phase

T[h]: Horizontal time of the propulsive phase

T[v]: Vertical time of the propulsive phase

Results and Discussion

The means and standard deviations of the I[h], I[v], F[h], F[v], T[h] and T[v] for each age groups of males and females are presented in Figure 1. In males, I[h] increased from 3-4 years to 18-19 years and after which it decreased. In females, I[h] increased from 3-4 years to 20-29 years and after which it decreased. The males exerted significantly larger I[h] than that in the females from 15-16 years old to 50-59 years old ($p < 0.01$). I[v] for both sexes showed an increase from 3-4 years to 20-29 years, and after that it showed a decrease. The males exerted significantly larger I[v] than that in the females at 17-18 years and 20-29 years age groups ($p < 0.01$). In T[v] and T[h], there were no significant differences with respect to age and sex from 3-4 years to 50-59 years.

These results suggest that the age and gender differences in impulses may depend on average forces.

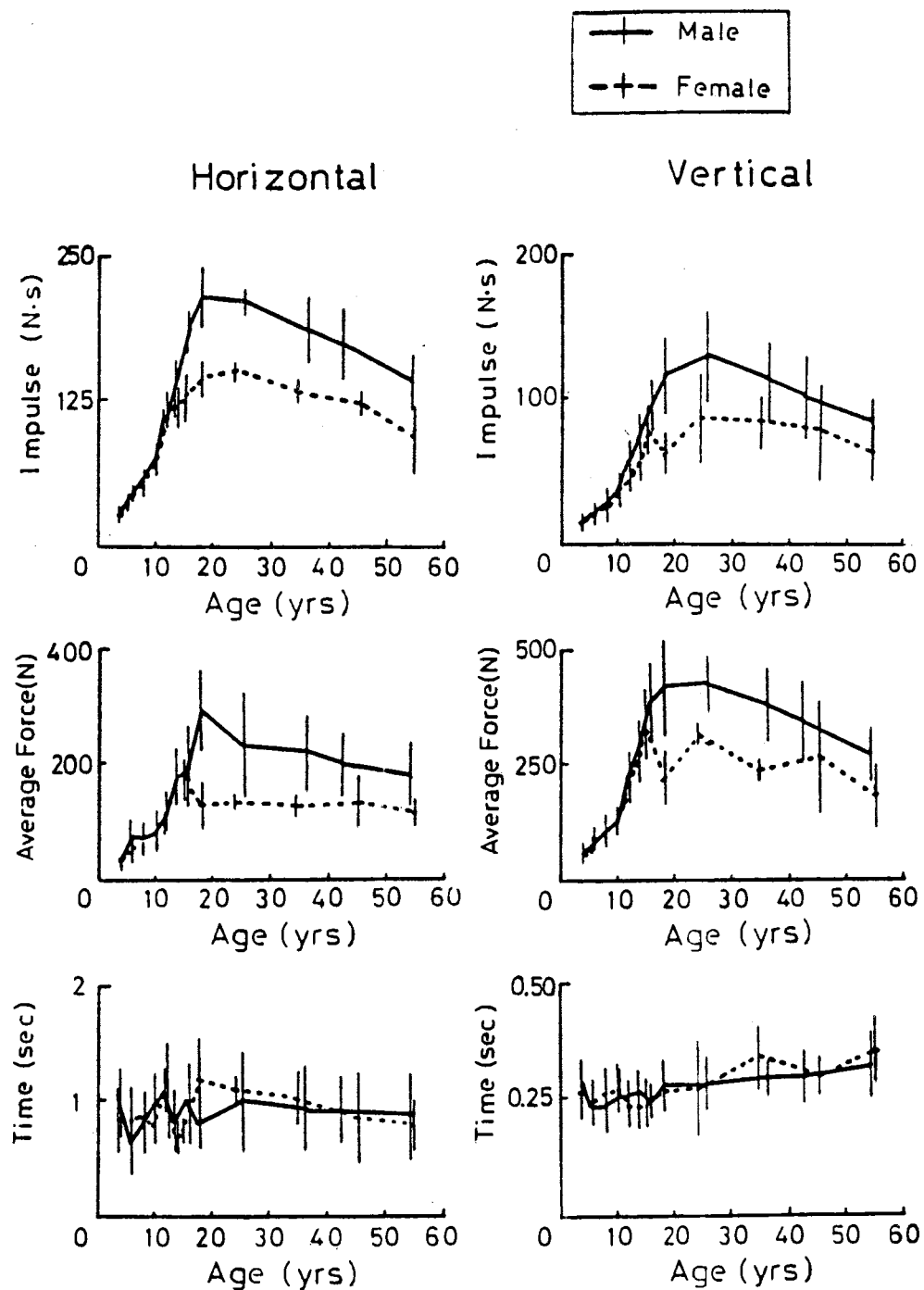


Figure 1. The means and standard deviations of the $I[h]$, $I[v]$, $F[h]$, $F[v]$, $T[h]$ and $T[v]$ for each age groups of males and females

ACHILLES TENDON FORCE AND EMG OF TRICEPS SURAE
DURING ANKLE HOPPING

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ABSTRACT

Mechanical behavior of tendomuscular system during ankle jumping was studied by direct measurement of in-vivo achilles tendon force (ATF) in man. A buckle-type transducer was implanted under local anaesthesia around the right achilles tendon of an adult subject. After the calibration procedure, in-vivo ATFs were recorded during ankle jumping on a force-plate together with the triceps surae EMG activities. Five jumpings were used: a) ankle squat jumping with the orthopedic cast that fix the knee joints(SJ), b) submaximal two legs hopping without cast (H-1), c) submaximal two legs hopping with cast(H-2), d) maximal two legs hopping with cast(H-3), and e) submaximal one leg hopping without cast(H-4). In all jumps the subject kept his hands on his hips.

Peak ground reaction forces (GRF/leg) and peak ATFs were 847N and 1335N in SJ, 1518N and 3786N in H-1, 1328N and 2258N in H-2, 2141N and 4868N in H-3, and 2376N and 6441N in H-4, respectively. In repetitive hoppings greater GRFs and ATFs generated compared to the maximal SJ defined as pure concentric movement. On the other hand, EMG activities in SJ and hoppings were almost same. Considering the results on ATFs and EMGs, the elastic characteristics in tendomuscular system was observed in the hopping defined as typical ballistic movement.

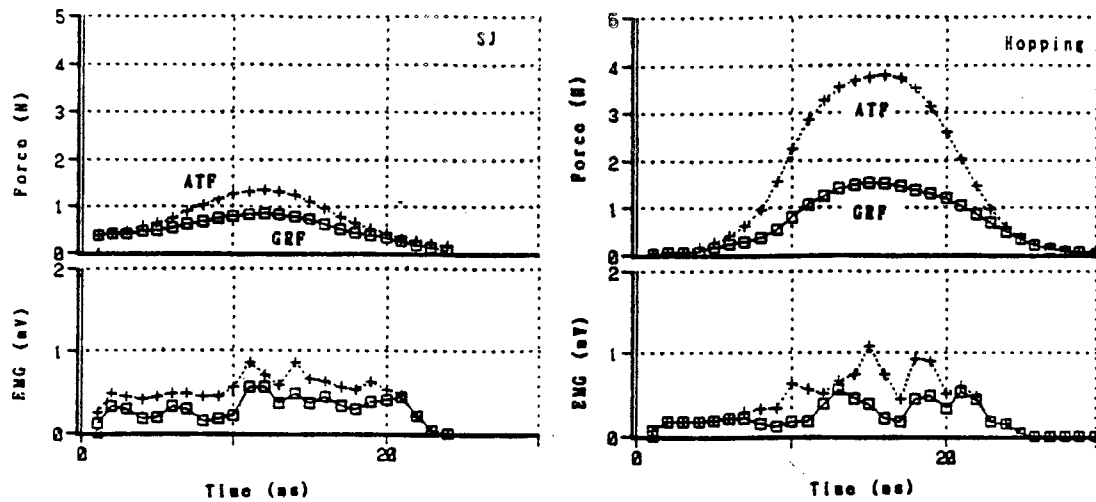


Figure 1. Ground reaction force (GRF) and Achilles tendon force (ATF) during ankle squat jump (SJ) and repetitive hopping (H-1).

* □ : EMG of Gastrocnemius
 + : EMG of Triceps Surae (Gastrocnemius + Solues)

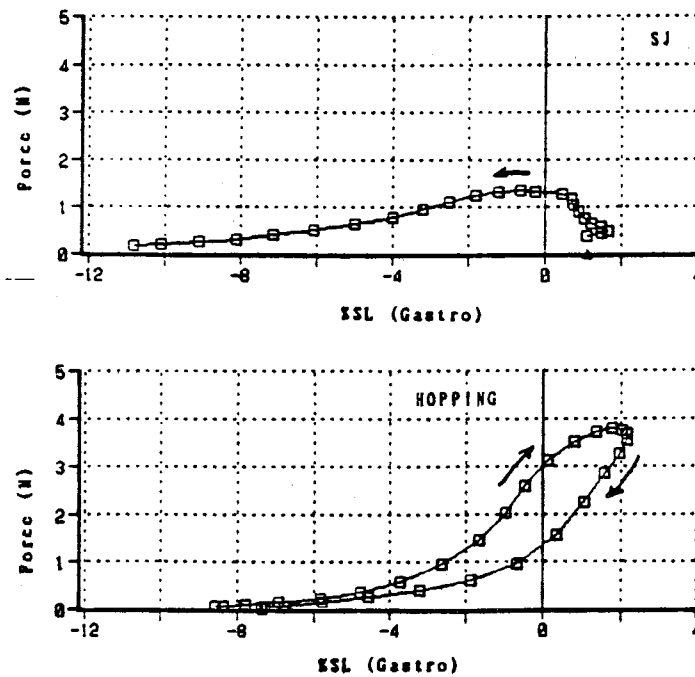


Figure 2. Force - Length curves during SJ and Hopping(H-1).

* Force: ATF
 Length: %length of Gastrocnemius

**Effectiveness of arm movement on performance
of the maximum vertical jump**

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Introduction

One of the property of skilled movements is that not only the directory concerning limb but also the other free limbs were enrolled for resolving a given task with the best performance and/or the least energy expenditure. Although movements are archived by a directly effective limb, skilled movements are always associated with simultaneous activity of the other free limbs. The purpose of this study is to determine the effectiveness of an elbow extension movement on the performance of the maximum vertical jump.

METHODS

Five subjects performed four kinds of the maximal vertical jump keeping their knee and hip joints fully extended. A jump was performed from a quiet upright standing posture without any arm movement. Another jump was associated with an elbow extension movement (EEM) which was asked to extend both elbows along with a lateral side of the body from an initial posture with both elbows flexed and put close to the lateral side of the body. These two types of jump were made under the conditions with no loads or holding 4 kg dumbbells. Our subjects were instructed to extend their elbows after the takeoff. Ankle joint angle, platform force, vertical head and hand accelerations, and takeoff and touchdown of the foot were measured with potentiometer, strain-gauge, accelerometer, and electrical switch, respectively, with simultaneous recording of electromyograms of triceps surae and tibialis anterior muscles using surface electrodes.

RESULTS

The jumps associatd with the EEM showed an increased flight time whether loaded or not, although its degree was different among the subjects, depending mainly on the timing of the EEM. Figure 1 shows averaged recordings of the head and hand accelerations and foot contact signal in the jumps accompanied by the EEM and without it, when a subject was loaded by the dumbbells. The foot contact signal shows that the EEM elongated the flight time measured by the time interval between the takeoff and touchdown. The head

acceleration with the EEM showed an increase of the upward acceleration after the takeoff, which means that the head was effected by force in the upward direction. The timing and shape of the incremental acceleration were comparable with the downward acceleration of the hand. Since the deceleration of the once accelerated hand occurred close to the touchdown, some part of the deceleration occurred after the touchdown. The platform force at the takeoff and the integrated electromyogram of the triceps surae muscle did not show any difference between the jumps with and without the EEM.

Conclusion

It was suggested that the performance enhancement associated with the EEM is caused by the transmission of force produced by the downward acceleration of the arm mass, and some of the force produced by the deceleration of the arm mass would not decrease the performance because the deceleration partly occurred after the touchdown.

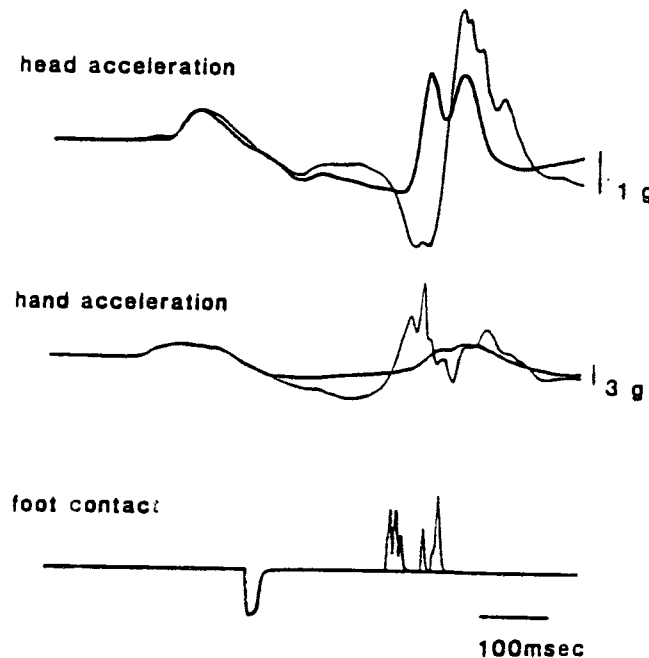


Figure 1 Averaged recordings of the head (upper traces) and hand (middle ones) and the takeoff and touchdown of the ball of the foot (lower ones) when a subject was loaded by dumbbells. Thin and thick traces show the jumps associated with the EEM and without it, respectively. Data were triggered by the takeoff of the foot (the wide downward pulse). Touchdown was shown by the average of the thinner pulses.

PRESSURE DISTRIBUTION PATTERNS INSIDE OF A RUNNING SHOE DURING UP- AND DOWNHILL RUNNING

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INTRODUCTION

The knowledge of the pressure distribution under the foot during dynamic locomotor activities is necessary for the understanding of the foot function. The pressure distribution inside of the shoe was measured during up-, down- and level running on a treadmill.

METHODS

Twelve healthy subjects were instructed to run on a "Woodway" treadmill uphill, level as well as downhill at given speeds of 3 m/s and 4 m/s (8% inclination). One type of running shoe with shock absorbing and stabilizing elements in the lateral heel, medial midfoot and lateral forefoot region was chosen for all subjects. Eight piezoceramic sensors (5 mm x 5 mm) with properties as described by Hennig et al. (1982), were taped onto the plantar surface of the foot at defined anatomical landmarks. The pressures were measured under the following foot structures: the lateral (LH) and medial (MH) heel, the lateral (LM) and medial (MM) midfoot, the first (I.M), third (III.M) and fifth (V.M) metatarsus and the hallux (H). The peak pressures and impulses from six trials in each condition were calculated for all subjects. From the eight local impulses a relative local load was determined by dividing each single impulse by the sum of all impulse values.

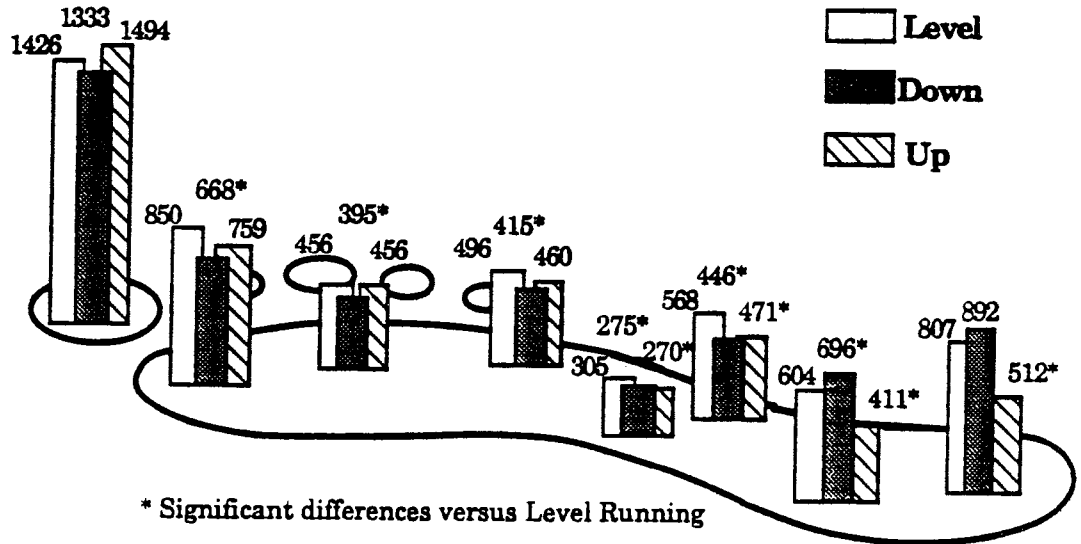
RESULTS

The statistical evaluation of the peak pressures and relative loads revealed significant differences between the three conditions in various regions of the foot. Figure 1 shows the means of the values of 3 m/s and 4 m/s. No significant differences could be found between the running speeds.

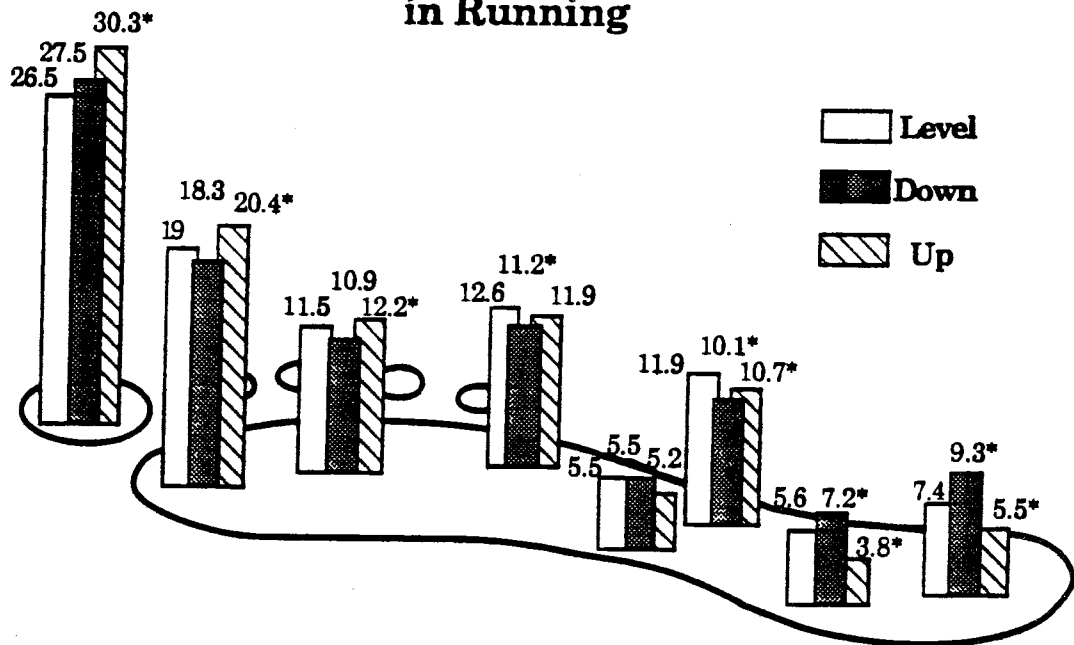
DISCUSSION

For all running conditions the large relative loads under the hallux and the first metatarsal head show the important function of the first ray to propel the body in running. As expected peak pressures and relative loads for downhill running were increased in both heel regions and reduced for the forefoot. Surprisingly small peak pressure differences were found between level and uphill running in the forefoot region, although a substantial reduction of pressures could be observed for the heel and midfoot region. However, the relative impulse graph demonstrates an increase and a shift of the load to the medial forefoot and the hallux. Unexpected was the substantial decrease of the load for in the lateral midfoot region for downhill as well as uphill running. Increased foot muscle activity may be responsible for a raised height of the longitudinal arch. Between subjects high variabilities were found for all regions. This finding indicates a pronounced individual running style of the subjects.

Peak Pressure Differences (kPa) in Running



Relative Load Differences (%) in Running



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FLEXION TORQUE IN ANKLE JOINT MEASURING SET

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Introduction

Measurements of torques in static conditions are the main method for evaluation of different sport competitors physical features. Various sets described by number of authors differ mainly in segmental position of the subject during measurement. This positions may significantly influence the results obtained.

In this study a general description of set for measurement in static of muscle torque in ankle joint is performed.

Construction

The set consists of stabilizing and measuring systems. The diagram of the set is shown in Fig.1. The stabilizing system constitutes of seat (1), shafts (2) and a thigh clasp (6). The clasp is pressed against the thigh and fixed by two screws (7). The measuring system is built of the pedals (3) equipped with a force sensor (4), an amplifier (8) and a meter (9). The force sensor is a strain gauge ring mounted at the fixed distance from the axis and works as a torque meter. Pedals as well as the stabilizing clasp can be gradually moved in vertical plane.

