# REAL-TIME MOTION ANALYSIS BY MEANS OF PASSIVE MARKERS

G. Baroni, G. Ferrigno
Dipartimento di Bioingegneria, Politecnico di Milano, Milan, Italy
Centro di Bioingegneria, Fnd. Pro Juventute Don Gnocchi, Politecnico di Milano, Milan, Italy

## INTRODUCTION

A method for real-time motion analysis based on passive markers is presented. The ELITE motion analyser[1] has been used as hardware platform and the features of the interfacing between the two levels of the system architecture have been exploited, in order to implement true real-time acquisition, processing and representation of two-dimensional and three-dimensional kinematics data. System and algorithm performances, in terms of maximal number of processed markers at the highest system sampling rate have been evaluated and are reported.

## METHOD

The developed method exploits features of the interfacing between motion analyser (MA) and PC, which is implemented on the Keithley Metrabyte Digital Interface Card PDMA-32, plugged into the PC ISA expansion slots. The digital data transfer is performed through a DMA (Direct Memory Access) process, which fills a 64 Kb RAM buffer. Philosophy of real time operation consists of performing the desired CPU elaboration (centroid calculation, 3D reconstruction, graphic representation) on the raw data belonging to a frame already transferred to memory, while the background DMA process writes in memory the information relative to the successive frames. For this aim, after having dynamically allocated the DMA buffer, in order to preserve any previously loaded TSR program, two pointers are defined at the base of the allocated memory segment. Every time a frame has been transferred to memory, the interface card generates an hardware interrupt, responding to a square-wave signal sent by MA. At this event, the control is transferred to an interrupt function, which updates the first pointer (Pa) at the beginning of the next frame. The second pointer (Pb) is then exploited to read and elaborate the current frame. After the processing, Pb reaches pointer Pa and the process starts again, only when Pa is updated by the hardware interrupt, notifying that the new frame has been completely transferred to the DMA buffer. The algorithm guarantees in any case the complete execution of the necessary elaboration on a frame, updating the start of the successive processing at the beginning of the most recent acquired frame. In this way, real-time characteristic is always maintained, even engaging the CPU in any desired processing, with the only consequence of the eventual loss of intermediate frames, i.e. a decrease of the effective acquisition sampling rate.

## EXPERIMENT AND RESULTS

Performance evaluation of the implemented real-time algorithm has consisted of acquiring a markers grid in different set-up and CPU (Intel 80486 DX2/66) work conditions, increasing the marker number on the scene till a decrease in acquisition sampling rate emerged. Final marker number seen by each active TV camera has been considered an indicator for the algorithm performance at the investigated condition. The actual real-time sampling rate has been monitored, exploiting two software counters. The first one was incremented by the hardware interrupt routine at each end-of-frame and printed on the screen every 100 frames (nominal sample rate as 100 Hz); the second counter was incremented at the end of the CPU elaboration loop, printed out with the first counter and reset to zero every second. The value of this second counter equal to 100, testified the elaboration and representation of every acquired frame; on the contrary a value less than 100 indicated a CPU delay with respect to the DMA transfer process and represented the actual sampling rate of the real-time acquisition. System performance have been evaluated progressively increasing the CPU work condition. The heaviest investigated situation was characterised by 4 active TV cameras, 3D graphic representation on 3 planes and contemporary data saving on HD. Up to 12 markers could be seen by each TV camera maintaining the nominal system sample rate of 100 Hz.

## DISCUSSION

The developed real-time method exploits intrinsic accuracy and flexibility of the considered hardware platform for the implementation of visual, digital and analogic real-time feedback for subjects, operators and control systems. A high optimisation of the most critical software steps, such as centroid calculation, 3D reconstruction and optional parallel data saving on HD, has allowed to obtain results, which seem to be suitable for most of the imaginable application of real-time motion analysis by means of passive markers

## REFERENCES

1. Ferrigno, G., Borghese N.A. and Pedotti, A. (1990) Pattern recognition in 3-D automatic human motion analysis. ISPRS Journal of Photogrammetry. Remote Sensing 45, pp. 227-246.

# INFLUENCE OF NUMBER OF MARKERS PER BODY SEGMENT ON KNEE AND ANKLE KINEMATICS - AN IN-VIVO STUDY

W. Barth[1], K. J. Deluzio[2], P. A. Costigan[2], U.P. Wyss[1]
[1] Sulzer Orthopedics Inc., Winterthur, Switzerland
[2] Clinical Mechanics Group, Queens University, Kingston, Ontario, Canada

## INTRODUCTION

When determining body segment position and orientation with the use of skin markers, the marker configuration changes throughout the gait cycle due to system noise and skin movement which invalidates the underlying rigid body model. Recently, several methods have been developed to obtain an optimal fit between the observed marker coordinates and the underlying model. Based on theoretical studies simulating system noise only (1)-(3), it is generally suggested to use redundant control points to improve accuracy. The purpose of this study was to investigate the influence of the number of markers on the kinematics of the ankle and knee joint in an experimental setup.

## METHODS

Gait trials were performed on 6 normal healthy subjects. Data was recorded at 50Hz during normal walking using the OPTOTRACK motion tracking system. Clusters of 5 markers were attached to foot, shank and thigh. One marker was on a 10 cm probe pointing anteriorly on shank and thigh. Kinematics were resolved as flex/extension, ab/adduction and int/ext rotation at the knee and as rotations about the upper and lower ankle axis of the ankle joint. Segment orientation was calculated using the least squares method described in (2). The analysis was repeated 3 times using 5 (5set), 4 (4set) and 3 (3set) markers per body segment. For the 3set, the analysis was repeated a second time without using the least-squares technique assuming error free data (3set/ef). Marker clusters were characterized by the condition number presented in (4). The results of the 5set were assumed to be the standard. Maximum difference (bias) and the RMS value (precision) of the differences between the 5set and the other sets were calculated. Analysis of variance techniques were used to find statistical differences between the marker sets for each of the 5 joint angles.

## RESULTS

The best (lowest) condition numbers were obtained with the 5sets. The highest condition numbers were obtained with the 4set on thigh and shank and with the 3set on the foot. Table 1 summarises the results for the maximum difference between the 5 marker set and each of the others over the entire gait cycle. Aside from three cases indicated (*) in Table 1 there were significant differences between the 5set and the other marker sets in calculating the joint angles. Precision as indicated by the RMS of the difference showed similar results. For the 3sets the RMS was at 1.5° for the ankle angles and 0.5° for the knee angles. The 4set which did not use the probe, had RMS values of 1.0° for both ankle and knee angles.

Table 1. Maximum difference in degrees (std dev)

| Marker Set | Lower Ankle | Upper Ankle | Flex/extension | Ab/Adduction | Int/ext rot. |
|---|---|---|---|---|---|
| 3set | 3.5 (1.3) | 2.1 (0.9) | 1.1 (0.1) | * 0.8 (0.4) | * 0.3 (0.3) |
| 3set/ef | 3.5 (1.2) | 2.1 (0.8) | 1.7 (0.3) | 1.1 (0.5) | 0.9 (0.3) |
| 4set | 5.5 (0.7) | 2.7 (0.2) | 1.3 (0.3) | 1.0 (1.1) | * 0.6 (1.6) |

* not significantly different from zero ($\alpha$=.05)

## DISCUSSION

The analysis of the maximum difference between the 5set and the other marker sets is conservative as an estimate for possible bias. Results indicate that the use of the probe is warrented due to improvements in condition number, precision and bias. Using redundant markers seems to have a greater effect at the ankle than at the knee. When using three markers, the least squares technique showed no difference from the error free method.

## REFERENCES

(1) Challis JH., Med Eng Phys 17(2):83-90.(1995)
(2) Challis JH., J Biomech 28(6):733-7.(1995)
(3) Ball KA, Pierrynowski MR., J Biomech 29(9):1213-7.(1996)
(4) Soderkvist I, Wedin PA., J Biomech 26(12):1473-7.(1993)

# A GENERAL KINEMATIC REPRESENTATION FOR HUMAN MULTI-JOINT SYSTEMS

Dale O Anderson, Louisiana Tech University, Ruston, Louisiana, USA

Anne M Hollister, Louisiana State University Medical Center, Shreveport, Louisiana, USA

Karin Hollerbach, Lawrence Livermore National Laboratory, Livermore, California, USA

## INTRODUCTION

Human limbs can be modeled as multi-link systems with the bones connected by linkages representing the joint mechanisms (Giurintano et al., 1995, Delp and Loan, 1995). This approach is useful for representation of the work spaces of the joints and limbs and is essential for mechanical analysis such as statics and dynamics. Robot and machine kinematics have standard modular kinematic algorithms, such as the Denavit-Hartenberg (DH) representation (Denavit and Hartenberg, 1955), which can be used to calculate the movements and moments of such multi-linked mechanisms. The DH representation allows the incorporation of a sequence of joint mechanisms such as revolutes, universal joints, and ball joints in a linked mechanism. The DH representation is limited to mechanisms in which the mechanisms are orthogonal. The authors propose a more general algorithm that allows representation of non-orthogonal linkages like those found in human limbs: the algorithm is a variation of the Sheth-Uicker representation (1971) for kinematic chains connected by lower pair joints. There is a local coordinate system for each bone, and the location of the joint mechanism is linked to that coordinate system. For joints with more than one degree of freedom, each component of the joint mechanism is linked sequentially, and the final joint mechanism is linked to the coordinate system of the next bone in the sequence. The link geometry transformations are separated from the joint actuation transformations because they are the most complicated, and are constant.

## METHODS & RESULTS

Two coordinate systems are defined in each link--one (XYZ) at the input bone and the other (UVW) at the output bone. The joint motion is then described by the relationship of the input (XYZ) coordinate system of the i+1st link with respect to the output (UVW) coordinate system of the ith link. On link i in Figure 1, the distances $a_i$, $b_i$, and $c_i$ are the linear offsets from the input origin to the output origin along the $X_i$, $Y_i$, and $Z_i$ axes respectively. The angle $\alpha_i$ is the twist angle between the $Z_i$ and $W_i$ axes about the $X_i$ axis. The angle $\beta_i$ is the cant angle between the $X_i$ and $U_i$ axes about $V_i$ axis. The joint between the two links is formed by making the $Z_{i+1}$ axis and the $W_i$ axis coincident. Two examples of the system are demonstrated: the joints of the ankle and the wrist.

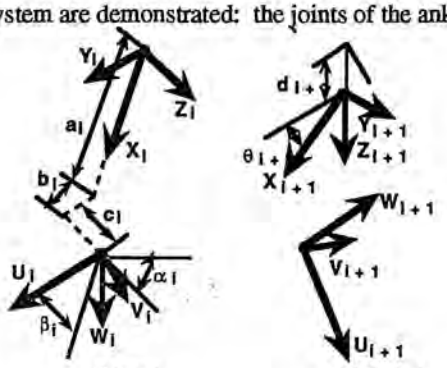



Figure 1. A representation of the ith and i+1st links in a kinematic chain.

## DISCUSSION

A variation of the Sheth-Uicker representation for kinematic chains connected by lower pair joints was developed. The new representation separates the link geometry transformations from the joint actuation transformations and is suitable for automated analysis of kinematic chains, because it requires that only the joint actuation matrices be recomputed for each new position. The new representation is applicable to robotic mechanisms as well as biokinematic systems.

## REFERENCES

Delp SL and Loan JP: *Comp in Bio Med*, **25(1)**:21-34, 1995.
Denavit J and Hartenberg RS: *J Appl Mech*, 215-221, 1955.
Giurintano DG, Hollister AM: *Med Eng Phys*, **17**:297-303, 1995.
Sheth PN and Uicker JJ, Jr.: *J Engr for Industry*, **93(1)**:102-112, 1971.

# ERRORS IN APPLYING THE SCREW DEVIATION METHOD TO SINGLE DEGREE OF FREEDOM MECHANISMS

K Hollerbach, Lawrence Livermore National Laboratory, Livermore, California, USA
AM Hollister, LSU Medical Center, Shreveport, Louisiana, USA
RL Van Vorhis, Lawrence Livermore National Laboratory, Livermore, California, USA

## INTRODUCTION

Instant helical axes or screw deviation axes (SDA) (Kinzell et al., 1972, Woltring et al 1985) have been used to describe the movements of bones relative to each other. An SDA is a translation along and a rotation about an axis of rotation and is calculated for each interval of joint motion recorded. Human joints often have two or more rotational degrees of freedom (DOF). Joint motion has been postulated to occur about evolutes, where the axis translates in the plane of motion as the joint moves. These mechanisms are quite different than a screw.

## METHODS

The authors generated motion paths for a body moving about a revolute, a revolute that translated in the motion plane as a function of rotation (evolute), and for a motion path about 2 revolutes. Data from these paths were then used to calculate screw axes, and the results were compared to the known axis' positions and orientations. The x, y, z trajectories of 4 non-coplanar points about each mechanism were calculated. SDA analyses were performed at 10 degree intervals because SDA accuracy is affected by step size. Screw deviation axes' positions, orientations, rotational angles, and translations were calculated according to Kinzell et al. (1972).

## RESULTS

The screw deviation method calculated correct results for pure revolutes and for prismatic joint mechanisms. Similarly, where translation occurred along the axis of rotation in true screws, the algorithm calculated the joint mechanism with virtually no errors.

An evolute is a single DOF mechanism with rotation about a revolute, and translation of the revolute in the motion plane as a function of degree of rotation. Screw deviation calculations for this case do not describe the location of the axis of rotation. The position of the calculated screw axis in the z=0 plane varies dramatically from the evolute's actual position. However, the joint angle and axis orientation results are as accurate as for a simple revolute.

A single DOF mechanism can occur with rotation about two revolutes in which rotation about the second is dependent upon rotation about the first. In this case, discrepancies occur in both the SDA rotations and displacements.

## DISCUSSION

SDA's should only be applied to joints whose mechanisms are a simple screw, a revolute, or linear mechanism. In all other situations, the method cannot measure the movements of the mechanism, nor can it represent the motion of the moving segments. Single evolutes and multi-degree of freedom systems cannot be accurately represented by the screw deviation method. The authors conclude that the kinematic mechanism must be known prior to application of the screw deviation method and that application of the method can produce reasonable results only in cases in which the joint mechanism is a screw or a revolute.

## REFERENCES

Kinzell GL, Hall AS, and Hillberry BM (1972) Measurement of the total motion between two body segments -- I. analytical development. *J Biomech* **5**, 93-105.

Woltring HJ; Huskies R, deLange A and Veldpaus FE (1985) Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. *J Biomech* **18**, 379-399.

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# AN INEXPENSIVE METHOD OF MEASURING ONE, TWO OR THREE DIMENSIONAL DISPLACEMENT OF THE HUMAN BODY DURING A RANGE OF HUMAN MOVEMENT STUDIES

Philip J. Rowe

Department of Physiotherapy, Queen Margaret College, Edinburgh, UK.

## INTRODUCTION

The ability to measure the linear displacement of key anatomical landmarks in one, two or three dimensions is a fundamental attribute of many motion analysis systems. In this way the scientist is able to describe human motion. Most motion analysis systems which record three dimensional displacements of the body contain a series of transducers (usually cameras) designed to determine the location of a point or series of points in three dimensional space. These opto-electronic systems are often expensive and complex and beyond the scope of many researchers.

An alternative method for measuring linear displacement was developed by Crosbie & Eisenhuth (1993) and has been used by Crosbie, Durward & Rowe (1996) to measure upper and lower trunk translations associated with stepping. This paper reports the continued development of the method to produce a system for the measurement of displacements of the human body with widespread applications in human movement studies.