Measurement technique

The set enables the measurement of a torque developed by the flexors of a foot in static conditions. The subject is

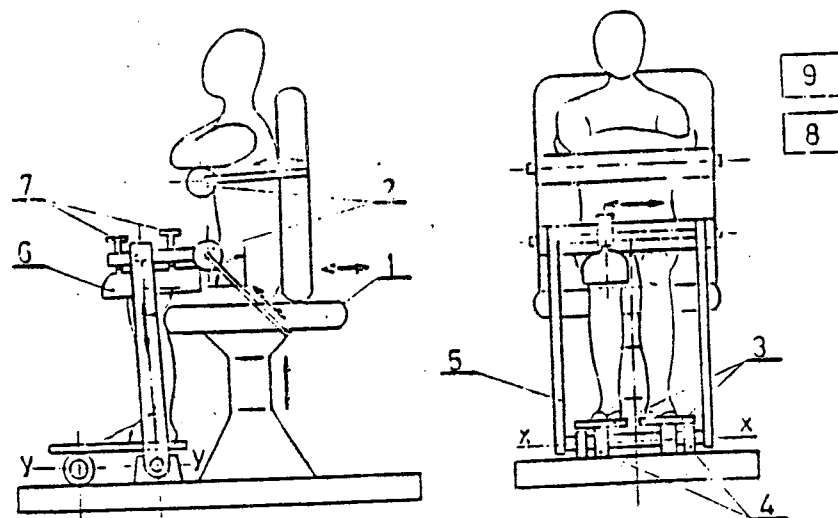


Figure 1. The diagram of the measuring set.

sitting on the seat with hip, knee and ankle joint angles equal 90° . The rotation axis of the measured joint is placed exactly above the x-x axis. The position of a subject can be assured by the adjustment of the seat, the back-rest, the clasp and the pedals. The stabilization is performed by pressing down and locking of the stabilizing shafts and of the thigh clasp. The value of a measured torque is digitally displayed on the meter. Shifting of the clasp enables the measurement of both legs.

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DEVELOPMENT OF FRICTION ANTICIPATION AND ADAPTATION IN THE HUMAN PRECISION GRIP

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INTRODUCTION

It has been well recognized that the fine motor-control system, which includes mechanisms responsible for the precision handling actions of the digits, develops ontogenetically during the early developmental stages of infancy (Lawrence and Kuypers, 1968). A series of recent studies dealing with healthy adults (Johansson and Westling, 1984, 1987; Westling and Johansson, 1984) has demonstrated that during manipulatory tasks involving the grasping of small objects by the tips of the index finger and thumb, the grip force (GF) was accurately controlled and adapted to the frictional condition of the object, providing a fairly small safety margin to prevent slips. Their studies have also shown that there are fundamentally three different mechanisms involved in the adaptive control of the forces to a given frictional condition: (1) preparatory control depending on the frictional demand during the previous lifts, (2) recognition and adaptation during the early phase of lifting, and (3) microslips and upgrading of the forces during the lifting and holding phase. The aim of this study was to investigate the development of anticipatory and adaptive control mechanisms in children to varied frictional conditions in the surface structure of an object during lifting and holding tasks using the precision grip.

METHODS

A total of 77 children with no known neurological impediment (13 months-14 years) and 15 healthy adults participated in the study. The experimental set-up consisted of an instrumented lifting object, a photoelectric position detecting unit, and an IBM personal computer (PC-AT) data acquisition and analysis system. The object was equipped with transducers for measuring the isometric GFs and the load force (LF: vertical lift force), as well as LEDs for providing a light source to the position detecting unit. An accelerometer was attached to record vibrations caused by small slips between the skin and the object. Two different experiments were carried out at separate experimental sessions. The first experiment was devised to examine the development of adaptation to friction during the static phase. The object was lifted and held in space for approximately 10 seconds, and it was then dropped by slowly separating the fingers (see Figure 1 phase B and C). The second experiment aimed to test the development of anticipation about friction, as well as of adaptive responses during the dynamic phase (Figure 1 phase A). The subjects performed a sequence of reach, grip, and lift actions in a semi-dark room while the surface structure was pseudorandomly varied between lifts. The surface materials used were a relatively rough sandpaper (SP) to provide a non-slippery condition and a finely textured silk (SLK) for a fairly slippery condition. The weight of the object was set at 210 g. All recorded force, vibrational (acceleration), and vertical position signals were fed into the computer via a 12 bits A/D converter sampling at 400 Hz for each channel.

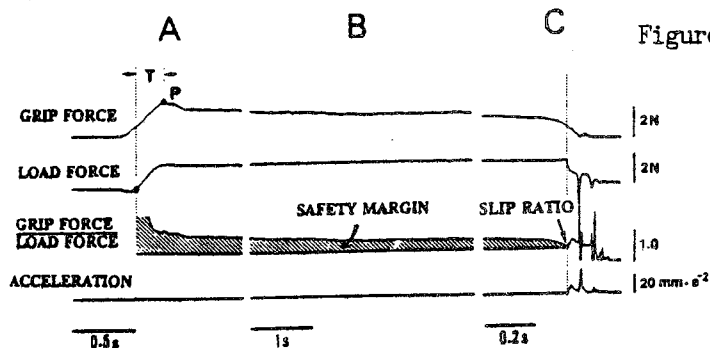


Figure 1. Typical GF, LF, GF/LF patterns and some of the selected parameters

A: Lifting phase
B: Holding phase
C: Drop phase

RESULTS AND DISCUSSION

Adaptation of the GFs to a given frictional condition between the skin and the object is of critical importance, particularly when smooth and gentle manipulative actions of the object are required. The magnitude of the forces must be large enough to prevent accidental slippings, but it should not be excessive to hamper the intended manipulations or to damage the object. Loss of adaptation occurs after cutaneous anaesthesia, indicating that tactile afferents are highly responsible in this operation (Johansson and Westling, 1984). The results of our study confirmed that the average GF/LF ratio for the SLK condition was significantly larger than the SP condition in all age groups (Table 1). Adaptative capacity after sensing the frictional difference during the holding phase seemed to be thus already intact even in 1 year-old infants. There was a significantly larger safety margin in children younger than 5 years of age compared to adults (Table 1). The value for the 1 year-old group was the largest among them, and it was indeed nearly three times the adult value, clearly reflecting the premature state of their sensory-motor connection.

To examine the capacity of anticipation of friction, data from the lifting trials in which the surface structure was switched were compared to those from the trials with unswitched conditions. The subjects were neither verbally informed about the new surface, nor could they visually distinguish the material. Responses associated with changes from more to less slippery material were examined by means of the peak GF (P value in Table 2). A comparison between the unchanged friction (SP followed by SP:SP-SP) and changed friction (SLK-SP) revealed that an overshoot of GFs resulting from anticipatory actions was significant for all age groups. The anticipation of the more slippery condition was accessed by the duration of the loading phase (T value in Table 2). Mean group data indicated that the duration was longer in the SP to SLK (SP-SLK) condition than in the SLK to SLK (SLK-SLK) condition for each age group. Statistical comparisons, however, revealed that the difference between the conditions in each of the younger age groups (1-3 yrs.) was insignificant.

Findings of this study suggest that preparatory control capacity for friction and its subsequent adaptation, develop during the early infant period, which seems to start even prior to the second year of life. To have a relatively efficient use of the capacity, on the otherhand, appears to require a further 5 to 5 to 6 years.

TABLE 1. RESULTS OF HOLDING PHASE ANALYSIS

AGE (Yr.)	N	AVERAGE GF/LF RATIO		AVERAGE SAFETY MARGIN	
		SP	SLK	SP	SLK
1	5	2.4±0.4	* 4.1±1.6	1.5±0.40	1.8±0.80
2	7	1.9±0.4	* 3.7±0.7	1.3±0.30	1.3±0.40
3	6	1.6±0.3	* 3.1±0.6	1.0±0.40	1.2±0.60
4	4	1.5±0.4	* 2.5±0.7	0.8±0.30	0.8±0.20
5	5	1.1±0.2	* 2.9±0.8	0.5±0.2	0.8±0.30
6-9	5	1.1±0.2	* 2.3±1.4	0.6±0.1	0.7±0.3
11-14	5	1.0±0.1	* 1.5±0.6	0.5±0.1	0.5±0.1
ADULT	7	0.9±0.3	* 2.1±0.4	0.5±0.1	0.5±0.1

* P<0.05: SP vs. SLK

0 P<0.05: EACH CHILD GROUP VS. THE ADULT GROUP

TABLE 2. RESULTS OF LIFTING PHASE ANALYSIS

AGE (Yr.)	N	P(N)		T(ms)	
		SP-SP	SLK-SP	SLK-SLK	SP-SLK
1	5	5.11±1.76	* 5.47±2.22	407±85	459±77
2	5	4.45±1.46	* 5.80±1.98	380±80	452±100
3	5	3.16±1.50	* 3.96±1.65	325±73	386±72
4	5	2.75±1.02	* 3.39±1.64	332±74	* 432±102
5	5	2.82±1.02	* 3.65±1.49	302±54	* 434±102
6-9	8	2.32±1.12	* 2.49±1.49	363±107	* 457±121
11-14	7	1.96±0.68	* 2.49±0.63	362±148	* 430±173
ADULT	8	2.09±0.90	* 3.10±1.64	309±74	* 372±70

SP:SANDPAPER, SLK:SLK

* P<0.05: SP-SP vs. SLK-SP, AND SLK-SLK vs. SP-SLK

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DEVELOPMENT OF ISOMETRIC FORCE ACTION PROGRAMMED FOR WEIGHT
DURING A LIFTING TASK WITH PRECISION GRIP

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INTRODUCTION

When lifting an object with known weight a person uses a centrally stored information to generate and adjust the grip forces (GFs) and load force (LF: vertical lift force) to a magnitude just exceeding the weight of the object. The capacity to use the central program appropriately can be tested by changing the weight of object without the knowledge of the subject (Johansson & Westling, 1988). Earlier Brooks (1979) differentiated the nature of motor output patterns depending on the extent of involvement by a centrally generated muscle command. He found that during a fast target tracking task the first derivative (velocity) of the arm motion formed a continuous or bell shaped curve compared to an irregular curve during a slow discontinuous movement. According to Brooks the continuous movement was performed in one sequence and was programmed in advance, in contrast to the polysequential advances during discontinuous movement relying mainly on ongoing feedback. In the study by Johansson and Westling (1988), it was demonstrated that in adults the first derivative of GF (DGF: dGF/dt) followed a pattern similar to the derivative curves of Brooks. The present study concerned with the development of motor programming ability with respect to the weight of lifted object in children of various ages.

METHODS

One hundred children (9 months - 15 years) with no known neurological impediment and 15 healthy adults served as subjects for the study. The experimental set-up consisted of an instrumented lifting object, a photoelectric position detecting unit, and an IBM personal computer (PC-AT) data acquisition and analysis system. The GF applied by each of the thumb and index finger and LF were measured independently by transducers installed in the object, and the vertical displacement of the object by LEDs and photoelectric position sensor. The moment when the object lifted from the table was then estimated from the position data. All force and displacement signals were fed into the IBM computer via a 12 bits A/D converter sampling at 400 Hz for each channel. The grip-force-rate (DGF) was calculated by digitally differentiating the stored GF data, and it was then low-pass filtered at 50 Hz to reduce the high frequency component (Johansson and Westling, 1988). The subjects were instructed to attempt a series of lifting trials from a table while the weight of the object was pseudorandomly switched out of the sight of the subjects. Children aged above 2 years lifted an object weighing either 200g or 800g, while those at 1 year old lifted either 200g or 400g. Adults were tested with all three weights.

RESULTS AND DISCUSSION

Confirming the previous findings by Johansson and Westling (1988), in a lifting series with constant weight by adults as well as older children (above 6 yrs.) in this study, the DGF, when plotted against the LF, followed mainly a bell-shaped path and was roughly proportional to the terminal force (see average trial trajectories for 200g or 400g in Figure 1 left graphs). Results in a lifting series with unexpected weight changes between lifts further elucidated that the fact that the DGF profile was programmed on the basis of the previous weight. Consequently, in the lifts with 400g, while the previous lift was with 200g, the initial bell shaped DGF profiles aimed at a load force target corresponding to approximately 2N force followed by numerous small force output

(Figure 1 upper-left graph). During such erroneously programmed trials the moment of lift-up was delayed and the DGF changed in a discontinuous fashion. When the current lift was preceded by a heavier weight, there was a pronounced overshoot of the GF which resulted in a larger DGF peak (Figure 1 lower-left graph). Observation of the DGF profiles in one of the young infants (see an example in Figure 1 right graphs), suggested that the programming capacity in this child was insufficient. To evaluate the degree of adequacy in the motor programming capacity in children, two parameters specifically pertaining to the adaptive responses to unexpected weight changes were selected. These were the peak DGF and the duration of loading phase (the period between the initiation of positive LF and the initiation of upward displacement). For each of these parameters, the ratio between the constant weight condition and weight change condition was computed for each child for each age-group (Figure 2). The results suggested that on average 1 year-old children were capable of motor planning actions, though accuracy and consistency of the observed force outcome were considerably poor compared with the adults. Rapid progress took place between 1 and 4 years, developed gradually from 5 and 6 years, and an apparent refinement of the capacity to a level similar to that of an adult appeared by around the age of 6 to 7 years.

Figure 1. An example of DGF profile in relation to LF

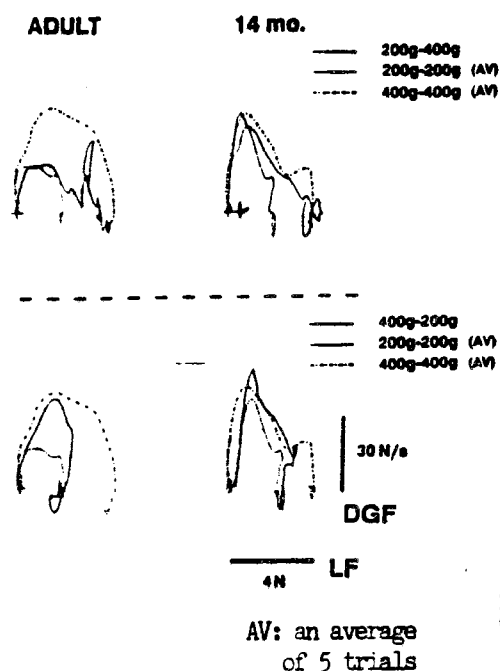
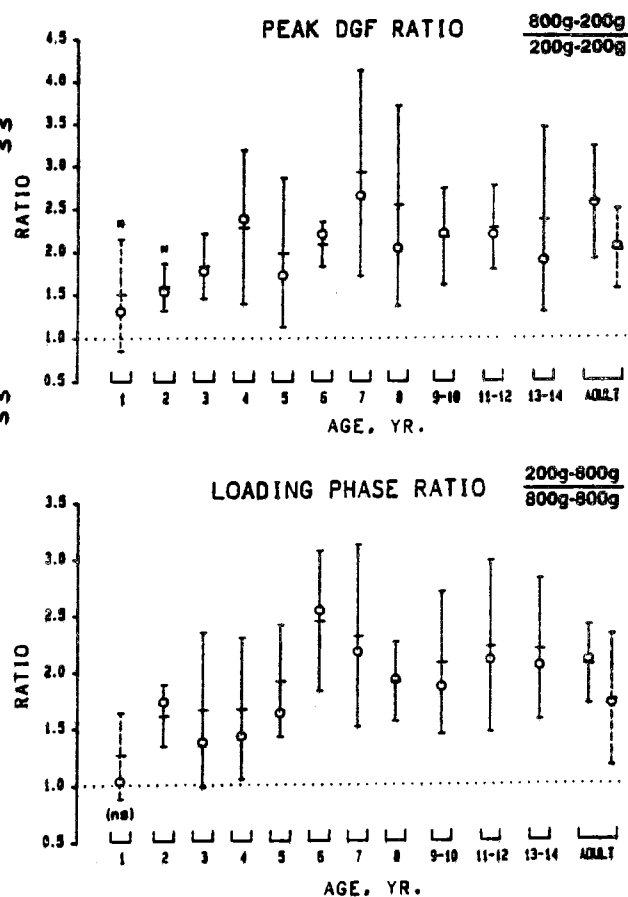


Figure 2. Development of programming capacity (—: mean, o: median, I: $\pm$ SD)



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The 1 yr. old group was test with 200g-400g series.

* $p < 0.05$:against adult mean

(NS) $p < 0.05$:against ratio 1

ANALYSIS OF THE VIBRATIONS OF THE VERTEBRAL COLUMN IN THE MAN-VIBRATING OBJECT SYSTEM

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The problem of the impact of vibrations on man was presented by means of a discrete model with many degrees of freedom. It was assumed that vibrations are transferred to the human organism through the environment of the vibrating object, e.g. a vehicle in the direction of the y, x, z axis. The vertebral column and other anatomical parts of the human body were presented as a system of solid bodies with masses m_i , moments of inertia I_{x_i}, I_{y_i} , joined with one another in a viscous-elastic way and arranged in a proper way in relation to one another as in Fig.1.

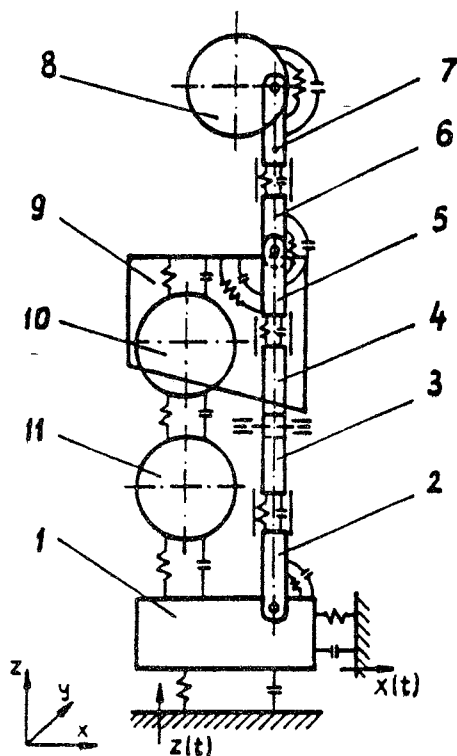


Fig.1-Biomechanical Model of Man in a sitting position for the investigation of the effect of vibration; 1-pelvis, lower part of the trunk; 2-lumbar vertebral column L5-L4; 3-lumbar vertebral column L3-L1; 4-thoracic vertebral column T12-T7 and T6-T1; 5-cervical vertebral column C7-C4; 6-cervical vertebral column C3-C1; 7-head; 8-thorax and shoulder girdle; 9-thoracic internal organs; 10-abdominal internal organs.

$\dashv \vdash$ - c_{ij}, k_{ij} - damping and rigidity coefficients between segments i and j ,

$\dashv \rangle$ - c_{tij}, k_{tij} - torsional damping and rigidity coefficient in relation to the y axis, x, y, z - coordinates describing the vibrations of the system.

The mathematical model was presented in the following notation

$$A\ddot{\bar{q}}(t) + B\dot{\bar{q}}(t) + C\bar{q}(t) = E\ddot{h}(t) + F\dot{h}(t) \quad (1)$$

where: A, B, C - matrices of inertia, damping and elasticity,
 E, F - matrices of input,
 $\bar{q}(t)$ - generalized coordinates vector,
 $h(t)$ - input vector.

The parameters of the system for numerical calculation were adopted on the basis of bibliographical studies: Nader (1985) Nader, Lachowicz (1987), Nader (1988). The effect of changes of the parameters on the model solutions was also investigated by means of the sensitiveness procedure. Examples of the results of the investigation of the vibrations of the vertebral column obtained through the analysis of the mathematical model (1) are presented in Fig. 2.

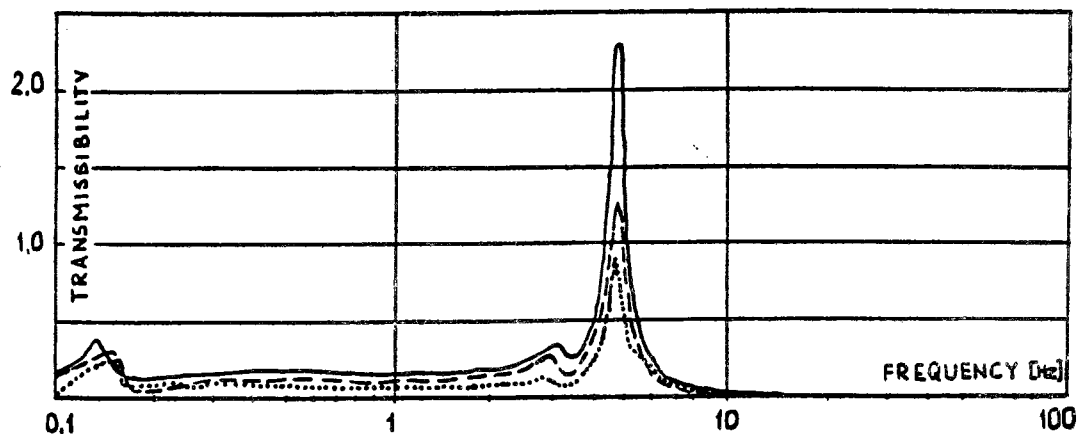


Fig. 2. Transmissibility of vertical vibration of the systems
 — seat/lumbar vertebral column L3-L1
 ---- seat/thoracic vertebral column T6-T1
 seat/cervical vertebral column C7-C4

Detailed results of the investigations are included in the author's works carried out within the framework of the Central Scientific-Research Programme 11.1 entitled "Protection of Man in Work Environment".

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VALIDATION OF A THREE-DIMENSIONAL KINEMATIC MODEL OF THE LOWER EXTREMITY OF A CYCLIST

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The objective study of cycling mechanics is a prerequisite to optimizing cycling efficiency and devising strategies to reduce the likelihood of injuries caused by incompatibilities between the cyclist and the bicycle. Lower extremity kinematics have been recorded with cinematography and videography (1,2). Force transducers have been used to measure the forces exerted on the pedals by the feet (3,4), and the activity of muscles providing these forces have been monitored using electromyography (3,4,5). For the sake of simplicity it has usually been assumed that the motions of the segments of the lower extremity during cycling are co-planar. However, a review of the nature of the common overuse injuries suffered by cyclists leads to the speculation that potentially-injurious stresses on anatomical structures may result from complex three-dimensional rotations (2).

On the basis of the anatomical structure and presumed kinesiological function of the segments of the lower extremity a three-dimensional kinematic model of cycling was developed. The model consisted of three rigid links attached by joints which permitted rotations about 10 independent axes. Motion of the distal end of the model was constrained to a circular locus. The matrix transformation method was used to obtain the inverse kinematic equations involved. Due to the fact that the model represents an indeterminate kinematic system, an iterative optimization algorithm was developed to compute joint variables and generate numerical solutions.