## METHODS

The new system uses a series of rotary shaft encoders connected to light weight plastic wheels and inextensible tensioned strings to measure the displacement of key anatomical points in space. Movement of the string produces rotation of the wheel and hence the shaft encoder and this gives rises to an asynchronous digital signal which can be fed directly into a computer using a standard digital data acquisition board. The moment of inertia of the wheel and shaft encoder is small and therefore the tension required in the string is minimal (0.5 N). The encoder produces 500 pulses per revolution and when used in conjunction with a 54 mm wheel this gives a precision of less than 0.3 of a mm. Using an IBM 486 computer it is possible to monitor and record the output of up to 6 devices at speeds up to 5 m/s. Repeat calibration of the device over 6 metres on 5 occasions showed high accuracy and repeatability with a mean residual error of 6.5 mm and a maximum residual error in all the tests of 19 mm.

## RESULTS

Measurement of linear motion is simple to record with a single encoder. A series of studies have been carried out to record the motion of the sacrum during normal gait. The data obtained shows good agreement with gait parameters published in the literature. Velocities and accelerations calculated from the displacement data are also in agreement with published data. The method has proved repeatable and reliable both within and between days.

Planar motion has been studied using two encoders mounted at separate locations but connected to the same anatomical landmark. By recording the length of the strings between the two encoders and the landmark and using trigonometry it is possible to determine the location of the landmark in two dimensional space. This has been carried out for a variety of anatomical landmarks during a number of functional activities including sitting balance, standing balance, rising to stand and sitting down and the data was again in agreement with knowledge of these activities published in the literature.

The addition of a third transducer to form a triangle with a transducer at each corner makes it possible to record three dimensional motion and this has been investigated during reaching and pointing tasks.

## DISCUSSION

The system has proved capable of measuring displacements in one, two and three dimensions in a variety of situations. These investigations were relatively easy to design and implement using the encoder system. Initially there was concern about slippage of the measuring thread on the transducer wheel. However this does not appear to be a problem provided sufficient tension is maintained on the thread. A tension of 0.5 Newton has been sufficient to eliminate slippage in these experiments. The maximum force on the anatomical landmark was therefore 1.5 Newtons during three dimensional analysis and this did not appear to interfere with the motion being studied. It is proposed that, provided the constraints of the system are considered, this method of measurement forms a simple, inexpensive, portable, accurate and flexible method of recording human motion. The system has the potential to measure a variety of functional human movements and is therefore a valuable addition to the armamentum of the motion scientist and the clinical researcher

## REFERENCES

Crosbie J. & Eisenhuth J. (1993) Transducer for the measurement of linear displacement of body segments. Medical and Biological Engineering and Computing, 31, 430 - 432.

Crosbie J., Durward B. & Rowe P. (1996) Upper and lower trunk translations associated with stepping, Gait and Posture, 4, 26 -33.

# ON EXPRESSION OF HUMAN BODY MOTION

Nishimura, Tetsu  Sakata, Takuya  Wada, Yuhei  Ito, Masaru
Department of Mechanical Engineering, College of Science and Technology, Nihon University

## INTRODUCTION

In order to study a human body motion, a human body is sometimes replaced by a mechanical links system composed of human body elements. The each link is supposed to be connected by a spherical joint, if we neglect a small deformation or distortion at a human joint. This replacement is effective for a study of a rough human body motion, without tremendous calculations on a vector analysis. A complex human body motion is performed by the superposition of the relative motion on many human joints. It is supposed that each relative motion is comparatively simple. Hence, in this paper, the method of resolving a relative motion at a human joint from the actual human motion is mentioned.

## MODELING OF HUMAN BODY MOTION

At first we study a motion of a serial connected links system such as a motion of shoulder and arm. The position vector of the shoulder is denoted by $\mathbf{X}$. The element vectors of the upper arm and the forearm are denoted by $\mathbf{r}_1$ and $\mathbf{r}_2$, respectively. The top of the second vector can be represented by the equation, $\mathbf{R}_2=\mathbf{X}+\mathbf{r}_1+\mathbf{r}_2$. Here we introduce the local coordinate system $\mathbf{H}_1$ and $\mathbf{H}_2$, z-axis of which are embedded to the vectors $\mathbf{r}_1$ and $\mathbf{r}_2$, respectively. The position vector $\mathbf{R}_2$ can be rewritten as follow,

$$\mathbf{R}_2=\mathbf{X}+L_1\mathbf{H}_1\mathbf{i}_z+L_2\mathbf{H}_2\mathbf{i}_z=\mathbf{X}+\mathbf{H}_0\{\mathbf{H}_0^{\mathrm{T}}\mathbf{H}_1(L_1\,\mathbf{E}+L_2\mathbf{H}_1^{\mathrm{T}}\mathbf{H}_2)\}\mathbf{i}_z\,,$$

where $L_1$ and $L_2$ are the length of each element vector, and $\mathbf{H}_0$ is the local coordinate embedded on the shoulder. The components of the unit vector $\mathbf{i}_z$ are zero except z-component. And $\mathbf{E}$ is a unit matrix. In the above equation, $\mathbf{H}_0^{\mathrm{T}}\mathbf{H}_1$ represents the relative coordinate which is observed on $\mathbf{H}_0$. Here we distinguish the initial state from the current state, by adding the subscript '0' to each vector and local coordinate system, For example, $\mathbf{r}_{10}$ and $\mathbf{H}_{10}$ explain the initial state of $\mathbf{r}_1$ and $\mathbf{H}_1$, respectively. That is, the first subscript indicates the element number, and the second indicates the initial state.

Now we discuss the movement of the j-th element to the current state generally. As the relative vector $\mathbf{r}_{j0}$ moves into $\mathbf{r}_j$, the embedded local coordinate system is also rotated from $\mathbf{H}_{j0}$ to $\mathbf{H}_j$. The current coordinate system $\mathbf{H}_j$ can be derived by the following equation.

$$\mathbf{H}_j=\mathbf{H}_{j-1}\mathbf{H}_{(j-1)0}^{\mathrm{T}}\mathbf{H}_{j0}\,\Phi_j\,\Theta_j\,\Phi_j^{\mathrm{T}}\,\Omega_j.$$

Where $\mathbf{H}_{j-1}$ is the coordinate system embedded to the element vector $\mathbf{r}_{j-1}$, to which the vector $\mathbf{r}_j$ connects. And the rotation matrix $\Phi_j$, $\Theta_j$ and $\Omega_j$ are defined by the following equations.

$$\Phi_j=\begin{bmatrix}1 & 0 & 0\\ 0 & \cos\phi_j & -\sin\phi_j\\ 0 & \sin\phi_j & \cos\phi_j\end{bmatrix}\quad \Theta_j=\begin{bmatrix}\cos\theta_j & -\sin\theta_j & 0\\ \sin\theta_j & \cos\theta_j & 0\\ 0 & 0 & 1\end{bmatrix}\quad \Omega_j=\begin{bmatrix}\cos\omega_j & -\sin\omega_j & 0\\ \sin\omega_j & \cos\omega_j & 0\\ 0 & 0 & 1\end{bmatrix}$$

where $\theta_j$ is the relative swinging angle of $\mathbf{r}_j$ to $\mathbf{r}_{j-1}$. This swinging motion is performed about the axis, which is positioned on the x-y plain and rotated by $\phi_j$ from x axis of the local coordinate $\mathbf{H}_j$. And the relative rotation $\omega_j$ designates the relative rotation of vector $\mathbf{r}_j$ about z-axis. That is, $\theta_j$ and $\omega_j$ explain the relative motion of the body element $\mathbf{r}_j$ on the joint.

## LIMIT OF MOTION

Each joint has a limit motion. As we discuss the motion of arm, the maximum movable range on a shoulder joint and an elbow are shown in Table 1.

Table 1. Movable Range on a Joint

| Joint | Degree of Freedom | $\theta$ (deg.) | $\phi$ (deg.) | $\omega$ (deg.) |
|---|---|---|---|---|
| Shoulder | 3 | 0～180 | 30～270 | -90～90 |
| Elbow | 1 | 0～160 | 90 | 0 |

## RESOLVING METHOD OF RELATIVE MOTION

An actual human motion is observed by such as a high speed video camera, and obtained image data are transformed into digital data. Therefore, we know the actual position of each nodal point. From these nodal points data, we can derive the relative rotation angle at joints by the following equation.

$$\mathbf{V}=(\mathbf{H}_{10}^{\mathrm{T}}\mathbf{H}_{00}\mathbf{H}_0^{\mathrm{T}}\mathbf{r}_1\mathbf{r}_{10}^{\mathrm{T}}\mathbf{H}_{10}\mathbf{i}_z)/L_1^2$$

As the components of the above vector $\mathbf{V}$ is denoted by$(v_x, v_y, v_z)$, the angular motion are

$$\theta_1=\cos^{-1}v_x,\quad \phi_1=\cos^{-1}\left(-v_y/\sin\theta_1\right)\ \text{and}\ \ \theta_2=\cos^{-1}(\mathbf{r}_1^{\mathrm{T}}\mathbf{r}_2\ /\ L_1L_2).$$

Finally, angular rotation $\omega_1$ is derived by comparing the relative vector $\mathbf{r}_2$ with $\mathbf{r}_2^*$ which is observed on the local coordinate $\mathbf{H}_1\,\Omega_1^{\mathrm{T}}$. The calculated angular rotations can be developed by numerical formulations for analyzing a human body motion.

# INFLUENCES OF INERTIAL PROPERTIES ON JOINT RESULTANTS

Adamantios Arampatzis, Jie Gao, Gert-Peter Brüggemann
Institute of Athletics and Gymnastics, German Sport University Cologne, Germany

## INTRODUCTION

Many significant studies towards a biomechanical understanding of human movement have been conducted in different areas of sport activities [1], [2], [3]. Mathematical modeling and inverse dynamics are often used for analysing mechanical factors of functional injuries in musculoskeletal systems. In inverse dynamic analysis, joint reaction forces and moments may vary with different sets of inertial properties even using the same mathematical models. The aim of this study is to explore how different inertial properties influence the results of joint resultants at different joints of the lower extremities during dynamic jumping movements.

## METHODS

The mathematical model consists of three rigid segments: foot, shank and thigh. The hip joint was modeled with 6 degrees of freedom (DOF), while the knee and ankle joints were assumed to possess 3 DOF for rotations. The relationships between the kinematics and dynamics of the 12 DOF model were established using the software program ALASKA for conducting 3D analysis of movements.

Data collection was carried out under laboratory conditions. Two cameras were used to capture the kinematic data (240 Hz), while a kistler force platform was used to measure the ground reaction forces (960 Hz).

Nine elite athletes volunteered to take part in the experiment and performed 5 high speed movements: (1) Counter-Movement-Jump (CMJ); (2) Drop-Jump (DJ) from a height of 40 cm; (3) Hurdle-Jump (HJ) with a hurdle height of 1 m; (4) Jump-Running (JR); (5) One-Leg-Jump (OLJ). All those movements were described as quick perfomance because the ground contact times were limited to 150 - 170 ms.

The resultant joint forces and moments at the different joints were calculated using inverse dynamics methods. In order to compare the numerical results, three sets of inertial properties were used in our model: the inertial characteristics of Zatsiorsky, the inertial parameters calculated by the Hanavan model and Zero inertial values. In the latter method it is assumed that all inertial parameters are equal to zero.

## RESULTS

The numerical results showed that using the inertial parameters of Zatsiorsky and Hanavan had only negligible effects on the estimation of resultant forces and moments at the ankle joint. The values of joint forces and moments at this level had a variation of only ±0.5%. At higher joints the influences of joint resultants increased. At the knee joint they varied by up to ±5%, and at the hip joint ±8%.

To estimate the contribution of the inertial forces to the total joint resultants for such a quick movement, the joint forces and moments using the inertial characteristics of Zatsiorsky were compared with those using the Zero inertial parameters. Both curves showed the same tendency in the time history. The difference at the ankle joint level remained within a range of ±3%, while the inertial part of joint resultants at the knee joint took up to ±20-40% and at the hip joint even to ±60-70%.

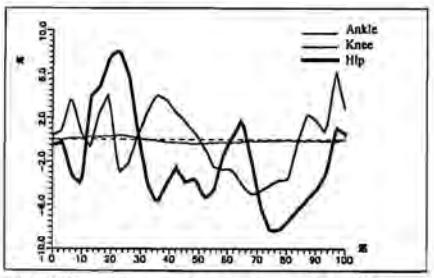



Fig1: Differences of joint resultant moments (in %) (ZATSIORSKY - HANAVAN)

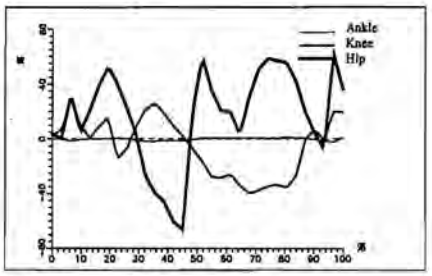



Fig2: Differences of joint resultant moments (in %) (ZATSIORSKY - ZERO Inertial Values)

## CONCLUSION

From our results it is reasonable to predict that the inverse dynamic analysis at the ankle joint level is not significantly influenced by chosing different inertial parameters, while the inertial forces have to be taken into account at the knee joint and hip joint for quick movements. However, the results of chosing inertial parameters of Zatsiorsky and Hanavan do not make any great differences even at the hip joint.

## LITERATURE

[1] Bobbert, M.F., van Ingen Schenau, G.J. (1988): Coordination in vertical jumping. J. Biomechanics 21, 3, 249-262.

[2] Pandy, M.G., Zajac, F.E. (1991): Optimal muscular coordination strategies for jumping, J. Biomechanics 24, 1, 1-10.

[3] van Ingen Schenau, G.J., Bobbert, M.F., Huijing, P.A., Woittiez, R.D. (1985): The instantaneous torque-angular velocity relation in plantar flexion during jumping. Med. Sci. Sports Exercise, 17, 4, 422-426.

## THE "AXIAL ROTATION RHYTHM" OF UPPER EXTREMITY

Wei-Ning Chang, *Fong-Chin Su, *Hong-Wen Wu, *Yi-Wen Chang, Yi-Chao Lu, Chi-Ying Wong
Kaohsiung Veterans General Hospital, Kaohsiung, Taiwan, R.O.C.
*National Cheng Kung University, Tainan, Taiwan, R.O.C.

### INTRODUCTION

The forearm axially rotates about 68-71 degrees of pronation and 74-88 degrees of supination, with a total rotation range of 120-160 degrees. The humerus rotates about 90 degrees both in internal and external directions (depends on the degree of arm elevation), with a total arc about 180 degrees. However, the contribution of the scapulothoracic, glenohumeral, and forearm motion has never been discussed together. Their dynamic relationship, and the role of the scapula in arm axial rotation is yet unclear. The purpose of this study is to investigate the composite motion of the scapula, the humerus, and the forearm in axial direction with percutaneous intracortical pinning on scapula and two special orthoses on the humerus and the forearm.

### METHODS

We developed a 3-demensional biomechanical model of the upper extremity using combined skin and skeleton-based marker sets. In the model, the trunk and the arm are treated as a 4-segment linkage system consisted of thorax, scapula, humerus, and forearm, each attached with 3 retroreflective markers. The subject did the axial rotation motion with the arm in 5 different positions. They are (1)resting position, (2)abduction 90 degrees with the arm perpendicular to trunk, (3)maximal abduction, (4)forward flexion 90 degrees with the arm perpendicular to trunk, and (5) backward extension to the maximal extent. The images of the markers were collected via six video cameras and digitized by Expert Vision motion analysis system (Motion Analysis Corp., Santa Rosa, CA, USA) during the test. These data were transferred to a Sun workstation, then downloaded to a personal computer for further calculation using an analysis program coded in MATLAB language. The calculation was based on the concept of vector projection on the plane perpendicular to the rotational axis.