The validity of the model was examined using 12 highly skilled, injury-free, male subjects who were members of the 1988 United States National Cycling Team. Each subject cycled on a bicycle which was mounted on a stationary roller system. Pedalling load was generated by the aerodynamic drag of a multi-bladed fan that was rotated by the motion of the rear wheel of the bicycle. Kinematic data were collected using a Vicon Motion Analysis System and a sampling rate of 50 Hz. Four cameras were placed so that at least any two of the cameras would record the spatial coordinates of 11 infrared reflective markers attached to the crank mechanism of the bicycle and the right lower extremity of the subject. The locations of markers on the thigh, the shank and the foot were selected so as to minimize the effect of independent motion of soft tissues. The distances between markers were carefully measured and the static locations of the markers during normal stance were recorded just prior to the collection of cycling data. The locations of the cameras and their optical axes were established from the recorded images of a series of markers attached to a calibration device that was temporarily positioned in the volume of space that was subsequently used for data collection.

Each subject was instructed to cycle at normal training cadence and pedalling load. When the subject indicated that he had achieved a steady cadence, data were collected for a total of 10 complete revolutions of the cranks. The positions of all of the markers as recorded by one of the

cameras were then reconstructed on a high resolution monitor. Each of the markers was identified and the process repeated for the data obtained from the other three cameras. Using the software and procedures recommended by the manufacturers of the system it was verified that the data were free from obvious artifacts and that the positions of each marker had been recorded by at least two of the cameras during data collection. Subsequently, all position data were established in a fixed system of orthogonal coordinates.

Data obtained from the subjects showed that all of the angle-time histories examined were periodic functions, but there were consistent differences between subjects. However, subject data were consistent with those obtained from model simulations. Discrepancies between subject and model joint angle-time histories varied from joint to joint, and so it was concluded that the accuracy of prediction will be improved as additional constraints are placed upon the kinematics of the model.

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EFFECTS OF THE HEIGHT AND LANDING TECHNIQUE ON THE TAKE-OFF MOTION IN DEPTH JUMPING

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[PURPOSE]

In recent years, investigations on the depth jumping have been getting popular remarkably. Most of studies have focused on the factors of gaining height on the depth jumping. Few studies have been analyzed the take-off motion from technical view point. It is more important to study the effective technique, especially for top-class jumpers, to improve their specific strength[2]. There is an important and most widely supported technical opinion for an effective jump take-off which the take-off foot or feet should be brought to the ground in a pawing movement. This action, also known as an "active landing," reduces the forward horizontal momentum at the touch-down and providing the better supportive body segmental alignment[1, 3]. The effects of the active landing, however, have not been investigated in the details. The purpose of this study was to investigate the characteristics of the depth jumping as a training exercise to develop a reactive-explosive strength and to evaluate how the active landing works as an effective technique to compare with the ordinal take-off motion, shortening and lengthening of the limb(s), as changing the drop height

[METHODS]

Twenty-one male high performance jumpers performed the depth jumping at full effort from five different heights - 0.3, 0.5, 0.8, 1.0 and 1.2m, by using two different take-off motions (normal landing and active landing) as shown in Figure 1. A high speed video motion analysis at 200fps, and a force platform system were employed to analyze the depth jumpings. The vertical ground reaction forces, contact time, lower limb joint angles (hip, knee, and ankle), the vertical displacement, and the vertical velocity of the C.G were calculated. These kinematic and kinetic data were averaged, and then normalized by the longest contact time and their body weight every 200fps, to compare with the two different landing techniques.

[RESULTS AND DISCUSSION]

The vertical ground forces during the take-off had the characteristic shapes as shown in Figure 2. The mean ground reaction force was increased during the take-off about 5 - 9 times/bw in the normal landing and 6 - 10 times/bw in the active landing as increasing the dropping height. The greater ground reaction forces, in which the dropping height 1.0m, was developed remarkably high during the former half of the take-off phase. That meant the difficulty to perform the intended take-off motion at that height. After the peak force, the force curve was reduced remarkably as increasing the dropping height at the mid-point of the support phase. From those results, it might be suggested that the dropping height of 0.8m seemed to be a limiting height from the technical view point to perform the rational take-off motion as generating the maximum force at the mid-point of the support phase.

The active landing showed less flexion of the lower limb joints than the normal one during the take-off. The maximum knee flexion at the mid-point of the support phase varied from $127 - 133^\circ$ as increasing the dropping height. That was more similar to the knee joint angle during the actual take-off in the jump events of the Track & Field than the normal landing. However, at the dropping height 1.0m, the knee joint flexion was decreased remarkably than at the other dropping heights by using the active landing. The significant difference was found in the change of the dropping height between 0.8m and 1.0m ($p < 0.01$). The active landing showed the shorter contact time than the normal one, and generated the greater force at the mid-point of the support phase than the normal one as shown in Figure 2. These results suggested that the active landing should be considered as an effective technique for the take-off motion in the sports which demanded to develop and improve the reactive-explosive strength within the limited short contact time.

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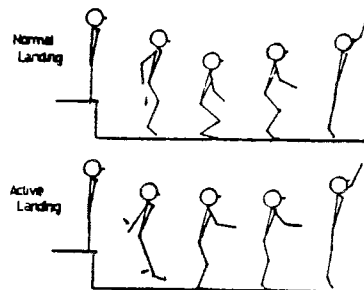


Figure-1
Two different landing techniques.

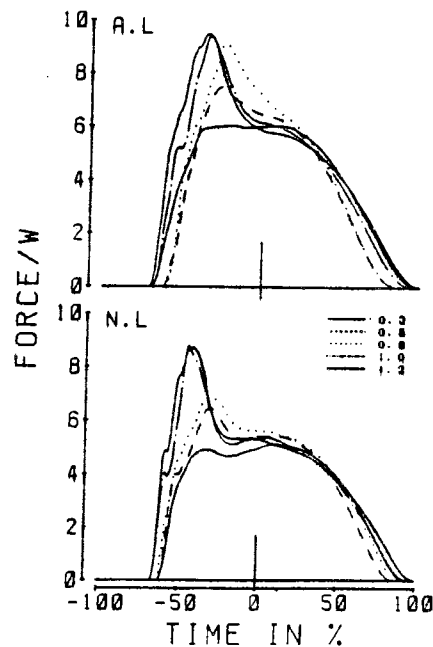


Figure-2
Change in vertical ground reaction forces of two different landing techniques as increasing height (N.L.-normal landing, A.L.-active landing).

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(ABSTRACT)

Biomechanical Analysis of SUKUINAGE(Scoop throw) in Judo

The purpose of this study was to analyze the throwing motion of SUKUINAGE (Scoop throw) in Judo.

Subjects were two top class judoists in Korea. SUKUINAGE motion was filmed on 16mm high speed cameras (100 fps, at front & side view).

After that SUKUINAGE was analyzed with respect to three phases : KUZUSHI (breaking the opponent's balance), TSUKURI (preparation before throwing the opponent), and KAKE (throw the opponent), from the biomechanical aspects.

The results were as follows;

In case of subject A :

- (1) The time elapsed from the beginning to the 1st stage of KAKE phase was 1.0sec., to the 2nd stage of KAKE phase was 1.04sec., and the section time was 0.04sec. from the 1st stage of KAKE phase to the 2nd stage of KAKE phase.
- (2) The knee angle of attacking left-leg showed 103° in the 1st stage of KAKE phase and flexed to 100° in the 2nd stage of KAKE phase.
- (3) The C.O.G. angle of UKE and Sub.A was 68° in the 1st stage of KAKE phase.
- (4) The C.O.G. of TORI (Sub. A: attacker) was 67.5cm forward from the beginning to the phase of KAKE and the C.O.G. of UKE was 3.4cm forward. Therefore, KUZUSHI of TORI was insufficient.

- (5) The time elapsed from the 2nd stage of KAKE phase to the touch of the hip on the mat was 0.92sec.

In case of Subject B :

- (1) In the position changing from the phase of KUZUSHI into the phase of KAKE, distance of KUZUSHI range was 7.9Cm in the 1st stage of KAKE phase and 13.5Cm in the 2nd stage of KAKE phase which was KUZUSHI of TORI Sub. B was insufficient.
- (2) In the position of pulling left shoulder of UKE, rotation(angular) movement was insufficient with the phase of KAKE following.
- (3) The time elapsed from the beginning to the 1st stage of KAKE phase was 0.98sec. to the 2nd stage of KAKE phase was 1.09sec., and the section time was 0.11sec. from the 1st of KAKE phase to the 2nd stage of KAKE phase.
- The section time was 0.11sec., longer than that of Sub.A by 0.07sec.
- (4) The moving angle of C.O.G. between UKE and TORI in the 1st phase of KAKE, in the 2nd stage of KAKE phase was 9° , 10° , respectively which was mechanically inefficient.
- (5) The time elapsed from the 2nd stage of KAKE phase to the touch of the hip on the mat was 0.84sec., which was shorter than Sub.A(0.92sec.) by 0.08sec. that
- (6) The knee angle in the positions of lifting UKE was 108° , and the knee angle of attacking left leg was 113° in the 1st stage of KAKE phase extending to 147° in the 2nd stage of KAKE phase.

Finally it was because of individual characteristics in the variation value of kinematic. SUKUI-NAGE was found to be one of the most important techniques to score. But on the whole, motion of KUZUSHI, the 1st phase of technique, was very insufficient.

THE HUMAN FOOT: A COMPOSITE SPRING WITH STRAIN DEPENDENT
EXPONENTIALLY INCREASING STIFFNESS

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Tested as isolated structures bones and ligaments show a load-deformation curve with a well defined linear part in the elastic region the slope of which indicates the stiffness of the structure. At the yield point the relation is transformed into a non-elastic region with gradually decreasing stiffness until the breaking point is reached. The result is that the isolated structure acts as a simple spring which yields continuously under increasing load (1,2).

Composite structures made from bones and ligaments as the vertebral column and the arch of the foot behave differently. They preserve their integrity by setting a limit to their deformation thereby protecting their joints, vessels and nerves. The present study investigates how the foot sets this limit as a composite structure.

Material and methods

Four unfixed foot specimens from persons aged 57 to 72 were tested. The specimens which consisted of bones, the long and short plantar ligaments, the calcaneonavicular ligament, and the plantar aponeurosis were loaded vertically through the tibia in a materials testing machine with controlled stroke and load. The strokes varied stepwise from 2 to 9 mm and the loads from 25 to 1200 N equaling two body weights. Strain rate was approximately 30 mm/s, and each load was maintained for 1 s and followed by a release period of 5 s. The specimens were tested with the supporting ligaments cut in sequences with the superficial ones first. Each step in the sequence was concluded by a stress relaxation test lasting 60 s. The signals sampled at 10 Hz were A/D converted and input to a personal computer.

Results

For loads up to two body weights the arch of the foot showed a steadily increasing stiffness or tangent modulus with no linear part of the load-deformation curve appearing. The curve was shown to fit an exponential function:

$$\text{Force} = \exp(a * \text{deformation} + b)$$

where "a" was found to decrease from 0.67 to 0.49 when the deep plantar ligaments were cut. Severance of the plantar aponeurosis did not affect factor "a" when the toes were in neutral position, but excluded the foot from a further increase in stiffness by dorsiflexion of the toes which is known to tense the aponeurosis. Factor "b" varied randomly between 4.2 to 4.6. All fits showed correlation coefficients above 0.97. The stiffness of the spring expressed as the tangent modulus at 3 mm deformation varied individually from 480 N for a small female foot to 915 N for a large male foot. The stiffness decreased 30-40% when the aponeurosis was cut and a further 15-20% when the deep ligaments were severed.

For stress relaxation a linear regression line with a slope of - 7-8%/s was fitted between force and the natural logarithm of time. The force decay was unrelated to the number of fibres and ligaments present.

Discussion

Adding fibres or spring elements in parallel to a carrying structure of a given dimension as a response to increasing loads will increase its stiffness and set a limit to its deformation. A composite structure as the arch of the foot can thereby act as a soft spring which readily yields under small loads while it under heavier loads will act as a stiffer spring which hardly can bottom out.

The foot is divided by a cleavage plane into a tibial and a fibular set or column of bones. The cleavage is spanned by the investigated ligaments which all run distally from the proximal part of the fibular set to the tibial one. When a load is transferred from the body to the foot it happens by way of the talus to the tibial set which thereby slides forward compared to the fibular one until it is stopped by the ligaments.

Instead of the concept of the arch as a simple spring a two-column sliding mechanism is suggested with a sequential engagement of the plantar ligaments and thereby with a strain dependent exponential increase of its stiffness.

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A TELESCOPING EFFECT OF THE HUMAN HAND AND FOREARM DURING HIGH ENERGY IMPACTS

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Introduction.

Impacts against or created by the human body or a part of it are very complex in nature because none of the structures act as solid masses during an impact, and the overall dynamics are an interaction between anatomical structures, tissues and body fluids, the energy flow between the segments and an input of mechanical energy from the muscles.

Materials and methods.

A karate punch very similar to the boxer's straight right was studied in 10 well trained karate students. A punching dynamometer was constructed (accelerometry) and calibrated with a series of controlled impacts (1). Punches were recorded by stroboscopic photography (100 Hz), high speed filming (500 Hz) and a triaxial force plate.

Results.

Peak impact force: 3334 N (2345 - 4866 N), maximal speed of the hand before impact: 9.5 m/sec (8.2 - 10.7 m/sec), effective mass from calibrated punching dynamometer: 1.4 kg (1.0 - 2.2 kg), effective mass from stroboscopic recordings: 4.0 kg (2.5 - 4.5 kg). The collisions showed inelastic characteristics (coefficient of restitution, $e = 0.3$ (0.0 - 0.4)).

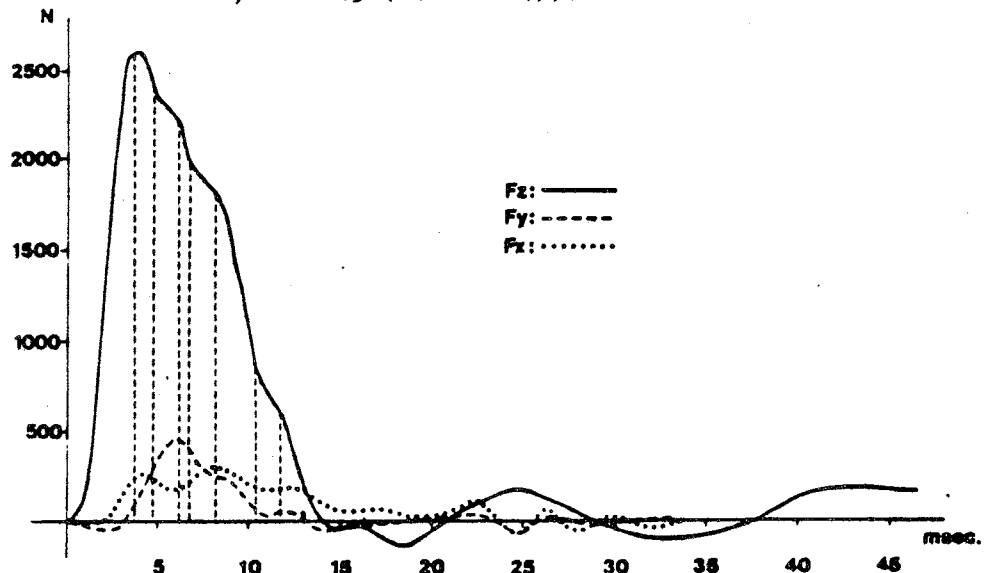


Figure 1.

Averaged force patterns during 8 submaximal vertical impacts of a fist against a triaxial force plate (horizontal) from one subject. The hand hits the platform with the knuckles of the second and the third metacarpal bone of the right hand.

Maximal speed of the hand before impact: 6.3 m/sec (5.1 - 6.9 m/sec).

Fx = to the sides, positive = ulnar side. Peak Fx = 329 N (253 - 385 N)

Fy = forth/back, positive = dorsal side of the hand. Peak Fy = 480 (369 - 573 N)

Fz = vertical, positive = down. Peak Fz = 2597 N (2370 - 2847 N)

The shape of the force curves in the punching direction (Fz, fig 1.) during the impacts of the unprotected hand was characterized by an asymmetrical shape (fast ascending part, slow descending part), by tripled impact time (average 15 msec) and by reduced peak force compared with impact curves from impacts between solid and homogenous masses. The descending part of the force curve showed repeatable waveform patterns with 1 - 3 deflections. The typical time history of Fz: peak force after 4 ms, 1. deflection after 5 ms, 2. deflection after 6 ms and 3. deflection after 10 ms. If the subjects were instructed to tense their fists hard before they delivered their punches, the first deflection on the Fz curve disappeared.

The high speed film recordings revealed a telescoping effect of the hand during the impacts, consisting of a compression of the tissues and changes in the position of the bones in the hand, and spring like vibrations between the structures.

Discussion.

In karate and boxing it is interesting to define the perfect knock-out punch. From the second law of Newton we have : $F(\text{impact}) \times t = m \times v$, and it is seen that a big effective mass with a high speed just before the impact and combined with a short impact time gives a high impact force. The effective mass behind a punch is very difficult to determine, because of the complex structure of the hand and the arm. Determination of the effective mass during punching from film recordings, calculated from the law of conservation of momentum (2), must be wrong because the muscles in the arm continuously are changing the energy level in the system and the residual velocity of the hand is affected in a drastic way. The experiment with the calibrated punch dynamometer showed that the peak impact force and the time to peak force in a punch is determined by the composition of the contact surfaces, and the speed and mass of the hand + forearm. The karate expert develops a fluid filled tissue cushion over the knuckles of the second and third finger, this cushion acts like a damper in the same way as the heel pad (3) and increases the stroke length in the impact between the knuckles and the target and thereby increases the impact time and reduces the peak force according to the equation above. The hand and forearm act as two (or more) masses connected with springs and dampers during an impact (4). The characteristics of the springs and dampers depends on the tension in the muscles, the filling of fluid compartments and the compliance of the connective tissues in the hand. The overall stroke length of the hand + forearm-telescope is determined by the thickness of the tissues over the knuckles, a dorsal flexion of the wrist, a rotation of the carpal bones starting with a rotation of the scaphoid bone to the ulnar side of the hand and the compliance of the ligamentous structures in the hand and wrist. The elastic part of the telescope is created by a loading of mechanical energy in the ligaments during the change in position of the bones and in the fascia's and the subcutaneous connective tissue during the shift of the fluids when the tissues are compressed. The first wave on the descending part of the F_z curve (fig 1.) is determined by the collision of the thenar + hypothenar with the flexed fingers in the palm corresponding to the peak force in the dorsal direction (F_y). The energy that creates the following waves is suggested to be the energy return from the loaded connective tissues. It was not possible to detect differences between the subjects in the measured variables.

Conclusion.

The maximal speed of the hand before the impact in a karate punch was 11 m/sec and the fist tolerates impact forces up to 5000 N without damage. To withstand this stress the hand act like a telescope, wich balances the energy absorption and energy return. The energy absorbing ability, the stroke length of the telescope is determined by tissue compression, changes in position of the bones and the compliance of the connective tissues. The energy return from the telescope is determined by the amount of mechanical energy loaded and returned from the ligaments and subcutaneous connective tissues. Determination of the effective mass active during a karate punch is complicated by the fact that the mass changes during the impact and the muscle energy de- or accelerating action.

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STUDY OF INTERACTION CURVILINEARLY GLIDING

SKATE BLADE WITH ICE

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Changing his posture (mutual location of the body segments) a skater controls a displacement of his general centre of mass (GCM) relatively to the blade of the skate. The blade has an oval cross-section with a curvature gradually varying in longitudinal direction and a constant curvature radius of the groove in transverse direction. The forces and moments acting on the blade determine the trajectory of a skater's gliding, and depend upon the contact zone of the blade with the ice, the tilt angle of the blade relative to the ice plane and velocity of gliding. Rational utilization of different zones of the blade will enable skaters to move along desirable trajectories with a minimal ice resistance preserving the balance and providing stability necessary during the performance of variety of elements.

The importance of the problem induced us to investigate at theoretical and practical levels the interaction of curvilinearly gliding blade with ice.

Measurements of trajectories of an actual figure skate and blade specimens with fixed oval curvature radius were performed by means of specially elaborated device. The device construction provided for changing the blade (specimen) tilt angle and longitudinal dislocating its contact zone on ice.

Under experimental procedure the skate (specimen) was propelled at a slow constant velocity, its trajectories (being circular with altered radii) changed according exper-

imental conditions, that is tilt angle variation, oval curvature, weight, skate motion direction, and availability of a groove.

The experimental material obtained and data processed made it possible to describe theoretically the interaction of skate moving along a curvilinear trajectory with ice.

The suggested model can be used for solving two kinds of the problem: the direct task is to determine the trajectory of gliding (the radius R of the arc) under the given curvature of the blade oval (r), its tilt angle relative to the plane of ice (α), velocity of gliding (V) and curvature radius of the groove (r_g). In the reverse task one needs to define the curvature radius of the blade oval (r) at a given trajectory of gliding (R), velocity (V), tilt angle (α) and curvature radius of the groove (r_g).

Skater's uniform gliding conditions were described by the system of equations defining the balance of forces and moments.

The obtained results showed the blade geometry is of crucial importance for the movement parameters and to be chosen individually; anthropometric and mass-inertial characteristics (MIC) of the body being taken into consideration.

ON THE ENERGY TRANSFER BY A MUSCLE DURING WAVE
MOTION OF HUMAN BODY'S SEGMENTS

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In previous study (Popov, 1987, V Symposium of ISBS) it was shown that the designing multisegment system of human body's movement as a transverse wave is accompanied by the energy transfer along body's segments.

It was logically to suggest that such energy transfer is accomplished through muscles which bring body's segments in action.

In multi-segment model of human body there were chosen two adjacent segments (the i -th and the $i+1$ -th segments). These segments were connected by a muscle which was modeled as a non-linear element (NE) including a contractile component (CC), a consistent elastic component (CEC), a parallel elastic component, and damping component (DC).

Wave motion $a \cdot \cos(\omega t - kx)$ propagates from the previous segments to the chosen segment (the i -th and the $i+1$ -th segments) and then to the following segments.

Active human actions were expressed as a Gauss "bell-shaped" pulse. It allowed us to come to muscle's model in the form of non-linear spring, consisting from CEC, PEC and DE, as in the study of Stilas R., Alexander R. (1972). The energy per cycle coming in the NE from the previous segments and then transferred to the following segments was calculated.

It was shown that the increasing energy transfer along the NE requires the generating impulse control moments, which bring the chosen segments in action.

These moments must be out of phase being equal to wave propagation time from the i -th segment to the $i+1$ -th segment. The increasing energy transfer occurs also during the NE ends's movement which are connected with the i -th and the $i+1$ -th segments being out of phase by 90° .

This condition determines the range $[0, 2\pi]$ of the energy transfer and energy absorption by the muscle.

The results allows us to explain the physical nature of the conclusions made by D. Robertson, D. Winter (1980) and by V.M. Zatsiorsky (1986) with respect to the mechanical energy exchange between body's segments.

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MATHEMATICAL MODELLING AND EXPERIMENTAL DETERMINATION
OF WAVE PROCESSES IN JAVELIN THROWING TECHNIQUES

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It was shown recently (Popov G.I., 1987 - V Symposium of ISBS) that wave processes of movement organization and energy transfer are observed in athlete's technical actions. The purpose of the present study is to assay the regularities of wave movement in javelin throwing.

The multi-link bar model of javelin thrower's body with hinge joint lower end free upper end with elastic hinge joint of links, distributed mass and rigidity parameters along links was used in this work. The motion of system was described by wave equations of forced transverse oscillations in partial derivatives of the fifth order.

To obtain experimental data shooting was with frequency 100 frames by camera "Actionmaster 500" and film processing by the film analyzer "Mac Sportias". Accuracy in determination of kinematic values was following: 1,5 % for displacements and 3 % for velocities.

The final phase of non-gliding javelin throwing made by elite athletes, was analyzed. Starts from the moment of putting the right leg upon the ground and continues until javelin release (0,7 s).

The spectral analysis of javelin thrower's movements showed the process of capacity distribution in spectrum on frequencies close to multi-link bar frequencies distribution. Spectral analysis of movements in javelin thrower's joints demonstrated expansion of spectrum frequency range and redistribution of oscillation capacity within this range towards high frequency harmonics from support to javelin (0-3 Hz for foot, 0-10 Hz for hand with javelin).

Wave movement spreads quickly along legs (10 m/s) and especially quickly along hands (10-15 m/s), more slowly along body (5-10 m/s). Wave movements oppositely directed are formed in upper and lower body extremities.

If all the javelin thrower's movement actions in running phase provide velocities of body, its links and javelin about 5-10 m/s, then thrower's actions in final throwing phase, organized in wave movement by the way of successive speed up of links from support to javelin accompanied by energy transfer along body provide increase of javelin velocity up to 15-20 m/s. Preliminary body speed up in the phase of running start unlike throwing (stand position) permits to use ballistic and inertial properties of links in progressive movement during wave movement formation.

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MECHANICAL POWER DEVELOPED DURING PULL MOVEMENT IN WEIGHTLIFTING

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The purposes of present study were to measure mechanical power outputs during pull movements in weightlifting and to evaluate the mechanical power related to athletic performance.

A dynamometer for the measurement of force, velocity and power was newly developed(Fig. 1). The loads were provided by adding the different weights of plates to the inertia wheel. Subjects pull the wire which was wound around the axis of inertia wheel. Instantaneous velocity and tension of the wire were respectively recorded by rotary encoder attached to the axis of inertia wheel and strain gauge mounted in the dynamometer. Mean velocity (MV) and mean force (MF) were respectively determined by integrating the velocity and force curves with respect to time. Instantaneous power curve was derived from instantaneous velocity and tension curves. Work and mean power (MP) were calculated by integrating the power curve(Fig. 2). Pull movement in weightlifting from starting position to final pull (S-1) was divided into three movements; i.e. the pulls in squat (S-2), hip extension (S-3) and shoulder shrug (S-4) movements. Mechanical variables were measured in each movement conditions. Subjects were twenty five weightlifters who consisted of ten Seoul Olympic lifters, five all Japan junior lifters and ten college lifters.

Mean values of MPs at the load of 15 kgw were 1024.4W, 999.9W, 561.8W and 299.3W at the movement condition of S-1, S-2, S-3 and S-4 respectively. In each movement condition, MF-MV and MF-MP relations were obtained from five different inertia loads. With increasing MF, MV decreased in hyperbolic manner and MP had a peak. In olympic lifter, higher velocity



Fig.1 Measurement of mechanical power during pull movement in weightlifting using a newly developed dynamometer (power processor).

and power were observed than those in junior or college lifters at identical MF level. Statistically significant positive linear correlation ($r=0.838$, $p<0.001$) was observed between MP developed during S-1 movement and performance expressed as the total weights of respective best records in snatch and clean and jerk (Fig. 3).

It was concluded that mechanical power measured during pull movement in weightlifting can well explain the performance of weightlifter.

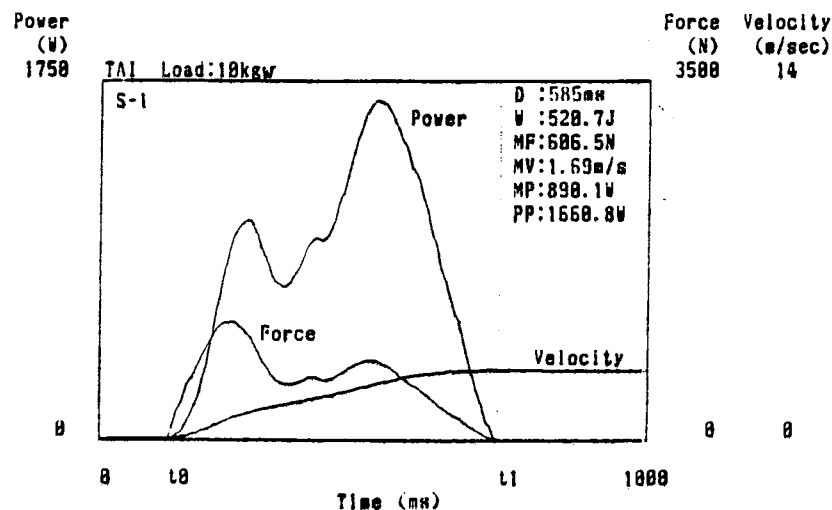


Fig.2 Typical recordings of instantaneous velocity, force and power curves.

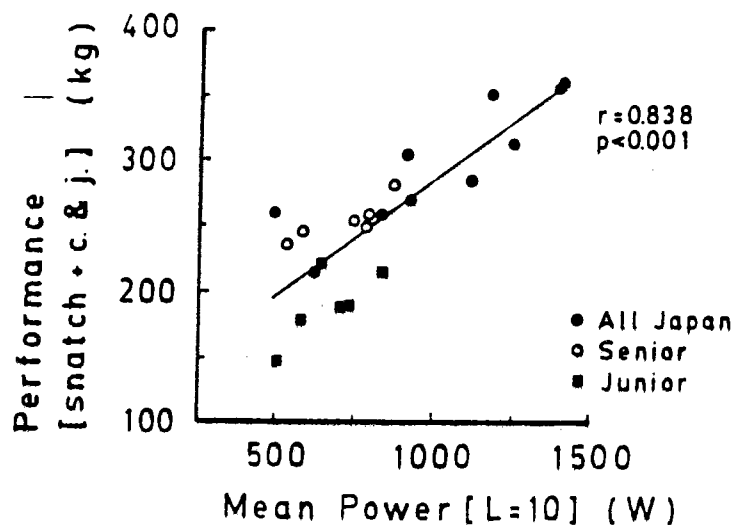
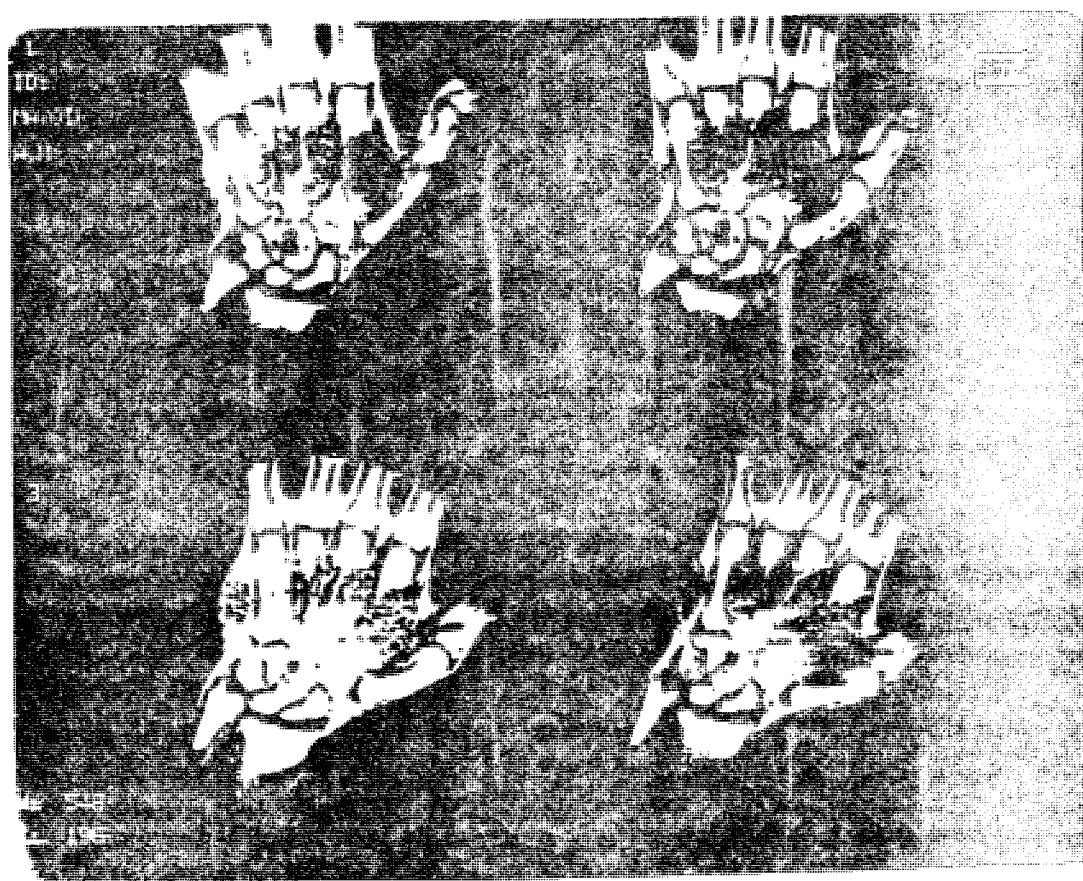
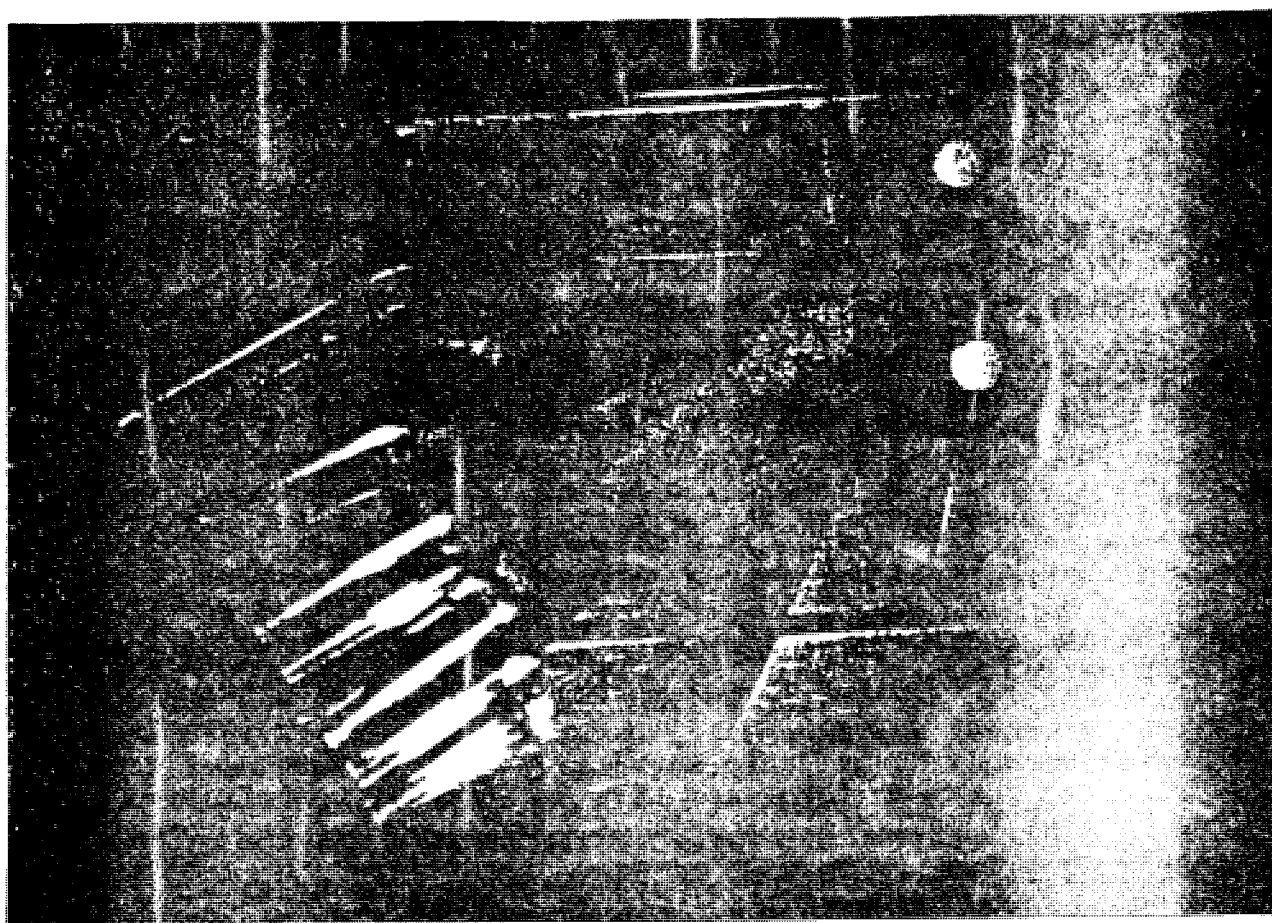


Fig.3 Relationship between mean power and performance in weightlifting.

KINEMATIC MRI: A NEW TOOL FOR DYNAMIC ASSESSMENT OF WRIST DISORDERS

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Wrist problems in the athlete remain a difficult diagnostic and therapeutic challenge. Epidemiologically wrist pain is most common in those sports that utilize the upper extremity for weight bearing including gymnastics, wrestling and football. Epidemiologically it is man's tedeologic evolution away from it's quadroped progenitor that adapts the wrist and hand to the function of fine finger movements and lifting. It is these sports that have high levels of acute and chronic compression tension and shear forces that result in the structural maladaptive response. Typically, the athlete has no pain at rest but motion especially coupled with the repetitive weight bearing causes pain and diminished in a performance. Recently, pathogenesis, diagnosis and treatment have been difficult by virtue of technical limitations. Diagnostics include plain films, arthrography and most recently static MRI. These studies are limited and do not simulate the pain eliciting motions. Kinematic MRI has been developed to define anatomic, pathoanatomic, kinematic and pathokinematic changes. The protocol is performed on a 1.5 Tesla magnetic and a specific coil is used as a transmitter and receiver. A plexiglass device is used to place the wrist motion during radial and ulnar deviation, flexion extension and concomitantly can load the joint. Spin echo pulses sequences were used generating T 1 weighted images. Acquisition parameters included two averages of 192 views from spatial revolutions and scan time of 1-2 minutes for each of six positions. It is then reformatted as a cine-loop and presented dynamically. Biomechanical parameters included carpal translations, centroid locations and motions of the proximal and the distal carpal row can be calculated. To date these protocols have corroborated previously defined carpal kinematics. Clinically, kinematic MRI has been applied to instability patterns, impingement syndromes, tears of the triangular fibrocartilage complex and idiopathic pain syndromes. It is the capability of providing a clinical analog in the face of defining anatomic and kinematic changes that make this technique an excellent scientific and clinical tool in defining wrist problems. The advantage of this technique is that it provides additional information regarding dynamic kinematic changes in the face of dynamic wrist disorders. This additional information can be potentially useful in further characterizing and defining the etiology of wrist pain, aids in surgical decision making as well as preoperative planning. In conclusion, kinematic MRI is an excellent adjunct for management of wrist disorders in the athlete.



COMPARISON OF VOLUNTARY AND OPTIMAL REGIMENS OF CYCLIC
MOVEMENTS IN PEOPLE OF DIFFERENT AGE

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Economy and performance are among the most common criteria of optimality of human motions. The aim of the investigation was to find out the difference between optimal and voluntary regimens of cyclic movements, chosen by people of different age and physical fitness. The study consisted of two series. 1) Voluntary chosen and most economic speed of even treadmill walking and running were determined in males of 6-60 years of age. 2) Voluntary chosen dynamics of power during maximal cycle ergometer work of 1.5 and 5 min duration was registered in subjects of 11-31 years of age. Every group consisted of 10-12 subjects.

Methods. Economy of cyclic locomotions was evaluated by energy cost of a unit distance. Energy expenditure was determined by summary energy production of oxygen, phosphagen, and lactic acid systems by modified (4) relations of Margaria (2) and Fox (1). Measured characteristics: 1) ergospirometry (gas analyzer Horizon, USA); 2) blood lactate (enzymatic method); 3) heart rate (Sporttester, Fin.); 4) speed of locomotions; 5) power of pedalling. Elucidation of optimal power dynamics was fulfilled by computer simulation of metabolic provision of movements (5).

Results (fig.1) proved that during even walking and running healthy adults voluntarily chose the most economic speed of locomotions, that confirmed minimal principle of Nubar and Contini (3). At the same time 6-7 year children in walking and 11-12 year boys in walking and running chose significantly higher speed. In patients with ischemic heart disease voluntary chosen walking speed was below energetically optimal.

Voluntary dynamics of mechanical power differed in sportsmen and untrained subjects (fig.2). Untrained chose almost constant power, but sportsmen - the most productive regimen "all-out". By analogy with minimal principle of biomechanics we suggest to call this law as "principle of maximal performance". According to it, individual trained in concrete mode of motor activity is capable to voluntary choice of optimal dynamics, providing the achievement of maximal performance in time-limit.

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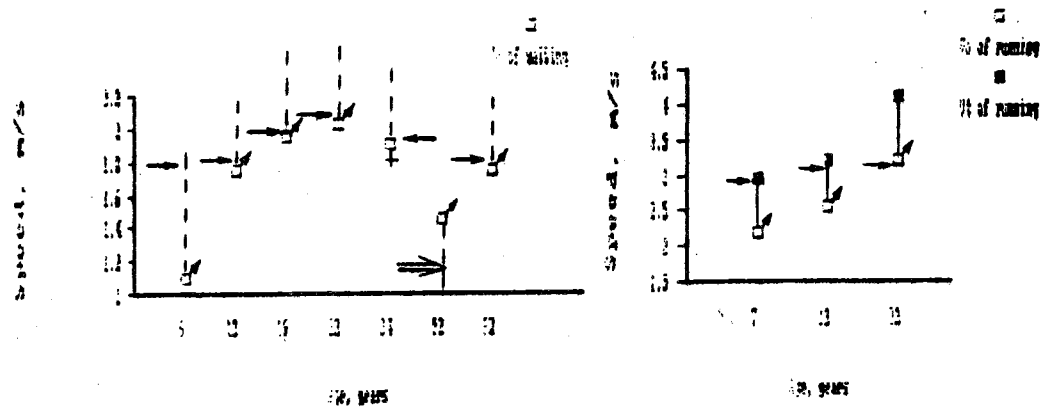


Fig.1. Speeds of walking and running, voluntarily chosen by subjects of different age and physical fitness (shown by arrows); empty squares - optimal speed, filled squares - anaerobic threshold speed; doubled arrow shows walking speed of patients with ischemic heart disease.

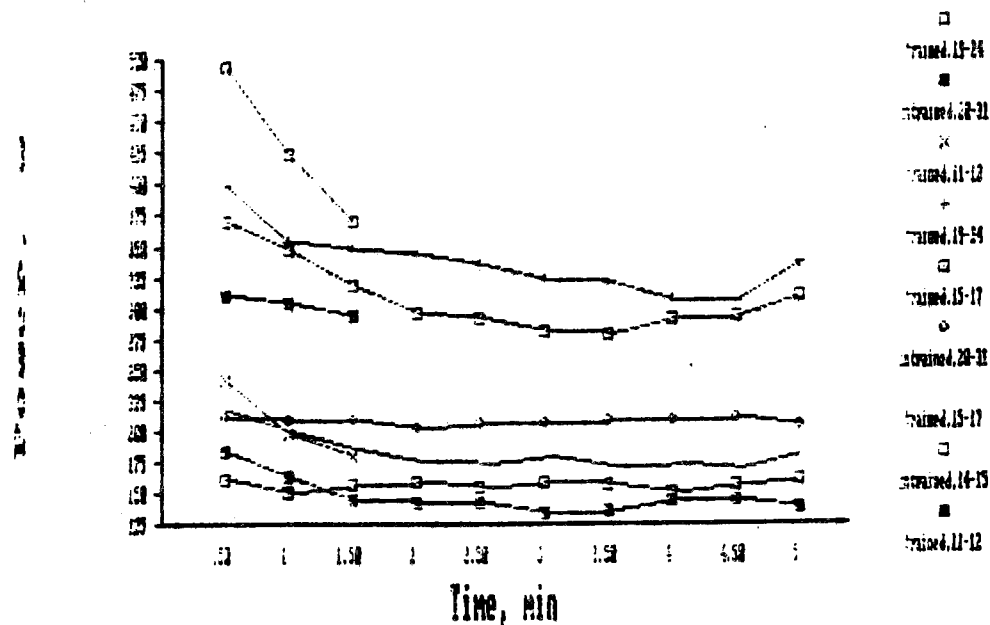


Fig.2. Mechanical power dynamics (mean by age group) during voluntary maximal pedalling of 1.5 min and 5 min duration in subjects of different age, sex, and fitness.