### RESULTS

The rotational motion was decomposed to represent composite motions of each of the 3 segments (scapula, humerus, forearm: Figure 1). We found that the initiation and the progression pattern of each segment differed greatly with the arm in different positions. For example, the humerus rotated more than the forearm during external rotation with the arm in backward extension while the forearm rotated more in other arm positions. The composite maximal axial rotation angles of each segment were plotted in bar chart as illustrated in Figure 2. The maximal overall axial rotation angles of the arm varied greatly with the arm in different position. (257 ° --maximal abduction; 356 ° --90 ° abduction). The forearm usually contributed about 140 ° during rotation, so the differences mainly came from the scapula and the humerus. The scapula axially rotated 66 ° with the arm in backward extension while only rotated 19 ° with the arm in 90 ° forward flexion. The glenohumeral joint (humerus) rotated maximally (143 °) with the arm in 90 ° abduction, while minimally (69 °) with the arm in maximal abduction.

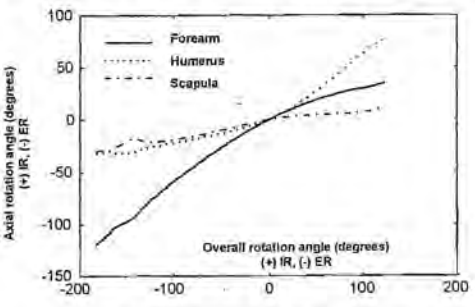



Fig 1: Axial rotation with arm in 90° abduction

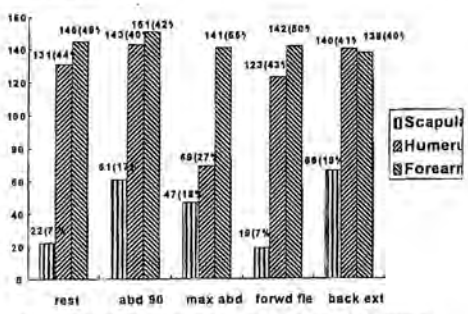



Fig 2: Maximal axial rotation angle of scapula, humerus, and forearm with the arm in different positions

### CONCLUSION

The motion pattern, the relationship, and the contribution of each segment of the upper extremity during axial rotation of the arm have never been reported in the literature. By using the skeletal marker of the scapula and the video-based motion analysis system, we determined the axial rotation rhythm of the upper extremity including the scapula, the humerus, and the forearm in one subject. However, since the information was obtained from a single subject, it is not representative. Further investigation in more subjects is necessary to draw more definitive conclusions.

# FEEDFORWARD AND FEEDBACK CONTROL OF CENTER OF GRAVITY DURING LATERAL ARM RAISING IN MAN

S.Vernazza[1], M.Cincera[23], A.Pedotti[23], J.Massion[1]
1 - Laboratoy of Neurobiology and Movements, CNRS, Marseille, FRANCE
2 - Dept. of Bioengineering, Politecnico di Milano, Milano, ITALY
3 - Centro di Bioingegneria, Fondazione Pro Juventute 'Don C.Gnocchi', Milano, ITALY

## INTRODUCTION

In motor tasks performed in gravity conditions, balance is guaranteed under static condition as long as Center of Gravity (CG) projection onto the ground is maintained inside the supporting surface.

While for many natural movement such as trunk movement [1] the maintainance of equilibrium requires postural adjustments by other segments not directly involved in the voluntary movement, some specific motor acts, like arm raising, do not require, from a mechanical point of view, postural adaptation, given the low mass of the segments in movement. The question then arises if a mechanism of postural control is present regardless of the mechanical constraint [2]. If this is the case, it is of interest the analysis of the organization between voluntary movement and postural adjustments.

## METHODS

Nine male healthy subjects (age range, 25 to 65) participated to the study. In response to a sound, each subject was instructed to raise the right arm in the frontal plane to the horizontal. Each subject performed a total of 40 trials in a randomize order in two standing condition (bipedal and monopedal standing) and two load condition (with no load and with a 3.5kg load at the level of the raising hand).

The movement kinematics were recorded using the ELITE system (100 Hz). Actual CG displacement ($\Delta CGa$) and theoretical CG displacement ($\Delta CGt$, that is CG displacement solely due to the biomechanical effect of the raising of the arm) were computed in the frontal plane. The following index of CG compensation was defined (as a function of time): $((\Delta CGt-\Delta CGr)/\Delta CGt)*100$. Index close to 0 and to 1 means respectively absence and presence of CG compensation.

## RESULTS

IC was first evaluated at the end of voluntary movement execution. 6 out of 9 subjects demonstrated significant CG compensation (on average, IC=54%) in bipodal standing without load; 8 out of 9 subjects demonstrated CG compensation (on average, IC=51%) in the remaining experimental conditions. IC was on average higher when considering monopodal standing with respect to bipodal standing, while the presence of load did not change significantly the amount of compensation.

When considering IC as a function of time (computed from 100ms after the start of the movement to 1000ms after the end of the movement), it was observed that CG compensation was significantly different from zero along the whole interval. Nevertheless, a common behaviour of the index was observed regardless of the experimental condition: IC was higher right after movement initiation (IC=69%), presented a minimum right after the end of the movement (IC=46%) and increased again at the end of the motor act (IC=57%).

## DISCUSSION

Results showed that during movement an active mechanism of control of CG postion is present, which is responsable for a minimization of CG displacement along the movement execution, even if this mechanism is not required from a strict mechanical point of view. Moreover, the 'gain' of the control mechanism depends on the support condition, since the CG compensation is higher in monopodal standing with respect to bipodal standing.

The dynamic behaviour of IC showed the possible presence of two different control mechanism [3]: a 'dynamic CG control', acting in feedforward manner, which stabilize CG position during voluntary movement execution and a 'static CG control' acting in feedback manner which stabilize the final CG position and which take place only when voluntary movement is accomplished. The time relationship between these two control systems might depend on the support and load condition.

## REFERENCES

[1] Alexandrov A., Frolov A., Massion J., 1994. Voluntary forward bending movement in human: a principal component analysis of axial synergies. In: Vestibular and neural fronts. K.Tagushi, M.Igarashi, S.Mori (Eds.). Elsevier Science B.V. ICS 1070: Amsterdam, pg.345-348.

[2] Massion J (1992) Movement, Posture and Equilibrium: Interaction and Coordination. Progress in Neurob, Vol.38: 35-56.

[3] Bouisset S, Zattara M, 1987. Biomechanical study of the programming of anticipatory postural adjustments associated with voluntary movement. Journal of Biomechanics, vol.20, pg.735-742.

# MEASUREMENT OF HUMAN WRIST FUNCTIONAL AXES MOTIONS: PRELIMINARY VALIDATION OF METHOD

R L Van Vorhis
Interdisciplinary Ergonomics Research Program
Lawrence Livermore National Laboratory
Livermore, California

A M Hollister
Department of Orthopaedic Surgery
Louisiana State University Medical Center
Shreveport, Louisiana

K Hollerbach
Institute for Scientific Computing Research
Lawrence Livermore National Laboratory
Livermore, California

P Tittiranonda
Interdisciplinary Ergonomics Research Program
Lawrence Livermore National Laboratory
Livermore, California

S R Burastero
Interdisciplinary Ergonomics Research Program
Lawrence Livermore National Laboratory
Livermore, California

## INTRODUCTION

Characterization of wrist motion is important to the study of hand function. Extremes of wrist posture has been identified as a risk factor for work related musculoskeletal disorders [1]. Representation of wrist motion in inappropriately assigned coordinates results in joint motion errors [2]. Functional axes for joint movement have been identified for many anatomic joints [cf. ankle[3] and knee[4] etc.). Youm et al [5] has characterized wrist joint function and identified two offset revolutes relating to flexion-extension (FE) and radial-ulnar (RU) movements. Pronation-Supination (PS) wrist motion can also be ascribed to rotation about a third revolute in the forearm. Wrist motion described relative to two anatomic revolutes is sufficient for describing the spatial nature of arbitrary wrist carpal-radial motion or posture. The authors describe validation of an experimental method for measuring wrist motion in which the locations of the FE and RU axes are located in each subject and the motion is described relative to these functional axes.