DESIGN PRINCIPLES FOR A ZERO-GRAVITY LOCOMOTION SIMULATOR

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The loss of calcium in the bones of the lower extremity remains a major concern in the planning of long term space flight. Data from the longest U.S. flight experience of 84 days (Skylab 4) indicated that the elevated levels of urinary calcium, which had signalled a negative calcium balance on previous missions, was maintained by all three astronauts throughout the flight with no tendency to plateau (Whedon et al 1977). Despite an in-flight exercise program, average mineral loss in the Os Calcis of the three crewmen was 3.9% while one individual lost 7.8% (Vogel et al. 1977). There is still considerable debate over the effectiveness of the various countermeasures that have been suggested to prevent the loss of bone mineral (Thornton 1989). Among the modalities under consideration for use in the Space Station are cycling, rowing, and locomotor exercise on a tethered treadmill. A major impediment to the examination of the various forms of locomotor exercise is the lack of a suitable ground based method to simulate weightlessness. Previous U.S. efforts in this direction have involved using simulators that either prevent the use of conventional gait analysis equipment or that interfere with the motion of the subject (e.g., Hewes 1969). If a suitable simulation of hypogravic locomotion could be achieved, it would be possible to experiment with different exercise protocols in order to determine those activities which maximize the forces that can be applied to the lower extremity in a reduced gravity environment.

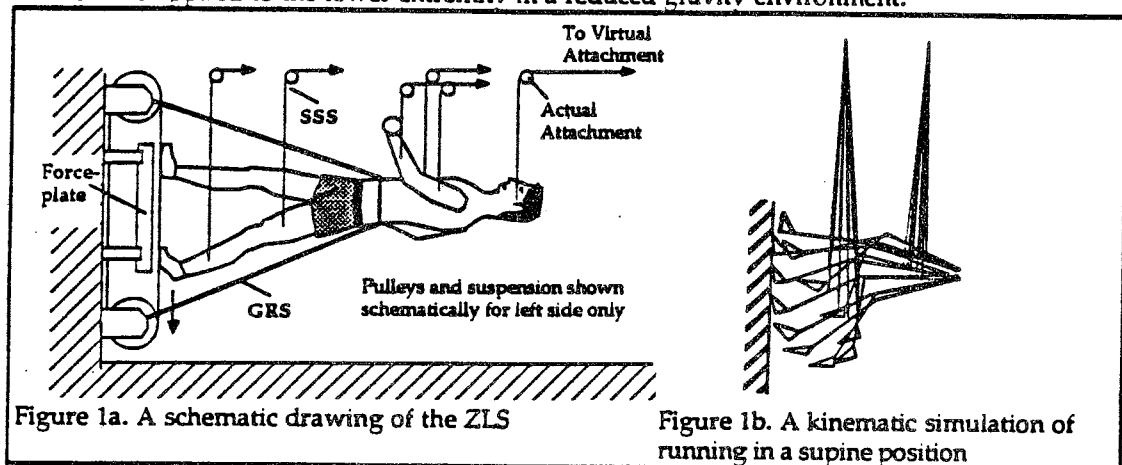


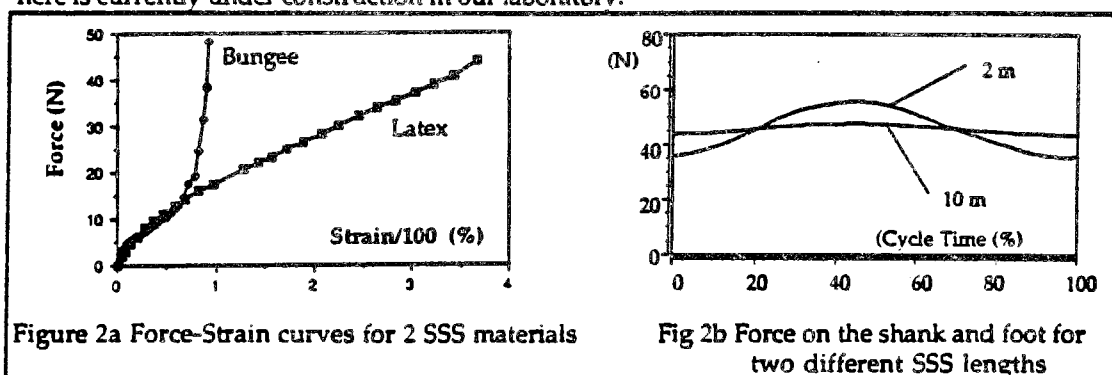
Figure 1a. A schematic drawing of the ZLS

Figure 1b. A kinematic simulation of running in a supine position

There is evidence in the literature that Soviet space scientists have used a supine suspension method to simulate locomotion in zero gravity (Grigor'yev et al. 1987). Although this simulator has apparently only been used for pre-flight training, it would appear that suitable modifications could be made to enable the device to be used for biomechanical experiments. The purpose of the present study was to examine the design principles of a supine treadmill for use as a Zero gravity Locomotion Simulator (ZLS). The basic components of the device are shown schematically in figure 1a. The subject is suspended in a supine position by elastic cords attached via close fitting sleeves to the segmental centers of mass. These cords form the Segment Suspension System (SSS). Their points of attachment are arranged such that, in the resting position, the subject's feet touch the surface of a treadmill mounted on a vertical wall. The treadmill can either be attached to the wall via force transducers or a force platform can be placed beneath the surface of the treadmill as described by Kram and Powell (1989). During any form of locomotor exercise in actual zero gravity, the subject will have to be restrained by a system, which we call a Gravity Replacement System (GRS), to accelerate the body towards the supporting surface. The same system is employed in the ZLS as shown in figure 1a.

The major task to accomplish in designing the SSS is to assure that each body segment is supported by a force approximately equal to its weight over the entire range of displacements and velocities encountered during exercise on the ZLS. During treadmill running at 3.4 m.s^{-1} , the distal end of the lower leg (approximated by the lateral malleolus) in a subject of average height exhibits a displacement along the AP axis about 1.5 meters. Figure 1b shows a kinematic simulation where lower extremity position data from treadmill running on earth has been rotated through 90 degrees so that the subject assumes a supine position. Straight lines have been drawn between the thigh and shank centers of mass and the points of attachment for the SSS above the subject. From this procedure it is possible to calculate the length of the suspension cables as functions of time and to determine whether or not the SSS would interfere with normal movement during the gait cycle.

Load displacement curves for various candidate suspension materials have been determined by conventional methods. Figure 2a shows two such results for latex tubing and bungee cord. The latex has the decided advantage that it can undergo considerable strain with relatively small increases in tension. When the measured mechanical properties are subjected to the displacements calculated from the kinematic simulation, the combined effects of SSS length, mechanical properties and segmental movement can be examined. If the virtual attachment is sufficiently distant, the suspension force on the thigh, for example, during the running cycle remains within 5% of the segment's 1g weight (see fig. 2b). A similar approach can be used to examine the required properties of the GRS. Based on the above experiments, it appears that the ZLS is a viable method to simulate locomotion in zero gravity and the device described here is currently under construction in our laboratory.



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A BIOMECHANICAL ANALYSIS OF HAND SPRING VAULT IN INDIAN FEMALE GYMNASTS

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The purpose of the present study was to elucidate the kinematic characteristics of hand spring vault performed by female Indian Gymnasts. A total of eight vaults performed were filmed and analysed for temporal, spatial and velocity parameters. The results of the present study reveal that the mean preflight time is more whereas the mean board contact time, horse contact time, mean post flight time and total vault time is lesser than that of 1984 olympians. The maximum height attained during the preflight and postflight time is similar to the gymnasts of USA. The horse departure velocity is lesser than that of the olympians. The results of the present study emphasised that each phase of the vault is important and the performance of one phase cannot be compensated by other phase.

Introduction

Handspring vault is the foundation for learning more complex vaults. Several workers have undertaken biomechanical evaluation of vaults performed by men and women gymnasts and have provided valuable information and guidelines for coaching(1-3). The purpose of present study was to determine the kinematic characteristics of handspring vault performed by female Indian gymnasts and to compare these with those reported in the literature.

Material and Methods

8 National level Indian female gymnasts took part in this study. A total of 8 vaults performed by them were filmed at 100 frames per second, using a Photosoni-IPL camera. The whole movement from the hurdle step to landing was recorded on 16 mm film. After processing the film, frame by frame analysis was carried out with the help of film motion analyzer (NzC-76). The centre of gravity of the gymnasts was determined using segmental method and was taken as reference point for the computation of various kinematic parameters.

Results and Discussion

The kinematic characteristics were studied under three areas i.e. temporal, spatial and velocity characteristics. These are presented in the table. The preflight time is 0.28s and is twice than those reported by Nelson et al.(1985) of 1984 olympians whereas the postflight time and the time of total vault were low. The vertical distance in the postflight was comparable to that reported by Rose-Hyvonen(1977) but was smaller than that reported by Nelson et al.(1985). The horizontal distance in the postflight is also lower than 1984 olympians (Nelson et al.,1985). The horizontal component of board departure velocity was lower and the vertical component was higher, whereas the horizontal and vertical components of both arrival velocity and horse departure velocity were lower than those reported for 1984 olympians(Nelson et al.1985).

As there are not much differences in the preflight phase but are in the postflight phase, the technique of arrival on horse and departure from horse needs to be improved to achieve better performance. Furthermore every phase is important and the performance in one phase cannot be compensated by other phase.

Table: Mean and standard Deviation of Selected Kinematic Parameters in Handspring Vault

Parameters	Mean±S.D.		Parameters	Mean±S.D.	
VELOCITY(m/s)			TEMPORAL (Time;s)		
1. Horizontal in last stride(foot-foot)	7.65	.36	1.Hurdle step	.29	.02
2. Horizontal 2nd last stride	7.65	.37	2.Board contact	.11	.00
3. Board arrival (horizontal)	6.49	.62	3.Preflight	.28	.02
4. Board arrival (vertical)	-1.44	.64	4.Horse contact	.17	.03
5. Board departure (horizontal)	4.47	1.11	5.Postflight	.57	.03
6. Board departure (vertical)	4.60	.48	6.Total vault	1.01	.04
7. Horse arrival (horizontal)	4.32	.55	SPATIAL(m)		
8. Horse arrival (vertical)	1.38	.79	a.Horizontal distance between contact points		
9. Horse departure (horizontal)	3.19	.64	1.2nd last stride	1.78	.08
10. Horse departure (vertical)	1.48	.43	2.Last stride	1.61	.08
			3.Hurdle step	2.91	.16
			4.Preflight	1.76	.06
			5.Postflight	2.03	.25
			b. Distance travelled by CG		
			6.Preflight (vertical)	1.93	.06
			7.Postflight (vertical)	2.27	.07
			8.Preflight (horizontal)	1.04	.05
			9.Postflight (horizontal)	1.73	.30

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AERODYNAMICS OF THE HUMAN BODY IN SPORTS

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The purpose of this study was to find the aerodynamic drag of the human body associated with different sports. In many sports, a large portion of energy is used to overcome air resistance. Therefore, an accurate prediction of the forces due to wind resistance is important to properly model sports such as running, cycling, speed skating, or skiing. This investigation will provide researchers with some necessary information on the drag and lift due to the air.

Many studies have been done on the aerodynamics of the human body, but most of them focus on only one sport. An accurate comparison of the air drag for different human positions studied by different people would be difficult due to experimental inconsistencies. Additionally, much of the data on human aerodynamics has not been published because of the competitive nature of some sports.

The results from more than forty tests provide comprehensive data on the aerodynamics of the human body for both lift and wind resistance. A life-size manikin and a human were used for wind tunnel models. Some positions measured with the manikin included running, long jumping, and cycling. The manikin was also used to measure the effects of hair, loose clothing, and crosswinds. The positions measured using a human included skiing, speed skating, standing, diving, as well many others.

The manikin, designed for use in the medical field, was proportional to a 5'8", 150 pound male. It had knees, hips, elbows, and shoulders which could be pivoted and tightened in place. Each joint was covered with a stretch bandage to smooth the air flow around it during testing. The human was 6'1" and weighed 165 pounds.

A force platform, manufactured by Advanced Mechanical Technology, Inc., was used to measure the forces in the wind tunnel for both the manikin and the human. The manikin was attached to the force platform with a vertical streamlined strut which was connected to the manikin's lower back. The strut was hinged at the top so that the manikin's body lean angle could be adjusted. The human stood on a wooden pedestal which was fixed to the force platform. All the testing was performed in a 10-foot diameter low-speed wind tunnel at California Technical Institute in Pasadena.

The pressure difference resulting from the airflow around a moving human is the main contribution to wind drag. Air drag is usually found using:

$$D = 1/2 \rho C_D A V^2 \quad \text{eq (1)}$$

where D is the drag force, ρ is the air density, C_D is the drag coefficient, A is the frontal area, and V is the velocity. The coefficient of drag is a dimensionless number which is a function of body position and size.

The tests were performed at speeds of 10, 15, 20, 25, and 30 mph. Rearranging eq(1) gives:

$$D/A = (1/2 \rho V^2) C_D \quad \text{eq (2)}$$

Performing linear regressions for each position tested on eq(2) using D/A as the ordinate and $(1/2 \rho V^2)$ as abscissa, returns the slope, C_D .

For the manikin, the correlation coefficients (r^2) for the linear regressions were all 0.999 except for one 0.998 (1.0 is a straight line). The correlation coefficients for the human were not as close because of the difficulty in holding and repeating the different positions, but the values were better than 0.99 except for two cases.

The table below shows a partial summary of the results. The drag coefficients varied from 0.56 for cycling to 0.92 for an erect standing position. A runner moving at 20 mph with a 3.5 mph crosswind has 8.8% more drag than the runner with no cross wind. A 7.3 mph cross wind raises the drag 12.1%. Large shoes raise the drag coefficient and the frontal area which both increase the wind resistance.

For the speeds and positions tested, the correlation coefficients have shown that the drag force increases linearly with the square of the velocity. This indicates that the drag coefficient does not change with speed. The results have also shown that clothing and hair can make a significant difference on the wind resistance of a human. Complete results of the tests are available from the authors.

Table: Partial results from the wind tunnel tests

Manikin	Frontal		Human	Frontal	
Position	Area (in²)	C_d	Position	Area (in²)	C_d
Runner	644.1	0.73	Ice Skating		
7.3 mph crosswind	644.1	0.82	Arms back	600.0	0.80
Large Suit	644.1	0.77	1 Arm back	642.1	0.75
Large shoes	663.4	0.74	Arms out	654.3	0.85
Long Jump	679.2	0.73	Skiing (crouch)	858.8	0.68
Standing	800.2	0.92	Standing	938.0	0.91
Cycling	484.7	0.56	Luge	208.3	0.62

SOME ASPECTS OF THE ANSYN-APPROACH TO MOTION ANALYSIS

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The complete analysis of an observed and recorded human motion encompasses the evaluation of all relevant kinematic and kinetic motion characteristics. In many applications, the kinetic characteristics, such as the joint moments, the components of the joint load vectors, and the segmental momenta, energies and powers are of particular importance to the analysis of the observed motion. However, the computation of these quantities requires the availability of the uncorrupted first and second time derivatives of the observed configurational function vector $q^*(t) = \{q_1^*(t), q_2^*(t), \dots, q_f^*(t)\}$, $t \in \tau$. In practice, the problem of determining these derivatives, in particular the second ones, is notoriously difficult because of the inevitable noise-contamination of the function vector $q^*(t)$ by systematic and random errors. This problem is particularly acute in three-dimensional motion analysis.

The present paper describes some aspects of an attempt to circumvent this problem altogether by replacing the solution of the notorious inverse dynamics problem by simulation of a specific kind. This approach is termed the ANSYN (Analysis-Synthesis) approach to motion analysis.

PROBLEM FORMULATION

Given an observed motion $q^*(t)$, $t \in \tau$, of an individual characterized by the set $\{P\}$ of subject-specific segmental, articular, and possibly myodynamic and myocybernetic parameters, find by synthesis of the individualized equations of motion (details in Hatze, 1981)

$$\dot{x} = A^{-1}(x) [B(x) + Q^E(x, t) + Q^C(t) + Q^M(t) + Q^L(x)], \quad x(0) = x_0, \quad (1)$$

where

$$x(t) = (q_1(t), q_2(t), \dots, q_f(t), \dot{q}_1(t), \dot{q}_2(t), \dots, \dot{q}_f(t)), \quad (2)$$

that motion $\hat{q}(t)$, $t \in \tau$, which minimizes

$$\int_0^\tau \sum_{i=1}^f \gamma_i (q_i^*(t) - q_i(t))^2 dt, \quad (3)$$

subject to certain constraints. The symbols γ_i denote weighting coefficients and f is the number of configurational coordinates.

This is the well known trajectory replication problem of optimal control with solution algorithms formulated by Jacobson (1968). A re-statement of the problem for gait applications was later given by Chao and Rim (1973). In the present problem, the muscular joint moments $Q^M(t)$ are the controls. The functions $Q^C(t)$ can be computed from the constraint equations (Hatzé, 1981) and $Q^E(x, t)$ are assumed given. It should be emphasized, however, that without the inclusion of muscle dynamics into the system (1) biologically unrealistic joint moments (e.g. bang-bang controls) may result.

GROUND REACTION FORCES IN THE ANSYN-TECHNIQUE

One of the great advantages of the ANSYN-technique is that environmental contact forces (e.g. ground reaction forces) need no longer be measured. They are computed directly from the dynamics of the motion by utilizing the equations that define environmental constraints. In practice this means that motions may be recorded without the subjects being restrained spatially by requirements of force plate contacts. In this paper we shall demonstrate one aspect of this technique by comparing measured and computed ground reaction force histories of a long jump analysis utilizing the 17-segment, 21-degrees-of-freedom body model of Hatze (1981) moving in the sagittal plane. Its segmental parameters (Hatzé, 1980) were determined for the 21-year old male athlete whose long jump motion was analyzed. Because of space limitations, only the vertical components of the ground reaction forces will be treated here.

In the present investigation, the optimal control problem described by expressions (1) - (3) was not solved directly because of its extreme complexity. Rather, the system (1) was simulated (program ANTOR) on the computer by guessing control functions $Q^*(t)$ until $q(t)$ was sufficiently close to the observed motion $q^*(t)$. The subjective judgment of closeness was based on a superposition of simulated and observed body model configurations on a video screen. The estimation of the controls $Q^*(t)$ was facilitated by functions $Q^{M*}(t)$ which were the result of a conventional motion analysis.

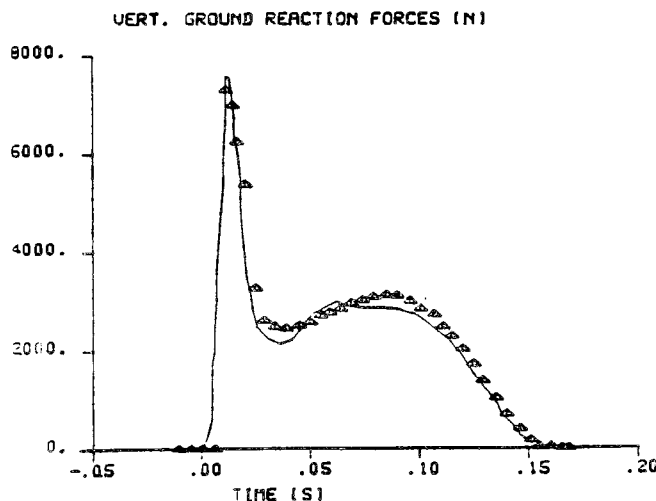


Fig.1: For explanation see the text.

From Fig.1 it is obvious that reasonable agreement exists between the measured (continuous line) and the simulated (symbols) vertical ground reaction forces. The larger discrepancies near 0.038 and 0.09 s may be attributed to the comparatively coarse method of visual motion superposition. Thus, the ANSYN-approach appears to hold some promise for future application in motion analysis.

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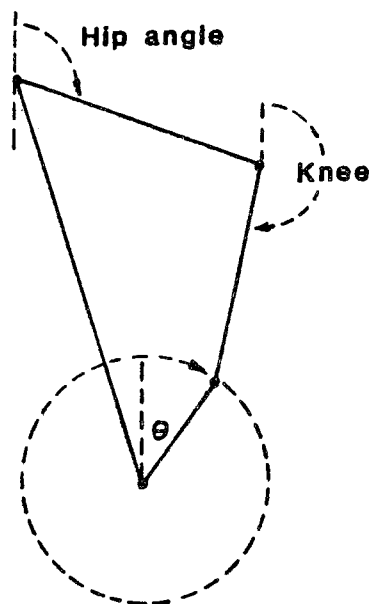
Using a Simulation Technique to Quantify Components of Hip and Knee Torques During Constant-velocity Cycling.

B.L.Davis

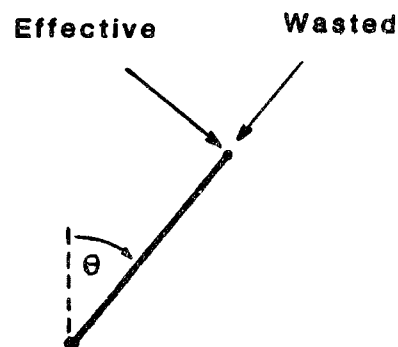
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It is generally the aim of every cyclist to maximise propulsion whilst minimising effort. In this regard it is commonly thought that by reducing the forces applied radially along the crank one could possibly decrease the "wasted" effort and in doing so, even increase the "effective" force perpendicular to the crank. The "criterion" diagram has frequently been used by cyclists to help them in this endeavor (Cavanagh and Sanderson, 1986).

The aim of this study was to take a closer analysis of these forces and in particular to calculate their effect on joint torques at the hip and knee. For this reason a mathematical model of a cyclist/bicycle combination was formulated using the simplifying assumption that it could be likened to a 4-bar linkage (in a way similar to that used by Andrews, 1987). The advantage of this approach was that the linear and angular acceleration profiles for the limb segments could be determined using explicit equations and thus not be susceptible to errors inherent in differentiating noisy kinematic data.



4 Bar Model



The input parameters required by the model were; geometry of 4-bar linkage, BSP's for thigh and lower limb, crank speed and components of force applied to pedal. The program then calculated hip and knee angles, acceleration profiles and joint forces and moments for each degree of crank rotation. The contribution of limb inertia to joint torque could be ascertained by setting crank forces to zero. Conversely, contributions of the "effective" and "wasted" forces could be determined by reducing crankspeed to almost zero. Typical results for this procedure (at 100rpm) are shown in the table below.

Angle	Hip Torque (Nm)				Knee Torque (Nm)			
	45°	90°	135°	180°	45°	90°	135°	180°
Inertial component	26	10	-19	-45	12	18	10	0
Component due to :								
wasted force	63	-74	-105	55	30	-55	-98	0
effective force	-14	154	220	63	-39	3	105	37
Total (N.m) :	73	91	96	74	3	-34	17	37

One interesting result of this study is that the concept of a radially-directed crank force being "wasted" needs further defining. At 135° for instance, were it not for the "wasted" component, the torques at the hip and knee would be 201 and 115 Nm respectively. Since neither of these magnitudes could be realised in practice the cyclist would be obliged to reduce the effective component and thus decrease propulsion.

By using the model in a similar way it is possible to ascertain the effects of increasing the "wasted" and "effective" forces. To demonstrate this, if at 135° the "wasted" component is increased by 20% the "effective" force can increase by almost 10% without requiring any greater torque at the hip or knee. This serves to illustrate that the torque that can be generated at these joints is probably not the limiting factor in determining the cycling speed.

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SIMULATION OF 3D MOVEMENT REGISTRATION BASED ON THE DIRECT LINEAR TRANSFORMATION METHOD*

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Introduction

In order to be able to determine the optimum experimental conditions for a particular movement research project, a computer program (DDD) was developed to simulate 3D movement registration based on the Direct Linear Transformation (DLT) method (Abdel-Aziz & Karara 1971). DDD is an extended version of the program that was used for an earlier paper on the effect of cinematographic parameters on the accuracy of 3D measurement using Subject Tracking Cameras (STC-method, De Haan & Den Brinker 1989). The extension not only concerns other registration techniques like high-speed video and pulsed IR-LED detecting systems, but also facilities to calculate the accuracy and computation time of various mathematical procedures to solve the DLT-equations. The program (in Modula-2) has an open structure, so that future extensions (in Modula-2, C, ADA or similar languages) may be added by both the designers and the users. The sources of noise related to the various recording techniques are modelled in the same way as was done by De Haan and Den Brinker (1989). For this reason the remainder of this paper will be dedicated only to the mathematical procedures.

Numerical solution of the DLT equations

In the DLT method, over-determined systems of linear equations arise in two contexts: while determining the DLT-parameters (if more than 7 markers are used) and in the process of calculating the 3 spatial coordinates (X,Y,Z) from the (at least) 4 DLT-equations. These situations can be represented as

$$Ax = b \quad [1]$$

where A is a $M \times N$ coefficient matrix (with $M > N$), b is a M -vector and x is a vector containing the N unknowns. In general, if A and b are the result of measurement, such a system does not have a unique solution. It is, therefore, necessary to find a N -vector x that minimizes the Euclidian norm of the residual vector $r = b - Ax$ (the linear least squares (LLS) problem). It can be shown (e.g. Stewart 1973) that the norm of r is minimized if x satisfies:

$$A^T Ax = A^T b \quad [2]$$

This square ($N \times N$) system is called the system of normal equations for the LLS problem. The simulation program exhibits two main strategies to solve the LLS problem:

1. Solving the LLS problem from its normal equations.

Once the normal equations are constructed, the system can be solved quite straightforwardly by Gaussian elimination (involving partial pivoting) and backward substitution. This is a fast procedure (see table, method I) that may be used in real-time applications, such as the Subject Tracking Cameras method. However, the procedure is numerically unstable, due to the fact that the matrix $A^T A$ in [2] has a tendency to be near singular. This leads to inaccurate solutions (see table, method I).

II. Solving the LLS problem using singular value decomposition.

A more stable procedure to solve the LLS problem is one that uses singular value decomposition, whereby matrix A is reduced to diagonal form by pre-multiplying and post-multiplying it by unitary matrices: $S = V^T A U$. The elements of S (the singular values) are in descending order from left to right along the diagonal. If the system [1] has one (or more) dependent equations, one (or more) of the singular values equal(s) zero. The singular value decomposition has important applications to the LLS problem, for it can be shown (e.g. Stewart 1973) that the Euclidian norm of $b - Ax$ is minimized if x satisfies:

$$x = A^+ b \quad [3]$$

where A^+ is the pseudo-inverse of A, defined as $A^+ = U S^{-1} V^T$ (In matrix S^{-1} all the non-zero elements of S are inverted.). This procedure has a higher accuracy than the one mentioned above, but it takes almost twice as long to evaluate (see table, method IIa).

A further refinement of this procedure is proposed in Van Huffel and Vandewalle (1985). Here, the Total Linear Least Squares (TLLS) method is described. This method, which is also based on singular value decomposition, is recommended if both matrix A and vector b in [1] contain errors. This is indeed the case in the DLT equations. The TLLS method is also implemented in the simulation program. At the cost of an increase of computation time a slightly more accurate solution vector can be obtained (see table, method IIb).

	method I	method IIa	method IIb
computation time	1.0	1.9	2.4
absolute error	0.92	0.24	0.17

Table: Relative computation time and absolute error [cm] for the various calculation strategies in a typical experimental situation.

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MUSCLE STRENGTH DEVELOPMENT IN BOYS AGED 7 - 14

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INTRODUCTION

The aim of this study was to estimate the natural trend of muscle strength pattern of boys of school age, in relation to their somatic and motor development growth.

The reason for this paper were the cognitive and practical. On one hand , the need of identification of level of strength generated by different muscle groups and defining the strength trend / dynamics and rate of growth / , and on the other hand , the need of refining norms of different muscle groups level of a population of Polish children and youth . These norms treated can then be as a biological reference system which is indispensable for controlling strength development in sport and therapeutic training.

MATERIAL AND METHODS

Maximal value of torques of flexors and extensors of right elbow , hip , and knee , also trunk , hand grip strength and torque of head-neck extensors were tested in static conditions with the use of strain gauge and mechanical dynamometry.

All measurements of torques were carried out in the special strength measuring stands : SMS-1 and SMS-2. It was assumed that the sum of torques of 9 tested representative muscle groups and the sum of relative torques is the parameter which generalizes muscle strength of human organism and its model of growth. Tests were of semilongitudinal character and they covered a population of several dozen boys , chosen at random. The research program lasted 4 years , with the same standard methods and measuring techniques.

Results of the tests underwent statistic - numerical analysis in which the following were calculated : full statistical characteristics ; significance of parameters' change ; inter - extrapolations ; structure of muscle strength and correlations of strength , somatic and motoric parameters and their model as well as growth prognosis.

The level of muscle strength asymmetry and ratio of torques of antagonistic muscle groups were also analyzed.

RESULTS

The highest increase of absolute values of torques was in hip and knee flexors - extensors ; it amounted about 519 - 338 %.

The smallest increase of all was in the group of head - neck extensors : about 160 %.
 The level of the sum of muscle torques increased by about 295 % , however , in relation to the sum of relative torques , only by about 74.9 % .
 Changes of absolute and relative values of hand grip strength amounted to about 205 % and 38 % , respectively.
 The progression of the running velocity and phosphagen power amounted to about 32 % and 252 % , correspondingly.
 The results confirmed our assumption about different models of strength of different muscle groups.
 It was confirmed that the process of strength development during this stage of ontogenesis is satisfactorily described by the exponential function form :

$$Y = a \cdot e^{bx}$$

where : Y - muscle torque
 e - base of natural logarithm
 a, b - factors
 x - time.

Accepting as a criterium of strength asymmetry the 10 % differences of muscle torque it was confirmed that the process of asymmetry was of a minimal character.
 The correlation of torques of antagonistic muscle groups / flexors - extensors / acting on the same levels of body segment was highly significant.
 The values of positive coefficients of correlation were from 0.419 to 0.659 , thus proving high influence of the level of force of antagonistic muscles on each other.
 Similar remarks result from an analysis of correlation of torques for the total combination of tested muscle groups and also from an analysis of correlation between the sum of torques and the hand grip strength.
 The sum of muscle torques is significantly and positively correlated with body height and body mass , LHM and phosphagen power.

**THE KINEMATICS AND KINETICS OF THE SHOULDER AND ELBOW JOINTS
DURING THE VOLLEYBALL SPIKE**

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Spikes of eight female intercollegiate volleyball players were filmed with two cameras. The direct linear transformation method of three-dimensional (3D) cinematography was used to obtain 3D coordinates of the ball and of 21 body landmarks. The resultant joint forces and torques at the shoulder and elbow joints were calculated, and various kinematic parameters were calculated to help describe the motions of the shoulder and elbow joints throughout the spike.

During the forward swing phase, the arm motion of some subjects was dominated by elbow extension, while internal rotation at the shoulder was the dominant motion in the others.

The external rotation of the upper arm was caused indirectly by the shoulder muscles that rotated the arm about an axis perpendicular to the upper arm and contained in the plane of the arm. The internal rotation muscles acted to prevent excessive external rotation. During the forward swing phase, the shoulder muscles acted to cause a forward somersaulting motion of the upper arm, as well as internal rotation. The elbow extension muscles acted to swing the forearm forward, in elbow extension. The varus torque exerted by bone-on-bone forces at the elbow acted to restrict the valgus rotation.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

AN ANALYSIS OF THE SPATIAL DISTRIBUTION OF MUSCLE FIBERS WITHIN THE TERRITORY OF A MOTOR UNIT

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INTRODUCTION: Based on visual inspection of glycogen depleted motor units (MU) the spatial distribution of fibers within a unit has been described as "random", "scattered", "irregularly spaced", "uniformly dispersed" and "quasi-Gaussian". Moreover, on the basis of these qualitative descriptions, investigators attempting to model the MU have made several assumptions regarding the distribution patterns of MU fibers. Some of these assumptions are: 1) that MU territories are circular in shape, 2) that the density of fibers within the territory is greatest at the center of the unit, and 3) that MU fibers are scattered randomly with few adjacencies between MU fibers. Clearly, more quantitative analyses are needed to adequately describe the distribution pattern of MU fibers. The purpose of the present study was to further describe the spatial distribution of MU fibers by evaluating the patterns which may exist over the entire MU territory. A detailed description of the patterns which exist in normal motor units should aid in the interpretation of single-fiber EMG data and provide a quantitative measure of assessing and diagnosing the involvement of degeneration and reinnervation in neurogenic muscle diseases.

METHODS: The muscle fibers innervated by a single motoneuron were identified using glycogen depletion techniques in 3 soleus (Sol) and 4 tibialis anterior (TA) units in 7 adult cats. Experiments were performed under pentobarbital anesthesia, 35 mg/kg, i.p., supplemented intravenously as needed to keep withdrawal and eye-blink responses suppressed. Glycogen depletion of the MU was achieved through repetitive stimulation of either the intracellularly identified motoneuron (Sol units) or the functionally isolated motor axon (TA units). Muscle fibers belonging to a stimulated MU were identified by their low level of glycogen which was assessed using the periodic acid-Schiff (PAS) reaction. From that cross-section which contained the greatest number of MU fibers, the x,y coordinate of the centroid of each depleted fiber was calculated and saved.

Three spatial analyses were used to study the distribution of muscle fibers belonging to a single unit. **ANALYSIS 1:** A quadrat analysis was used to assess the variability of MU fiber density within the territorial boundaries of the MU. For each MU, a rectangle was drawn to enclose the fibers belonging to the unit. This area was divided into quadrats and the number of MU fibers in each quadrat was counted and used to calculate the index of dispersion which is a sensitive measure for detecting a lack of homogeneity in a distribution of points. **ANALYSIS 2:** The distances between each MU fiber and all other MU fibers were calculated to provide additional information regarding the spacing of fibers within the territory. For each unit, a subarea of the territory was chosen and the interfiber distances between MU fibers were calculated. For each fiber in the sample area, the distance between it and every other MU fiber was calculated as the distance between the x,y coordinates of the centroids of each fiber. A histogram of interfiber distances was plotted and compared to a 95% confidence interval generated using Monte Carlo simulation techniques in order to assess whether the distribution of interfiber distances significantly differed from a random distribution. **ANALYSIS 3:** A point-to-nearest neighbor analysis was performed in order to test for the existence of "holes", i.e., areas containing no MU fibers, within the MU territory. In each MU, a subarea of the territory was chosen and points were randomly placed within this area. The distance from each random point to the nearest MU fiber within the subarea was calculated. The distribution of point-to-nearest neighbor distances was plotted and compared to a 95% confidence interval generated using Monte Carlo simulation techniques.

RESULTS: The spatial distribution pattern of MU fibers was analyzed in 3 slow-twitch Sol units, 1 slow-twitch TA unit and 3 fast-twitch TA units. Figure 1 shows the location of a fast TA unit within the muscle cross-section and distribution of the MU fibers within the outlined MU territory. In general, the fibers belonging to a unit were not distributed across the entire muscle cross-section, but were localized to a particular region of the muscle. The relative territory size of each unit, calculated as a percentage of the whole muscle cross-section, ranged from 8-76% with the Sol units extending over a larger percentage of the muscle cross-section than the TA units. In contrast, there were no significant differences in the absolute size of the Sol and TA MU territories. Sol and TA territories ranged from 1.6 to 4.7 mm². In general, MU fibers were not distributed evenly throughout their territory (see Figure 1b). Of the 7 units, 6 units (3 TA and 3 Sol) had significant values for the index of dispersion, indicating that MU fibers were distributed in a heterogeneous manner. The point-to-nearest neighbor analysis revealed that in 5 of the 7 units there was an excess of points with large distances to the nearest MU fiber. This result indicates the existence of holes in the MU territory. The calculation of all interfiber distances revealed a tendency for the MU to be more "grouped" or "restricted" than dispersed because of the predominance of short interfiber distances. This result was observed in 5 of the 7 units.

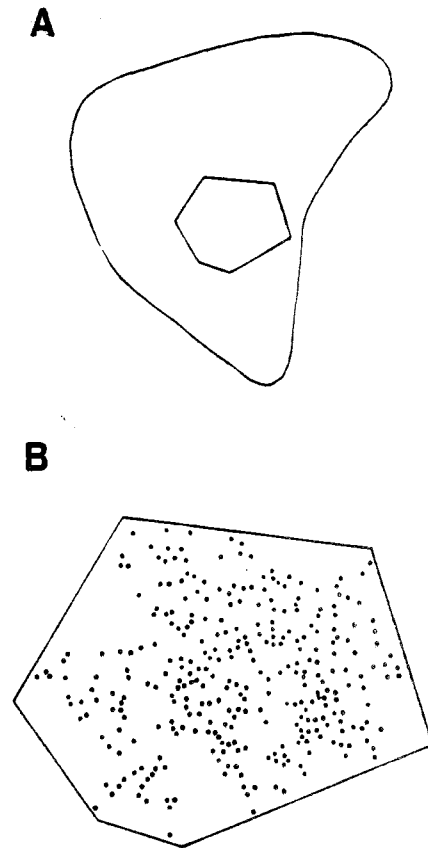


Figure 1
Location of a fast TA MU within the whole muscle cross-section (A) and distribution of all depleted fibers within the outlined territory of the unit (B).

SUMMARY: In the Sol and TA, MU fibers tended to be localized to rather discrete regions of the muscle, as opposed to being distributed across the entire cross-section. Within the MU territory, MU fibers were not evenly or homogeneously distributed. Instead, the grid analysis showed that there were regions of very high and low fiber density, with the high density area usually not coinciding with the center of the territory. Further, visual inspection of the units suggested that there were clusters or subgroups of fibers within the territory. The point-to-nearest neighbor analysis confirmed the existence of areas void of MU fibers, lending support to the hypothesis that there are groups of fibers which are more closely associated to one another than to other fibers in the MU. These subgroups of MU fibers may be a consequence of the branching pattern of a single motor axon. Additionally, the distribution of MU fibers within the cross-section may be a consequence of the processes that occur during synapse elimination during development. Supported by NIH Grant 16333.

ARCHITECTURAL AND FIBER TYPE DISTRIBUTION CONSIDERATIONS RELATIVE TO REPEATED BIOPSIES IN RHESUS MONKEY HINDLIMB MUSCLES.

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INTRODUCTION: Needle biopsies are used routinely to estimate the fiber type composition of individual mammalian muscles. Further, repeated biopsies have been used to assess the adaptability of the muscles under a variety of conditions (Saltin and Gollnick, Handbook of Physiology, Skeletal Muscle, Chpt. 19:555, 1983). Two implicit assumptions in these studies are that 1) the relatively small sample of fibers in the biopsy material is representative of the muscle cross-section and 2) the repeated biopsies are independent.

The purpose of the present study was to determine the feasibility of performing reliable, repeated biopsies in two plantarflexors, the soleus (Sol) and medial gastrocnemius (MG), and a dorsiflexor, the tibialis anterior (TA), of the Rhesus monkey. The consistency in the fiber type distributions within specified regions of the cross-section of each muscle at three levels along their proximo-distal axis and the relatively short muscle: fiber length ratios indicate a clear potential for the success of these studies.

METHODS: Three adult Rhesus monkeys were overdosed with sodium pentobarbital following a terminal experiment. On one side in two monkeys and on both sides in the other, the Sol, MG and TA were removed, cleaned of fat and connective tissue, weighed (wet weight) and prepared for qualitative histochemical analysis. Approximately 5mm thick blocks were taken from the proximal, middle and distal thirds of the muscle, mounted on cork and quick frozen in isopentane cooled to -160°C using liquid nitrogen. Subsequently, serial 10um thick sections were cut in a cryostat maintained at -20° . Fibers were stained either for myosin ATPase at an alkaline or acidic preincubation, succinate dehydrogenase or alpha glycerophosphate dehydrogenase and typed qualitatively as fast glycolytic (FG), fast oxidative, glycolytic (FOG) or slow oxidative (SO) as described by Peter et al. (Biochem. 11:2627, 1972). At each level, representative samples (200 to 400 fibers) were taken from the medial and lateral aspects of each muscle.

Prior to removing the muscles (see above), in two of the monkeys needle biopsies (approximately 10 to 30 mg) were taken post-mortem at a consistent depth (region) and at the same levels as described above in the proximal, middle and distal thirds of each muscle. Biopsy sites were chosen based on their accessibility in an *in vivo* condition. The biopsies were prepared for histochemical analysis as described above.

In two monkeys similar in size to those used for histochemical analyses, one hindlimb was removed above the knee and prepared for gross architectural determinations as described by Sacks and Roy (J. Morphol. 173:185, 1982). Briefly, the limbs were skinned, the muscles of interest freed from connective tissue and immersed in 10% formalin for fixation. Following partial maceration in 15% sulfuric acid, muscle, small bundle and fiber lengths were determined at several sites throughout each muscle. Isolated individual fibers were mounted on glass slides in glycerine and average sarcomere lengths were determined using a calibrated eyepiece at a magnification of x400. Fiber lengths were then normalized to a sarcomere length of 2.2 um to allow for intermuscle comparisons. Subsequently, average fiber:muscle length ratios were calculated.

RESULTS: As observed in a variety of mammals (Ariano et al., J. Histochem. Cytochem. 21:51, 1973) to include other strains of monkeys (Acosta and Roy, Anat. Rec. 218:136, 1987), the Sol had the highest (>85% SO in any sample) and the TA the lowest (<20% in any

sample) percentages of SO fibers among the three muscles studied. Generally, the fiber type percentages in each region of the Sol and MG were consistent (<10% deviation) along the length of the muscle. In addition, medial to lateral differences at any proximo-distal level were minimal (also <10% deviation) in these two muscles. The TA showed more variability than the Sol and MG across regions within an animal and within a region across animals. The right vs left limb comparison in one animal showed similar trends for each muscle. The size of the muscle biopsies ranged from ~100-400 fibers. The fiber type distributions of the biopsies closely resembled the regional data from the muscle cross-sections in each monkey sampled.

Average, normalized muscle fiber lengths were 32, 27 and 48 mm for the Sol, MG and TA, respectively. Fiber and small bundle lengths were generally similar throughout the muscle, with the highest variability observed in the TA. Fiber:muscle length ratios were 0.24, 0.27 and 0.32 for the Sol, MG and TA, respectively.

CONCLUSIONS: The consistency in the fiber type distribution within and across each region studied in the cross-section of the Sol and MG and in some regions in the TA strongly suggest the feasibility of taking reliable, repeated biopsies from these muscles in the Rhesus monkey. Similarly, the relatively small fiber:muscle length ratios indicate the possibility for repeated biopsies with little danger of sampling the same fibers more than once.