## METHODS

A jig was fabricated with reference to bony prominences which repeatably positioned the hand/forearm in a neutral position. Once in the jig, the locations of the wrist functional axes could be established in a jig-based coordinate system using anatomic relations [5]. Arrays of three 7mm retroreflective markers [spaced _1.5cm] were attached to the carpus of the hand, on the forearm over the distal radius and on the jig. From this, the location of the functional axes of the wrist was established relative to hand and forearm marker triads.

To validate the measurement method, it is necessary to be able to successfully measure motion of the wrist while it is constrained to undergo exclusive FE or RU motion. Motion about these axes are the two degrees-of freedom of this anatomic system. Marker position was simultaneously recorded using both the MacReflex system and a graphical method. Position of the retroreflective markers is established using a MacReflex system sampling at 50 Hz and processed in LabView on a Macintosh computer. Motion of the wrist was constrained to each axis by interfacing the carpus to a plane oriented normal to each axis. Paper placed on the plane was marked for each hand position facilitating accurate independent angle measurement.

## RESULTS

Motion about each unconstrained axis calculated from markers measured with the MacReflex system correlated well with axis rotation calculated using the graphical method. In addition, the corresponding constrained motion remained nearly constant.

| FE motion with RU constraint | | | RU motion with FE constraint | | |
|---|---|---|---|---|---|
| | correlation | $r^2$ | | correlation | $r^2$ |
| FE | 0.953 | 0.999 | RU | 0.889 | 0.992 |
| RU | -0.084 | 0.102 | FE | -0.068 | 0.325 |
| | | n = 948 | | | n = 900 |

Table 1. MacReflex vs. Graphical angle correlation.

## DISCUSSION

The results demonstrate that the method is valid for establishing independent motion of the FE and RU wrist axes. Investigation of the effects of axes location variation and improvements in axes location identification is planned.

## REFERENCES

[1] Silverstein B, et al, Occupational factors and carpal tunnel syndrome, *Am. J. Ind. Med*, 1987, 11:3, p343-58.

[2] Hollerbach K, et al, Mathematical analysis of errors resulting from choice of reference frame coordinates in measuring human joint motion, 2nd Triennial Int'l Hand Wrist Biomech. Symp. 10 Sep 1995, S.F., CA.

[3] Inman, VT *The Joints of the Ankle* MacMillian Philadelphia 1976.

[4] Hollister, A, et al, The axes of rotation of the knee. *Clin Orthop* 290:259-268, May, 1993

[5] Youm Y, et al, Kinematics of the wrist. I. An experimental study of radial-ulnar deviation and flexion-extension. *J. Bone and Joint Surg.* Am. V., 1978 Jun, 60(4):423-31.

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# AN EXPERIMENTAL INVESTIGATION OF THE UPPER ARM AXIAL MOTION RANGE

Xuguang Wang*, Frédéric Mazet‡, Nilo De Castro Maia‡, Karine Voinot‡, Jean Pierre Verriest* and Michel Fayet‡
*LESCO/INRETS, Bron, France; ‡GMC/INSA, Villeurbanne, France; Email: wang@inrets.fr

## Introduction

The use of computerized human models in a computer-aided design (CAD) environment allows for the ergonomic evaluation of a product at a very early stage in design. Through simulating human movements and postures, human models are used to make vision, reach and fit analyses. In order to predict natural looking simulation, joint limits must be taken into account. However, most data of joint limits reported in the anatomical literature (e.g. Kapandji, 1994) are described in a qualitative and conventionalized terminology: flexion, extension, abduction, adduction, etc. For the shoulder complex, Engin and Chen (1986) were the first to give a three-dimensional mathematical modelling of the human shoulder complex based on a statistical in-vivo data base. In their work, only the shoulder joint sinus, which is defined as the total motion range of the upper arm circumduction, has been investigated. The upper arm axial rotation was supposed to be independent of the position of the upper arm in the shoulder joint sinus cone, and its motion range has not been reported. In the present work, the shoulder complex, which refers to the combination of the glenohumeral joint and the shoulder girdle including the clavicle and scapula and their articulations, is simplified as a 'lumped' spherical joint between thorax and humerus. The purpose of the work is to extend the Engin and Chen's database by quantifying the upper arm axial motion range.

## Materials and method

Six voluntary male subjects of ages 23 through 34, who have neither musculoskeletal abnormalities nor any history of trauma in the shoulder complex region, were chosen. With help of a cardan support system, the upper arm could be easily positioned in any position within the shoulder joint sinus cone. Eight latitudes at an interval of 20 degrees were used to specify the upper arm elevation. For each latitude level, a different number of longitudes, which varied from one when the upper arm is vertical to a maximum of twelve when it is horizontal. For an elbow position, each subject was asked firstly to make the maximal internal rotation along the longitudinal axis of the upper arm, and then to make the maximal external rotation along the same axis. Using a sonic digitizer, the following surface landmarks were measured for these two limit arm postures in order to calculate their corresponding attitudes:
- three torso related surface landmarks which are seventh cervical vertebra (C7), sternoclavicular joint (incisura jugularis (IJ)) and bottom of the sternum (processus xiphoideus PX), which allowed the torso-fixed coordinate system to be defined,
- lateral epicondyle (LE) and medial epicondyle (ME) which gave the elbow joint center (EL),
- two markers at the wrist to locate its center (WR).

The shoulder joint center was approximated by the center of the sphere which fits the most the measured elbow positions. From the measured positions of these surface landmarks, three Euler angles, which represent the upper arm attitude with respect to the torso in terms of latitude, longitude and axial rotation (twist), were calculated. Inverse distance weighting interpolation was firstly applied to the raw data of both maximal internal and external upper arm axial rotation angles so as to provide estimates at the node of a regular upper arm latitude-longitude grid, which is based on the Engin and Chen's shoulder sinus. Then a global polynomial method was used to smooth interpolated data over the regular grid.

## Results

A statistical data base of the upper arm axial motion range has been established using a similar approach as that proposed by Engin and Chen (1986). Both maximum internal and external upper arm axial rotation surfaces were fitted into a third order polynomial function based on the experimental data from six male subjects. The sample mean and the confidence limits for the population mean were derived under the hypothesis of normal distribution. The present study shows that the axial motion range of the upper arm depends strongly on the position of the upper arm in the shoulder sinus cone and varies on average from 112 degrees to 166 degrees.

## References

Engin A.E. and Chen S.M.(1986), Statistical data base for the biomechanical properties of the human shoulder complex - I: Kinematics of the shoulder complex, Journal of Biomechanical Engineering, August 1986, Vol.108, pp.215-221.

Kapandji, I.A.(1994), Physiologie articulaire, Tome 1: Member supérieur, 5th Edition, MALOINE.

## MORPHOLOGICAL DATA ACQUISITION USING MRI/CT: GEOMETRIC TRANSFORMATION OF A HUMAN SHOULDER MODEL

B.L Kaptein, F.C.T. van der Helm
Dept. of Mechanical Engineering, Delft University of Technology, The Netherlands.

INTRODUCTION In our lab., a 3D biomechanical model of the human shoulder, with a representation of all muscles and bones, has been developed. This model is used to analyse the kinematic and dynamic behaviour of the shoulder in order to improve the diagnosis, treatment, and prevention, of shoulder disorders [1]. The geometric parameters for this model like muscle size, location of muscle attachments, bony contours and location of joint rotation centres have been obtained by means of a cadaver study [2]. As a result of this, the model is based on the geometry of a specific individual human being. To get a better agreement between the simulated movements of the model, and the measured movements of a living person, it is necessary to base the model on the geometry of that person. This involves that the geometric structures of a living individual human being are to be measured. This can be done by creating 3D geometric models of the bones of the subject from MRI or CT images. These models transform the generic shoulder model to an individual model. This transformation involves rotation, scaling, translation as well as a deformation. The question that has to be answered is: Is a transformation based on the shape of the bones sufficient to transform all the geometric parameters of the shoulder model. In this study we describe the procedure used to check the similarity of muscle attachment locations between different subjects after a transformation based on bone geometry.

METHODS In the past, a cadaver study has been performed in order to acquire data for inertia, geometry and muscle contraction parameters. Muscle attachments were measured on the shoulder bones of these cadavers, with respect to a local coordinate system defined by five metal markers (little screws in the bone).

CT images of the same bones have been made at Leiden University Hospital. Using a semi automatic contour detection method [3], contours of these bones are extracted, and connected by triangular elements to form a surface model. Using the position of the metal markers in the CT images, the muscle attachment contours can be mapped on these bone models.

For this study, we used three sets of these bone models; M1, M2, and M3. We transformed M1 to M2, M1 to M3, and M2 to M3, using an elastic matching procedure[4]. This procedure transforms the points of the source model to the surface of the target model. This results in exact agreement between the surfaces of the source and target model.

After transformation the similarity between the muscle attachment contours of the reference model and the transformed models can be checked. For this, muscle attachments are modelled as polygons consisting of 1000 points. The difference measure used for the similarity check consists of three parts: Dmin, the minimum of the distances from the points of the reference contour with the transformed contour and vice versa; Dmean, the mean of these distances; Dmax, the maximum of these distances. We use these three values to be able to distinguish between different types of errors: First, muscle lines that have a shift, or length difference in their direction. These have a Dmax equal to this shift, a small Dmean, and a small Dmin. Second, lines that have a shift perpendicular to their direction. These have a Dmax, Dmean, and Dmin equal to this shift. Third lines that have different directions. These have a large Dmax, smaller Dmean, and a very small Dmin. Of course, combinations of these types are also possible. Because the original measured data are available, we have compared model simulations based on measured data with simulations based on a transformed model. Results of these simulations will be presented at the conference. Muscle attachments of humerus, clavicle, scapula, and thorax have been processed.

RESULTS Difference in muscle attachment contours of different subjects after a geometric transformation based on bone models. In general, the muscle attachments that were very difficult to measure also show larger differences than the easy to measure attachments. For six line shaped muscles on the scapula of the three subjects, Dmin is around 1 mm, and Dmean around 6 mm. From this we can conclude that the method gives a good prediction of muscle attachment contours. Dmax differs for the different pairs of subjects, from 9 mm to 18 mm. This difference is caused by muscle attachment contours having a length difference in their direction. For the plane shaped muscle attachments, the results are more or less the same as for line shaped attachments.

DISCUSSION Looking at the results for the line shaped muscle attachment contours on the scapula bone, we can conclude that the main source of these differences is probably the inter individual difference, in combination with the errors in the assessment of muscle attachments during data acquisition. Using the proposed method, we can make a reasonably good prediction of the location and shape of muscle attachments. The effect of the differences on the simulation results of the shoulder model look very promising because differences in moment arm of the muscles are smaller than the differences in attachment. The only drawback of this method is that the difference in muscle architecture between different subjects is ignored

REFERENCES [1] Van der Helm FCT (1994) J. Biomechanics 27(5), 551-569. [2] Van der Helm FCT et all. (1992) J. Biomechanics 25, 129-144. [3] Kaptein BL et all. (1996) EMBS Conf., Track 4.4.1-2. Amsterdam. [4] Kaptein BL et all. Exact geometric matching of 3D triangular surface models. Submitted to CVPR Conference Puerto Rico (1997).

# SHOULDER INJURIES ON SWINGING MOTION OF ARM

S. Yanagi, T. Nishimura, N. Watanabe, Y. Wada, M. Itoh, H. Yamashita, K. Tsurumi
Dept. Mech. Eng., Col. Sci. & Tech., Nihon University, Tokyo, Japan

## INTRODUCTION

Many baseball and volleyball players often suffer from shoulder injuries. The injuries mainly the dislocation of the long head tendon. Because the shoulder joint is the one of the unstable joint, the muscle and tendon support the joint not to dislocate. Generally the muscle and tendon have enough strength against the longitudinal force, however, they are weak against the lateral force. It is imagined that the lateral force could cause the shoulder injuries.

In this study, we analyze the actual volleyball spiking motion with assuming the arm as the two mechanical links system, the first link is the upper arm and the second is the forearm. We calculate the driving force at the shoulder joint and resolve the force into two direction, longitudinal and lateral direction. Here the longitudinal direction is the direction from shoulder to throat because the injuring tendon is attached on the upper side of the scapula. Finally, we discuss the possibility that the lateral force will cause the shoulder injuries.

## METHOD

The actual volleyball spiking motion are taken by two high speed video cameras (1/200[sec]) for getting the three dimensional data. From the obtained video image, we get the displacement of the link element. However, it is impossible to obtain the displacement without error. So, we get the velocity and acceleration by the Fourier series.

The relative position vector of the shoulder, elbow and hand are defined as $\mathbf{R}_0$, $\mathbf{R}_1$ and $\mathbf{R}_2$, respectively. The driving force, $\mathbf{F}_0$, at the shoulder joint can be obtained by

$$\mathbf{F}_0 = (\ddot{\mathbf{R}}_0 + \lambda_{G1}\ddot{\mathbf{R}}_1)m_1 + (\ddot{\mathbf{R}}_0 + \ddot{\mathbf{R}}_1 + \lambda_{G2}\ddot{\mathbf{R}}_2)m_2 \qquad (1)$$

where, $\lambda_G$ and, $m$, means the ratio of the position of the mass center to the link and mass of the link (1:first link, 2:second link), respectively.

Since the injuring tendon is located on the upper side of the scapular, let us consider to resolve the driving force given by Eq.(1) into the longitudinal (from shoulder to throat ) and the orthogonal plane to the longitudinal direction, that is the lateral force.

## RESULT

In order to reduce the influence of the error included in the displacement, we adopt the Fourier series for the velocity which is derived by the first divided difference. Here, it is very important to decide the adopted numbers of the Fourier series. For the displacement and velocity, we adopt the least number of the Fourier term which can express the obtained data well. And for the acceleration, in order to avoid amplifying the higher term, over 4 terms are cut.

By substituting the obtained acceleration into the Eq.(1), the driving force can be derived. The calculated lateral force is show in Fig.1 with the actual spiking motion. The doted circle indicates the position of the long head tendon. Since the calculated lateral force is generated by the muscle contraction, the reaction force is loaded on the opposite direction of the lateral force. From fig.1, the lateral force on fourth phase pushes the tendon strongly. It is the shearing force to the tendon, about 230 [N]

## DISCUSSION

The lateral force on the shoulder joint can be derived by the our suggested procedure. With considering that the only muscle and tendon on the shoulder joint make the lateral movement restrict, there is a high possibility that the lateral force cause the shoulder injuries.

Our final goal is to find effective form without the shoulder injuries. For that, it is very important to examine with support of medial knowledge whether our suggested form will have the possibility to cause the injuries. The cooperation of the mechanical and medical filed must be necessary to prevent the spots injuries.