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MEASUREMENT OF FROG SEMITENDINOSUS MUSCLE FORCE AND TENDON LOAD-DEFORMATION AND LOAD-STRAIN PROPERTIES

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INTRODUCTION: An understanding of the interaction between muscle and tendon properties is important in elucidating their function during normal activity. A number of studies suggested that tendons perform functions which are more complex than simply transmitting muscle force to bones (see review by Proske and Morgan, 1987).

We previously modelled the relationship between sarcomere length and hip and knee joint angle in the frog semitendinosus (ST) muscle (Mai and Lieber, 1989). These data were used to model a normal frog hop using the kinematic data of Calow and Alexander (1973). This model assumed zero tendon compliance. In light of the studies mentioned above, we have measured the load-deformation and load-strain properties of the ST tendon in order to determine its influence on normal movement.

METHODS: Tendons from the distal insertion of the frog ST were dissected free of the muscle proximally and left attached at the tibial insertion site. The size of the specimen was approximately 0.5 mm wide x 5 mm long. The tibia was clamped and the proximal portion of the tendon stump glued and clamped into a specially-designed clamp to minimize stress concentration at the clamp site. Dye lines were then placed on the tendon with a gauge length of about 2 mm, and the tendon specimens were deformed at a rate of about 0.03 mm/sec corresponding to a strain rate of approximately 0.1 %/sec. Load-deformation curves were recorded in real-time, while load-strain curves were generated off-line using a video-dimensional analyser. In this way, for each specimen, a load-deformation and load-strain curve was generated.

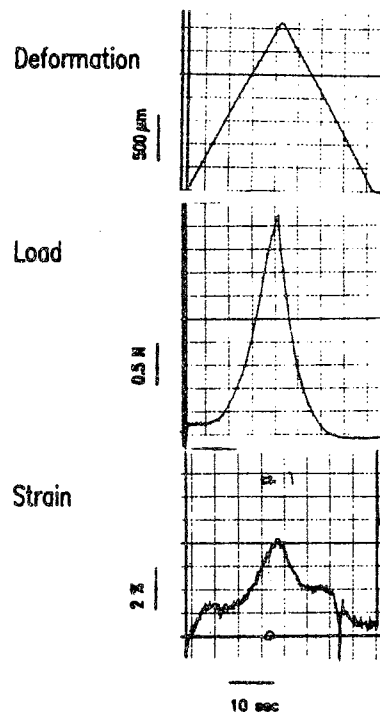


FIGURE 1

As a result of the difficulty in accurately measuring the specimen cross-sectional area, stress calculations were not made. Loads were then related to the maximum tetanic tension (P_0) of the isolated ST.

RESULTS AND DISCUSSION: A typical data set is shown in Figure 1 with records representing deformation (upper trace), load (middle trace) and strain (lower trace). The summarized data (mean $\pm$ SD) for all samples are shown in the Table. Load values for the load-strain table are also expressed relative to ST P_0 using a value of $P_0 = 0.3 \pm 0.05$ N (Lieber and Boakes, 1988).

In order to determine the relationship between muscle contraction and tendon properties, muscle fiber length, tendon length, and sarcomere lengths from Lieber and Boakes (1988) were combined with the load-strain properties measured in the present study. In addition, tendon slack length (Zajac, 1989) was measured directly as a function of muscle sarcomere length with the tendon origin and insertion intact. Slack length occurred at a sarcomere length of 3.1 ± 0.2 μ m.

Deformation (mm)	Load (N)	Strain (%)	Load (N)	% P_0
0.15	.25 $\pm$.19	1.0	.18 $\pm$.14	60
0.30	.61 $\pm$.50	2.0	.37 $\pm$.29	120
0.45	.87 $\pm$.28	3.0	.90 $\pm$.48	300
0.60	1.46 $\pm$.26	5.0	1.2 $\pm$.44	400

Tendon strained approximately 2% at P_0 (see Table). Thus, during a maximum isometric contraction (sarcomere length 2.2 μ m), sarcomere length would shorten to approximately 2.0 μ m. Current studies are underway to further model the relationship between tendon and muscle properties.

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EMG OF UPHILL RACING - A NEW QUALITATIVE APPROACH OF CYCLIC MOVEMENTS

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From viewing professional uphill-racing in European major competitions (Tour de France, Giro d'Italia,...) we know that different competitors take different positions, ranging from sitting backward on the saddle to pedaling while standing, to sitting on the forward top of the saddle. We know that lightweight climbers sit more forward than the more all-round "muscular type" of cyclist.

As saddle-height, crank-length and other frame-dimensions are individually designed according to the morphology of the cyclist, we can expect a correct positioning of most professional competitors for road-racing (zero degree inclination). Similarly, the same professional cyclist will use a different designed bicycle for track-racing, because the field situation has become different. The combination of the above allows us to assume that not only morphological variables determine the design of the bicycle, but that field variation such as uphill-cycling can be considered as an influencing factor too.

Summarized, we are dealing with a complex combination of ergonomics and cycling-position biomechanics specified to uphill-racing. This complexity is increased with the different interpretations of functional anatomy of two-joint muscles during the pedaling cycle (see Despires, 1974 ; Faria and Cavanagh, 1978 ; Sjogaard et al, 1982 ; Gregor et al, 1982, 1985 ; Andrews, 1987) and by the over-emphasis of the Lombard's paradox (van Ingen Schenau, 1981). Recently, Clarys et al, (1988) simplified the muscular movement model and denied the existence of the Lombard paradox in cyclic and rhythmic movements.

The purpose of this study was to investigate (i) the muscular activity of the cyclist's position in nine different circumstances, namely a combination of three saddle positions in three different road inclination simulations ; and (ii) the ergonomics of a newly designed climbing-bicycle with a moving saddle (Figure 1).

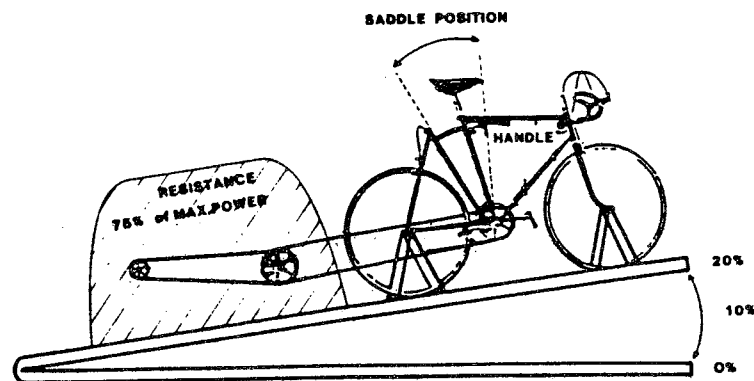


Figure 1 : Test set-up: bicycle with moving saddle and simulated road inclinations.

DATA ACQUISITION

To fulfil this aim, we measured a series of parameters simultaneously, such as EMG, crank forces, crank resistance, pedal frequency, heart rate, energy consumption, high speed film, video and position photography, and anthropometry-goniometry in different circumstances,

700

i.e. the combination of three saddle positions with three different slopes (0%, 10% and 20%). EMG is measured on-line on a seven channel FM data recording system (TEAC) with pre-amplified active surface electrodes and an additional portable amplification unit (Clarys and Public, 1987).

A six-channel portable regulation unit provided the supply for the pre-amplified electrodes and also took care of the supplementary amplification. Signal strength was adjustable from 0 to 40 dB to match the input sensitivity of the portable seven channel data recorder (TEAC HR30). This external calibration device allows adjustment (0.1-10V peak to peak) of all six channels to the individual EMG signal prior to the actual dynamic measurement. The synchronization of the pedaling cycle is generated electrically during each crank revolution at the (right) bottom death centre.

RESULTS AND DISCUSSION

Discussing all the results of this interdisciplinary ergonomic study would lead us too far, therefore we focus the discussion on the normalized (to the highest peak) muscle activity through its linear envelope, average EMG and dynamic component of the contraction. In addition, the qualitative muscular specificity between the nine test situations is verified using the IDANCO-system (Clarys and Cabri, 1988). The integrated EMG of the normalized linear envelopes is used as an indication of muscular intensity.

The dynamic activity of muscular activity is presented in circle diagrams divided in octants and representing the circular movement.

It is confirmed that the hamstrings are the prime movers of circular movements and we have found a decrease in dynamic muscular activity with the saddle forward on a 20% slope. A decrease that mostly is to be contributed to a significant decrease of arm muscles intensity. If we compare the 0% slope with the 20% slope, we see for both saddle forward and backward an increase of intensity with increasing slope. Although the oxygen uptake related to power (gross-efficiency) indicates for the saddle forward a decrease in efficiency in function of an increasing slope, we find no reflection of this phenomenon in the qualitative IDANCO-analysis of the different saddle and slope combinations. As a matter of fact the muscular patterns between the different situations are 95% analogue patterns, indicating that change of slope and change of saddle (position) does not affect the pattern of contraction but only the intensity. The cyclist's position with the saddle forward seems to offer the best muscular condition for cycling uphill, but it might be an explanation for the decrease in gross-efficiency, also.

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INFLUENCES OF ALPRAZOLAM ON EMG IN MAXIMAL VOLUNTARY EFFORTS.

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Anxiety is thought to be a feature of life in contemporary society. It is a subjective experience in which emotional and physical discomfort are associated with increases in the central and autonomic nervous activity, and, consequently, has some somatic expressions. Although benzodiazepines (BZ) are known for their anti-anxiety and muscle relaxant activity, it is still questioned how a centrally-acting drug can mediate this particularity at the level of the muscle itself. Since the discovery of BZ, various experiments in different areas have demonstrated their beneficial effects in depressing anxiety, in their relaxing and anticonvulsatory action, and with a limited rate of toxicity and side-effects. However, there is but little evidence of their influences when a human subject is put under abrupt physical stress such as all-out exercise. The basic question submitted in this study was: what is the effect of orally ingested doses of alprazolam (a BZ) on the electromyographic (EMG) signal during maximal voluntary effort?

METHODS.

Twenty healthy male subjects (mean age: 21.6 years (19.0-24.11), mean weight: 73.3 kg (66.2-87.5), mean height: 180.3 cm (165.5-196.5)) were asked to participate in the study. All subjects signed a written consent knowing the purposes of the study, the restrictions asked before each test session and the eventual risks of the protocol. The test protocol consisted of a randomized controlled double-blind protocol. The experimental group was divided into 5 subgroups (placebo, alprazolam 0.5, 1.5 and 3.0 mg) of 4 subjects to be tested on one day. In order to determine a standardized physical effort, a computer-linked isokinetic dynamometer (Kin-Com, Chattex Corp., TN) was used for the experiment. The subjects were asked to perform a maximal effort, both during concentric and eccentric knee extension at angular velocities of 3.66, 2.04, 1.05 and 0 rad.s⁻¹.

Three active electrodes, fixed on the M. vastus medialis, were used to measure the EMG output [gain: 10; CMRR: 90dB], the signals were then amplified [differentially; gain: 300; CMRR 85dB] and recorded on an FM-tape recorder for later analysis (Clarys & Publie, 1986). The signals (both EMG and dynamometer output) were low-pass filtered at 500 Hz [second order Butterworth] and sampled at 2000 Hz using the Electromyography Signal Processing and Analysis System (Cabri, 1989). The EMG signals of the respective angular velocity tests were full wave rectified and then normalized in time and in respect to the highest peak (Clarys & Cabri, 1988). Both integrated EMG (iEMG) and maximal EMG amplitude (mEMG) were averaged for all three electrodes.

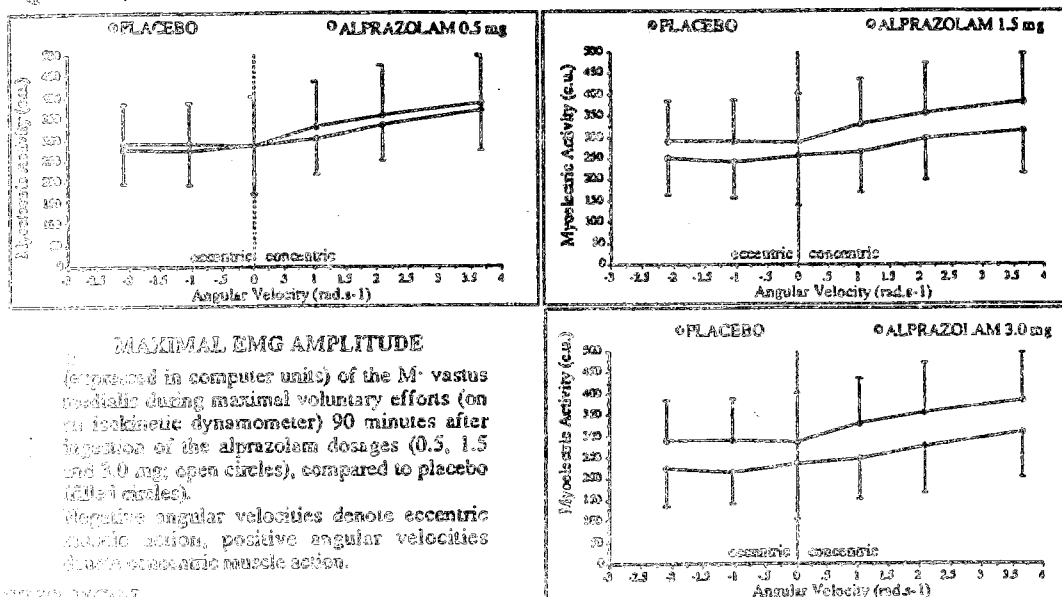
The statistical analysis included Kolmogorow-Smirnov test for the normal distribution of the data, analysis of variance, and a-posteriori protected least significant difference (PLSD) and Scheffé tests. The level of significance was chosen at $p < 0.05$.

RESULTS

The analysis of variance, pointed out that doses alprazolam administered were having a significant influence ($p < 0.05$) on the mEMG. Maximal EMG amplitudes were significantly lower after the ingestion of 1.5 and 3.0 mg alprazolam, when compared to placebo. When the alprazolam concentrations were compared, mEMG at 1.5 and 3.0 mg was also significantly lower ($p < 0.05$) than 0.5 mg. However, the 1.5 and 3.0 mg were found not to be different from each other ($p > 0.05$). After 360 minutes, significant lower activity was found for only for the 3.0 mg drug administration, in comparison to the placebo trials.

At all doses, the time point at which the tests were performed, showed a significant influence on the iEMG ($p < 0.05$). When the 90 and 360 minutes sessions were considered separately, significant differences were found between the drugs tested: after 90 minutes, when compared to the placebo trial, the 1.5 and 3.0 mg alprazolam dosages demonstrated significantly lower

mean iEMG values ($p < 0.05$). The lowest value was found after 90 minutes of administration of 3.0 mg, which was also significantly lower in iEMG when compared to the 0.5 mg alprazolam ingestion. After 360 minutes, these differences did not show any significance any more ($p > 0.05$).



DISCUSSION

The results from this study showed a clear trend, i.e. that ingestion of a dose of alprazolam decreased electrical activity (both maximal amplitude and iEMG) in a significant way. Plausible explanations of the mechanisms may be found in the action attributed to BZs. It is well documented that BZ enhance the action of gamma-aminobutyric acid (GABA), in that they amplify the inhibitory action of GABA (e.g. Man et al, 1986). When tension in the muscle is produced, the afferent Ia tracts will increase its release of GABA (presynaptically) and will cause an increasing influx of Cl^- , resulting in a hyperpolarization of the postsynaptic membrane. When this happens in the presence of BZ, the influence will be greater, for the opening of the channels will be more pronounced, and Cl^- will be increased. Thus, postsynaptic inhibition, whereby the release of the neurotransmitter (i.e. GABA) into the mono-neural synapse, leads to hyperpolarization of the post-synaptic membrane and a subsequent reduction in the ability of incoming excitatory impulses to depolarize the motoneuron. But presynaptic inhibition, that is inhibition whereby the nerve terminals are partially depolarized so that incoming impulses are less effective in releasing the excitatory neurotransmitter, is also forwarded as a causal mechanism (e.g. Young & Wiegner, 1986). The observed relaxing effects are reduced after 360 minutes, indicating efficient clearance of the effects of the drug.

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THE PATHOGENY OF ANKLE SPRAIN: EVALUATION OF A HYPOTHESIS

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Ligamento-muscular protective reflexes are assumed to prevent ankle sprain. This hypothesis can be tested by determining: 1) the reflex latencies of the foot evertors in response to an inversion movement and 2) the time to produce an ankle sprain during a varus movement. The latencies are measured on 7 healthy subjects. The second delay is obtained by simulating the landing of a vertical jump (± 30 cm) where one foot lands on a pavement kerb. To implement the computer simulation, we investigated, in 3 healthy volunteers, the relationship between in vivo torque and angle both in static and dynamic conditions, with the evertor muscles—either relaxed (passive torque) or during maximal voluntary contraction (active torque).

The results show that—during this ankle sprain situation, (1) the peri-articular passive structures are unable to absorb the mechanical energy of the inversion sprain, and (2) the time necessary to reach the angle of capsular-ligamentous rupture is inferior to the reflex latencies we observe. We have performed several simulations with variations of the model parameters and it seems that in this particular situation, an ankle sprain cannot be avoided.

Two-page abstract not received in time for publication.

Short abstract is reproduced to left.

AN APPROACH TO A BIOMECHANICAL PROFILE OF TOP ATHLETES'
TAKE-OFF TECHNIQUES IN HORIZONTAL AND VERTICAL JUMPS

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Purpose. The purpose of the study is the establishment of a biomechanical profile for horizontal and vertical jumps in athletics. Whereas horizontal jumps are characterized by the long jump and the triple jump, the high jump is the classical paragon of vertical jumps. The approach also tries to analyze characteristics of the long jump and of the high jump which are determined by sex. The paper deals with athletes on the highest international level, including the finalists of the Olympic Games in Seoul 1988. In addition data from the World Championships in Athletics in Rome 1987 are also taken into account.

Method. The jumps of the finalists are recorded with three (in triple jump and long jump), resp. with two (in high jump) synchronized highspeed cameras (LOCAM). The nominal camera speed is 200 fps. The control of the panning angles is guaranteed by numerous landmarks in the object field; their positions are well marked and defined. For the analysis the film has to be digitized frame by frame. A 14 segment model of the human body is used. The arrangement of at least two cameras facilitates a three-dimensional analysis of each of the body points. The determination of the centers of masses of the segments and of the body is followed by a calculation of translatory and rotatory velocities and accelerations. Before their differentiation the raw data set is digitally smoothed by the help of a second order Butterworth filter.

Results. The factors used for the establishment of the profile are based on models which have been presented inter alia by HAY (1985) and HAY et al. (1986) for the disciplines dealt with. The analyses show that the differentiation of athletic jumps in approach, take-off preparation during the last three run-up steps, take-off and flight, as well as landing (including further take-offs in triple jump) is a quite effective one. This effectiveness is well-founded both from a biomechanical and a sports-practical perspective.

It is remarkable that maximum run-up speed, which has been brought out as the classical and most important factor in horizontal jumps, does not allow any differentiation in our top-sample in regard to maximum performance. The only exception here is LEWIS. A statistically significant correlation between jumping distance and maximum run-up speed can neither be found in the finalists of the Olympic Games in Seoul nor of the World Championships in Athletics in Rome.

This shows that maximum run-up speed in athletics on the highest level of performance becomes a necessary, however, not the sufficient prerequisite for successful jumps. In high jump, too, it is not possible to identify a definite correlation between maximum run-up speed and performance in the finalists examined. Take-off preparation and the take-off as such serve as vehicles to explain differences in performance. Interesting trends of preparatory movements in the frontal plane, aiming at favourable take-off conditions (inter alia for the prolongation of the path of acceleration) are noticed and taken into account as constituent elements of profile. In our analyses the phase of preparing the take-off is given special attention because of our applied perspective.

The degree of profile elements, foremost of the take-off and of the direct take-off preparation, in all the jumping disciplines examined, show remarkable inter-individual differences. Last not least for this reason it is tried, on the basis of multiple trials of several athletes, to establish the intra-individual variability, resp. the degree of profile. The approach supplies most interesting results when comparing the data of the World Championships in Athletics in Rome 1987 and of the Olympic Games in Seoul 1988 in top-athletes such as KOSTATINOVA in high jump and LEWIS in the long jump.

POTENTIAL MECHANISMS FOR ENHANCEMENT OF MUSCLE FORCE-VELOCITY RELATIONSHIPS DURING MOVEMENT

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INTRODUCTION

In-vivo recording of muscle force and length changes during overground locomotion in cats has indicated a significant enhancement of force-velocity properties over those predicted from in-situ isotonic shortening experiments (Gregor et al.,1988). The enhancement was of a magnitude sufficient to double the power output of the soleus muscle. Experiments were therefore undertaken to verify these findings in an in-situ preparation by simulating the lengthening-shortening cycles measured during locomotion on tetanized muscles and calculating the resultant force-velocity relationships. Further experiments were then undertaken to investigate possible mechanisms for such an enhancement.

METHODS

Rat soleus muscles were tetanized in-situ and subjected to controlled lengthening and shortening excursions using a Cambridge lever system. The imposed length changes were simulated from digitized step cycles or generated from a triangular waveform. All waveforms were adjusted to provide a maximum muscle length of L_0 (Optimum muscle length for the isometric tetanus). The stretch phase of the triangular waveform was adjusted to give a range of stretches from none to a maximum of 5mm. The velocities of the imposed stretch and shortening were chosen to provide representative records for all regions of the force-velocity curve.

RESULTS

Lengthening-shortening cycles similar to those observed in the cat soleus muscle during locomotion confirmed that force-velocity properties may be greatly enhanced in locomotion.

Quick release of a tetanized muscle to an isokinetic shortening velocity without prior stretch also enhanced force-velocity properties during shortening (Figure 1a). This enhancement occurred in the initial shortening phase when force had not reached a constant level.

Stretch of a tetanized muscle prior to release to an isokinetic shortening produced an elevation of the initial tetanic tension to values considerably greater than P_0 plus an even greater enhancement of force velocity properties (Figure 1b). Again, the enhancement occurred during the initial shortening phase when force had not reached a constant level. Elevation of tetanic force due to stretch was transient, decaying to normal tetanic levels in approximately 50ms if the stretched position was held.

DISCUSSION

The force-velocity curves described in this work are not the classic isotonic force-velocity curves described by Hill (1938), describing a shortening contraction as a single point. In this case a single contraction is described by its own curve and therefore shows the changing force-velocity relationship throughout a single contraction. By definition, the isotonic shortening velocity is measured when a constant force level is apparent. Measurements under these conditions indicated normal force-velocity relations.

All of the force enhancement observed in these experiments was found to occur in the initial phase of the shortening where force was falling rapidly. It has been suggested that this phase of shortening is purely elastic (Civan and Podolsky, 1966) and may correspond to the release of stored elastic strain energy as force in the muscle decreases. It should be noted that release of stored elastic energy can only occur as force falls since the elastic energy stored is proportional to force in the system. Any movement which is accompanied by a fall of muscle/tendon force utilizes stored elastic energy and if it is desirable to

increase utilization of stored elastic energy then the drop in force throughout a movement must be increased (cf. Avis et. al., 1986).

Greater enhancement after stretch must be due in part to the increased force generation within the muscle providing greater stresses on elastic elements, thus increasing storage of elastic energy. The transient nature of this phenomenon has been reported by several authors (eg, Cavagna and Citterio, 1974), indicating that if countermovements are used to enhance work output, any delays before a desired activity will result in a loss of performance.

CONCLUSIONS

Any compliant material will store energy when stressed. This energy will only be released when the stress is decreased. A prerequisite of any movement utilizing release of stored elastic energy must therefore be that stresses (muscle forces) must decrease throughout the energy recovery phase.

Pre-stretch greatly enhances muscle force development, therefore providing higher stresses on compliant elements with a resultant increase of stored elastic energy.

In the design of activities to take advantage of stored elastic energy it should be remembered that energy release can only occur as stress in the system falls and if countermovements are made, they should be designed to maximise stresses on series elastic elements and should be followed quickly by the power stroke of the movement.

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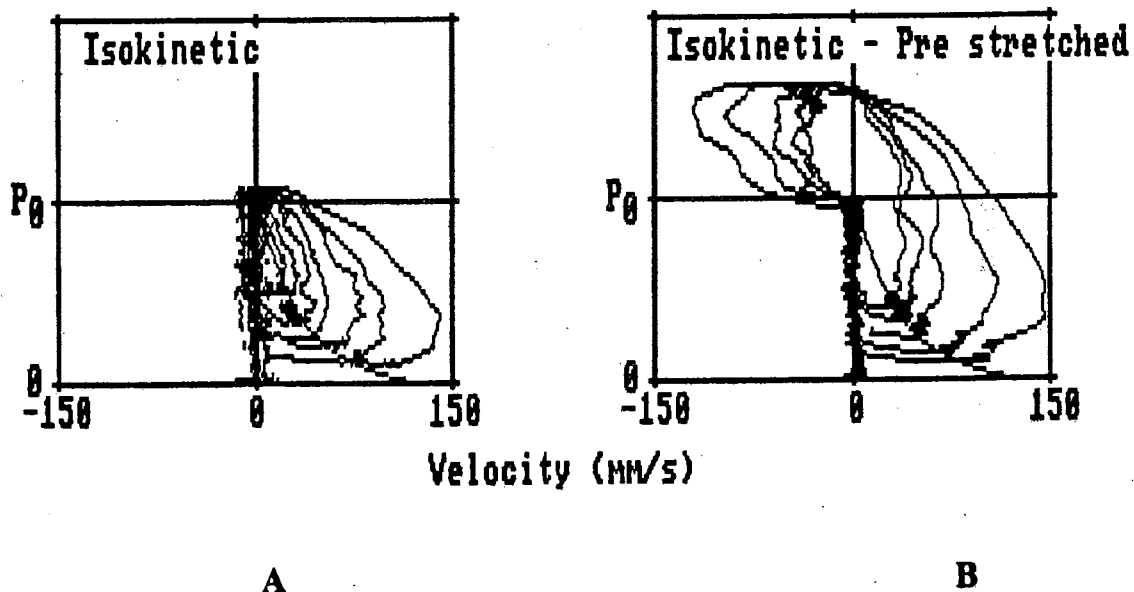


FIGURE 1

Superimposed force-velocity curves of individual contractions in quick release experiments to isokinetic shortening velocities. The isometric force-velocity curve is indicated on each figure. All releases were from L₀. a) without pre-stretch b) with a 2.5mm pre-stretch.

The Contribution of Individual Muscles to the Ankle Moment Produced in the Cat Hindlimb

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INTRODUCTION - Principles of inverse dynamics have been used to determine joint moments produced during movement. Using this approach, a generalized muscle moment (GMM), dominated by muscular forces acting about the joint, may be determined. The contribution of individual muscles to this moment has been estimated using EMG-driven models or optimization algorithms. Muscle forces, however, can be directly measured in freely moving animals using transducers, chronically implanted on muscle tendons. The purpose of this investigation was to compare moments developed by individual muscles with the generalized muscle moment at the ankle during the stance phase of locomotion.

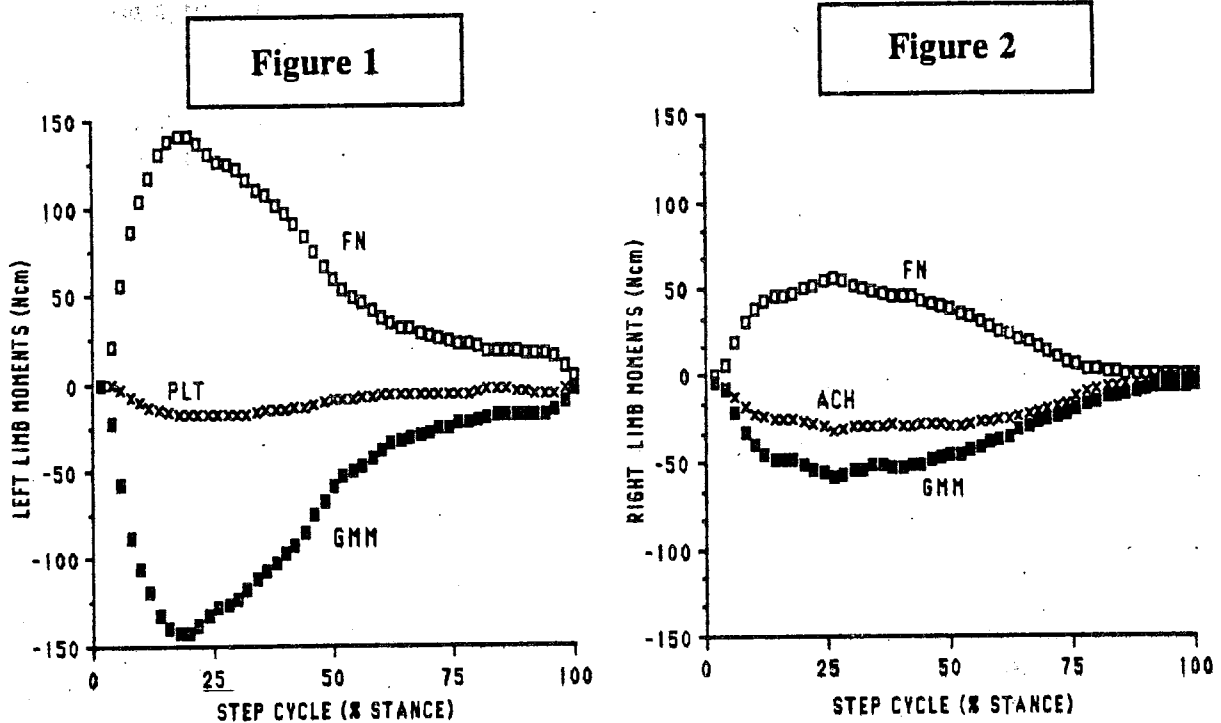
METHODS - Four adult male cats (weight = $4.2 \pm .57$ kg) were trained to walk within a plexiglass enclosed walkway. Tendon force transducers were placed, under sterile conditions, on hindlimb muscles randomly selected for each animal. Muscles examined included soleus, medial gastrocnemius, lateral gastrocnemius, plantaris, and tibialis anterior. The entire Achilles tendon was also examined in one animal.

Force recording and high speed filming occurred approximately one week following surgery. Both hindlimbs were shaved and markers placed on the iliac crest, greater trochanter, approximate knee joint center, lateral malleolus and base of the fifth metatarsal. The cat was placed in the enclosed walkway which contained two concealed force platforms placed in succession on the side of the walkway floor closest to the camera. Each force plate contained two piezoelectric elements capable of measuring vertical and horizontal components of the ground reaction force. Limb movement was filmed at a rate of 100 fps, by a camera placed orthogonal to the plane of movement. Film, tendon forces, and ground reaction forces were synchronized using a DC pulse and light in the camera field of view. Each tendon transducer was calibrated during a terminal experiment *in situ*. The limb was secured by pins and clamps and the force transducer left undisturbed on the tendon which was connected distally to a force lever. The tendon transducer output was then compared to the calibrated lever output during a series of muscle contractions of varying intensity. A calibration factor was obtained for each buckle transducer (Sherif *et al.*, 1983).

A photographic method was used to determine the instantaneous center of joint rotation and ankle muscle moment arms for ranges of motion observed at the ankle and knee *in-vivo* (Lieber and Boakes, 1988). Individual muscle moments were then computed by multiplication of muscle forces by their corresponding moment arms. Coordinates (x,y) of skin markers were obtained from serial film frames. Each frame was projected onto a screen and limb position-time data calculated from the digitized coordinates of segment endpoint markers. Kinematic data were smoothed using a digital filter and derivatives computed via finite differences (Whiting *et al.*, 1984).

Ground reaction force records were acquired on line to a IBM PC, multiplied by appropriate conversion factors, and point of force application on the plantar surface of the paw calculated. Anthropometric data (segment mass, center of mass, and moment of inertia) were obtained for each segment using regression equations with cat mass and segment lengths as predictor variables (Hoy *et al.*, 1985). The cat's shank, hindpaw, and forepaw were modeled as a planar, three-segment, rigid body system with ground reaction forces acting at specific locations along the paw. Equations of motion were formulated using Newtonian mechanics and the generalized muscle moment, representing the net effect of all muscles and periarticular structures acting about the joint center, determined for the ankle.

RESULTS - Exemplar data are shown in Figs. 1 and 2 for one animal whose right Achilles and left plantaris were instrumented. Peak moments due to the vertical component of ground reaction forces (FN) ranged from 56.2-157.4 Ncm. As acceleration of limb segments was minimal during stance, the GMM was essentially a mirror image of these moments. TA moments were approximately zero during most of the stance phase. Peak plantaris moments (PLT) ranged from 13-15% of the generalized muscle moment (19-21 Ncm). Peak Achilles moments (ACH) varied from 39-56% (33-37 Ncm) of the GMM. The difference in magnitude between achilles and generalized muscle moments was unexpected. Additional *in situ* achilles transducer calibration experiments were subsequently performed which compared standard muscle force transducer calibrations with those obtained while varying the relative contribution of extensor tendons within the achilles complex, more closely simulating *in vivo* conditions. The results suggested that the calculated *in vivo* achilles moments were underestimated using the standard calibration procedure.



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USE OF SURFACE AND FINE WIRE ELECTROMYOGRAPHY IN THE STUDY OF LOWER EXTREMITY FLEXORS DURING CYCLING

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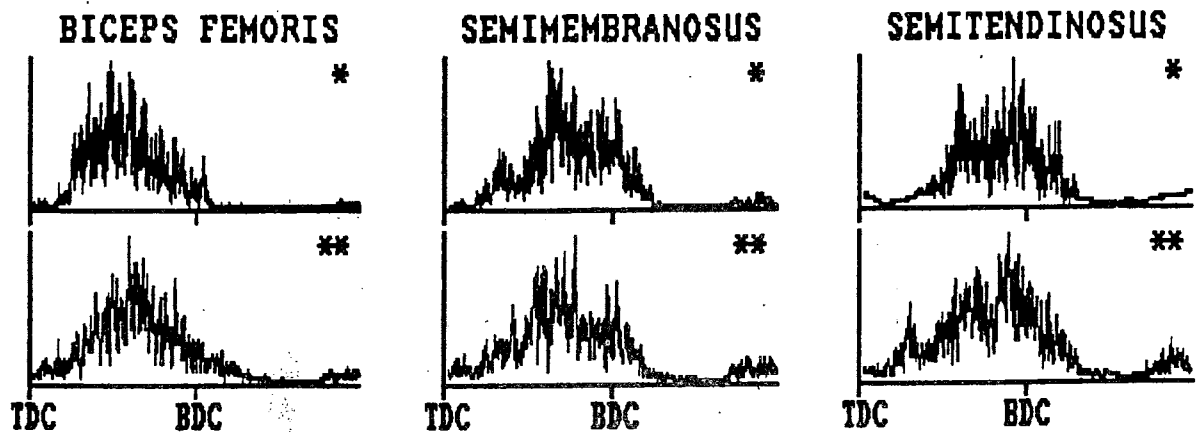
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INTRODUCTION - The pattern of electrical activity of muscles (EMG) provides useful information regarding the assessment of neural control of muscle function. While either surface or indwelling electrodes are commonly used to obtain this information, surface electrodes have been employed exclusively in the collection of lower extremity EMG during cycling. Comparison of data reported by a number of investigators (Jorge and Hull, 1986; Gregor et al., 1982) indicates that EMG patterns observed during cycling in major lower extremity flexors and extensors are repeatable. This suggests that the data reported are reliable, and accurately reflect how the muscular system is being controlled by the nervous system. However, recent surface EMG data reported by Ryan et al. (1988) indicated that although the EMG patterns for the extensors were relatively immutable, the flexor muscles showed a great deal of variability. This variability was most pronounced in the biceps femoris (BF) muscle, where two separate activation patterns were observed. While this contradiction with previous reports may be explained, in part, by individual rider variation and experimental conditions, further comparison of EMG patterns is needed. To further study the issue of variability, selected lower extremity flexor EMG patterns were monitored using both fine wire and surface electrodes in an effort to compare output obtained from each preparation with output previously reported in the literature.

METHODS - Five recreational cyclists rode a racing bicycle at 90 RPM (250W) at their preferred seat height. The bicycle was mounted on a Velodyne Trainer which simulated inertial and non-inertial loading. Fine wire electrodes (50 μ m) were inserted into the muscle bellies of the long head of biceps femoris (BF), semitendinosus (ST), and distal semimembranosus (SM). Electrical stimulation was used to verify placement of the wire electrodes. A bipolar pair of silver-silver chloride surface electrodes was placed over each muscle approximately 1cm either side of the insertion of the indwelling electrodes. EMG data were collected from surface and fine wire electrodes for two muscles simultaneously. Connections were rotated through subsequent pairs to insure that records were obtained from each muscle at least twice during the experimental sessions. Data were sampled at 2KHz, conditioned by Universal amplifiers (Gould) with a bandwidth of 30-300 Hz on line to an Amiga 2000 microcomputer. Signals were full wave rectified and spike trigger averaged across 10 pedalling cycles. Average patterns from both surface and fine wire electrodes were compared for relative timing and burst duration. High speed film (100 f/s) was taken during each trial. Joint angle changes across cycles were obtained from the kinematic data to verify consistency of pedalling action across trials.

RESULTS - A comparison between surface and fine wire EMG collected from the same muscle shows that the relative timing of EMG signals from each electrode pair was approximately the same. Burst duration from the surface electrodes appears longer than that collected from the fine wire electrodes (Figure 1). Additionally, both surface and fine wire electrodes recorded a small burst of EMG just prior to top dead center in the BF and SM but this burst was observed only in the surface recording of the ST.

FIGURE 1 - Fine wire (*) and surface electrode () recordings for BF, SM, and ST from top dead center to top dead center (TDC).**



CONCLUSIONS - During cycling, the flexors of the lower extremity show the greatest variability in activation patterns. EMG activity collected from the knee flexors via surface electrodes was compared with EMG patterns obtained from indwelling electrodes. The relative timing and general pattern of activation was similar between the two methods. Differences were observed at onset and cessation of activity with a more gradual change in activation observed from the surface electrodes. This could be due, in part, to the larger sampling area of the surface electrodes and a greater number of motor units being picked-up when compared to the fine wire preparation. Recordings from the ST indicate the possibility of cross-talk in the surface electrodes, however the BF and SM showed apparently similar patterns during the cycling task. In general, the EMG collected in this study suggest that the activation patterns obtained using surface electrodes provide a sufficient and reliable means to assess muscle involvement during the cycling task. A more detailed analysis may require use of a fine wire preparation.

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EXTRINSIC FEEDBACK AND THE LEARNING OF CYCLING KINETIC PATTERNS

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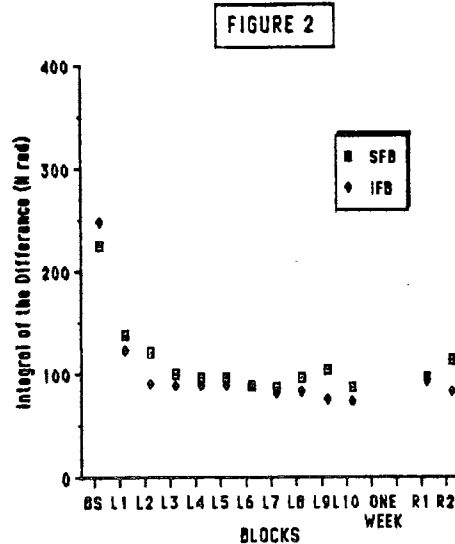
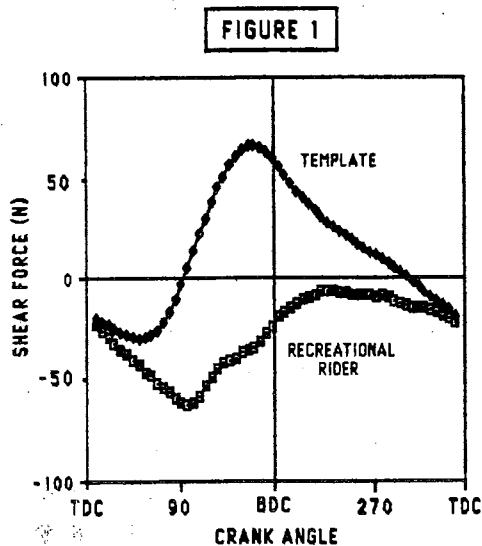
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Introduction - Researchers generally recognize that augmented extrinsic feedback enhances motor learning. The recently proposed **guidance hypothesis** (Salmoni et al., 1984; Schmidt et al., in press) suggests, however, that feedback given immediately after each response may interfere with learning by generating a response dependency on the feedback. Scheduling methods such as summarizing, delaying, or reducing the frequency of feedback are postulated to minimize the detrimental guiding aspects of feedback.

The effects of feedback scheduling and the **guidance hypothesis** have traditionally been studied using discrete tasks. Recent advances in on-line data acquisition permit the use of concurrent feedback in continuous task learning environments (McLean and LaFortune, 1988). Although the benefits of concurrent feedback during continuous tasks may seem obvious, the effects of feedback and its scheduling on the learning of such tasks remain unclear. The purpose of this study was to evaluate the retention of cycling kinetic patterns using two different feedback schedules, and to assess the general application of the **guidance hypothesis** to complex continuous tasks.

Methods - Eighteen inexperienced cyclists rode a racing bicycle mounted on a Velodyne trainer at 125 W (78 RPM). Pedal reaction forces were monitored by dual piezoelectric transducers mounted within the pedal. Subjects, assigned to an immediate (IFB) or summary (SFB) feedback group, performed 50 one-minute practice trials, receiving right pedal shear force feedback and a criterion pattern emphasizing "effective" shear near 0 and 180 degrees. IMF subjects received concurrent feedback 140 msec after the completion of every other pedalling cycle. SFB subjects received averaged feedback between trials. All subjects performed 10 retention trials without feedback one week later. Shear forces during eight to ten consecutive pedalling cycles were collected during the final 20 seconds of each trial and averaged for error analysis. Seven error measures described pattern deviation from the criterion (integral of the difference, magnitude (3), timing (2), and composite).

Results - The criterion shear force pattern and the average baseline pattern produced by the test subjects are shown in Figure 1. Practice and retention data describing the integral of the difference for both feedback groups is shown in Figure 2. Both groups improved significantly on all seven error measures during practice. Dramatic improvements occurred early in practice; thereafter, gains were slight. Performance decay in retention was negligible for both groups. Although group differences were not significant, the IMF group tended to perform better on all seven error measures during both practice and retention.



Discussion - High IFB group proficiency in retention indicates that the detrimental aspects of immediate feedback, defined in this task as concurrent feedback delivered after every other pedal cycle, were not apparent. Some subjects in the IFB group displayed large cycle to cycle pattern variations early in practice (e.g., the first two trials). Post-test comments from the subjects indicated that one-to-one associations between response execution (one pedal cycle) and the subsequent feedback describing that particular cycle were very difficult, if not impossible, to form. Consequently, it is suggested that IFB subjects may have packaged several pedal cycles and several feedbacks into an informational unit defining the relationship between movement and feedback. Such packaging may underlie the insensitivity of continuous tasks to the guiding properties of immediate feedback.

High SFB group proficiency in retention suggests that large changes in cycling kinetic patterning are achievable with relatively few feedback presentations. In fact, SFB subjects attained performance levels in retention equivalent to IFB subjects despite receiving 15 to 18 times less feedback.

These data suggest that biomechanical feedback, administered during or after a cycling task, can be used effectively to modify existing pedalling kinetic patterns, and that practiced patterns are retained despite the subsequent removal of feedback.

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