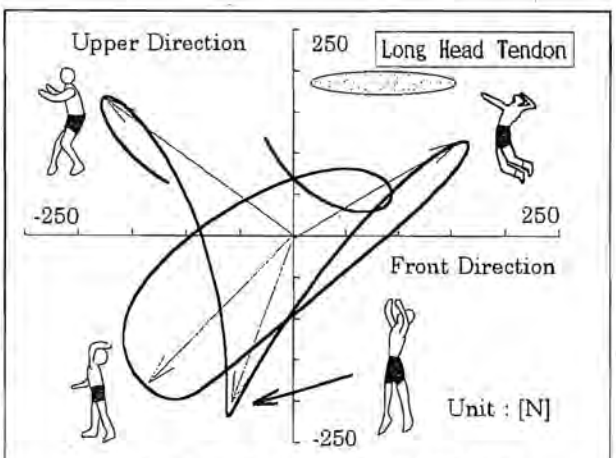


Fig.1 Lateral Force on the Shoulder Joint

# BIOMECHANICAL CRITERIA FOR DETERMINATION OF SITTING POSTURE

Akinari Hirao and Nobutoshi Yamazaki

Department of Biomedical Engineering, Faculty of Science and Technology, Keio University

Yokohama, JAPAN

## INTRODUCTION

Individual differences of sitting posture were observed in the angle and position of pelvis, and lumbar curve. The skeletal posture seems to be determined by internal loads caused by muscles and elastic elements. The aim of this study was to discuss about these inherent criteria for sitting posture. Synthesis of sitting based on the criteria would bring us reasonable chair design. Goosens and Snijders (1995) mentioned the criteria as the reduction of shear forces acted on the body surface. In this study, we compared internal loads at the comfortable sitting with the enforced sitting using a musculo-skeletal model (Hirao and Yamazaki 1996). Using the results, we clarify the proper criteria for determing the comfortable posture.

## METHODS

A musculo-skeletal model was constructed of 13 rigid segments and 63 muscles in a sagittal plane (Fig.1). The chair was modeled simply as external forces acted on the body segments. Muscle forces, intervertebral disc forces and intra-abdominal pressure were calculated from measured external forces and sitting posture. The experimental chair with adjustable cushion was used for measuring the external forces. Body landmarks were measured by arm type three dimensional digitizer (Kosaka Lab. VSC-27-A). Seat dimensions, angles and cushion hardness of the experimental chair were set on the conditions of a standard office chair (Kotobuki FA-1151). Following five kinds of sitting postures corresponding to VDT works were measured; comfortable posture, ischial position enforced moving forward/backward 5 cm from comfortable one, foot position enforced moving forward/backward 15 cm. Fifteen healthy normal male subjects (height:165~185cm, weight:45~80kg) were examined.

## RESULTS AND DISCUSSION

Individual standard deviation of stresses on all back muscles was minimized on comfortable posture (Fig.2). This tendency was observed on 9 out of 15 subjects with ischial positions and 9 subjects with foot positions. The result shows individual standard deviation of stresses on all back muscles is one of the most important factors for maintaining comfortable posture, in order to prevent local muscle fatigue. Sum of intervertebral disc shear forces was also minimized on comfortable posture (Fig.3). This tendency was observed on 5 out of 15 subjects with ischial positions and 10 subjects with foot positions. The shear forces seem to be another important factor in order to keeping spinal curve. Forces of abdominal muscles, neck muscles, leg muscles and intra-abdominal pressure did not show clear tendencies.

## CONCLUSION

Individual standard deviation of stresses on all back muscles and shear forces on intervertebral discs were minimized on comfortable posture. These were reasonable candidates of biomechanical criteria for determination of sitting posture.

## REFERENCES

Hirao, A. and Yamazaki, N. (1996) Estimation of internal loads on sitting posture by musculo-skeletal model, Book of abstracts of 10th conference of the European Society of Biomechanics, 345.

Goosens, R.H.M. and Snijders, C. J. (1995) Design criteria for the reduction of shear forces in beds and seats, Journal of Biomechanics, 28, 225-230.

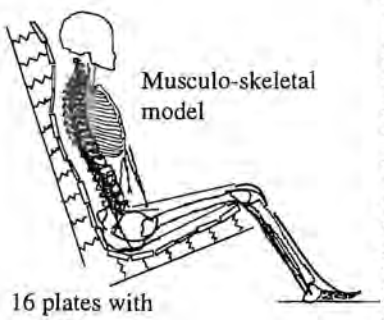



Fig.1 Simulation model

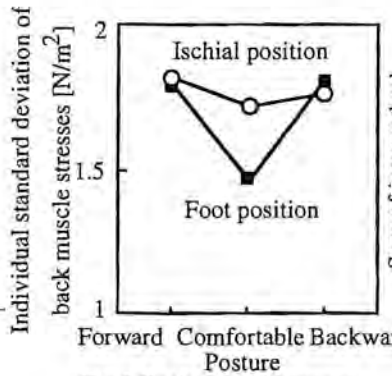



Fig.2 Back muscle stresses

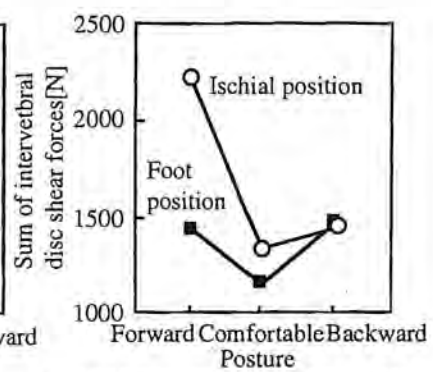



Fig.3 Intervertebral disc shear forces

# CLOSED-LOOP BALANCE CONTROL IN STANDING POSTURE

Paul Bourassa, François Prince
Université de Sherbrooke, Sherbrooke, Qué,Canada

## Introduction

Winter, Prince and Patla [2] have proposed that balanced control in the standing posture may be achieved using ground pressure variables as feedback . Newton and Neale[1] ,after investigating posture with 17 parameters measured over 24 body landmarks concluded on the need for addressing the posture problem with a dynamic model. Also, Winter, Prince, Stergiou and Powell [2] have shown that medio-lateral control of posture with feet side-by-side is achieved by the hip abd/adductor. This paper proposes a dynamic model to study the influence of level and centre of mass feedback on posture.

## Model

The frontal plane posture model is a multi-body four bar system for which the Kane dynamic equations of motion were solved numerically using the Autolev Software. Outputs are analyzed and animations are produced. Models with ground constraints having a single degree of freedom have advantages and drawbacks and are discussed. A model with elastic supports under each foot ends up with 5 degrees of freedom and has been used to investigate two types of internal muscular torque control, one whose amplitude is based on the perceived level or inclination of the horizon, $T = -k\, q_2\, b_3$, the other which is based on the ground projection of the center of mass. It is important to introduce ground damping in the horizontal and in the vertical direction in order to get a ripple free integration of the differential equations of motion for the achievement of smooth torque control.

## Simulations and analysis

A small initial angular velocity input of .1 rad/s of the left leg body was imposed to represent the effect of a lateral impact and to produce a momentary loss of balance. The first 1 dof constraint model is well suited for sway control through horizon level correction but it do not provide, per se, outputs for ground reactions unless auxiliary variables are introduced . In the 5 dof second model large values of ground stiffness are required in order to raise the corresponding natural frequencies well above the sway frequency. The ripple ground oscillations in turn impose the use of very small time steps for good convergence. Critical damping has thus been introduced to smooth out the variables and ease the control using these output variables. The amplitude of the torque constant was varied over a range of practical interest. Small changes in the torque constant resulted in important changes of the sway motion near the practical limit of sway stability. The sway variables do indeed exhibit an hyperbolic type of representation over the range of practical torque amplitude. The sway sensitivity to this parameter is therefore very significant near the stability boundary. Mathematical expressions have been devised for the work done by gravity during sway, initial kinetic energy and also for the work done by the hip muscles during a cycle. Based on energy conservation principles, maximum sway angles have been obtained and good agreement has been found. Simulations were carried for two types of control over a stable range of torque constants. Results for sway angles, foot reactions, c.o.p c.o.m coordinates and sway cycle period will be discussed and shown with animation.

## Discussion

One of the major difficulty associated with these models is in the identification of cases where either constraints or elastic restoring forces are producing negative ground reactions which of course lead to instability. The 5 dof model with elastic supports do provide all the proper information for the forces evaluations which are indeed proportional to the associated elastic deflections which are part of the state vector. On the other end, these elastic variables cause vibrations to appear and a ripple output ends up in all other variables, including control. If the control is operating on very small variables, the sign of the control input may be reversed. Proper damping and damping constants must thereof be introduced in order to eliminate or at least minimize these effects.

## Conclusion

The 5 d.o.f. model with critical ground contact damping gives a ripple-free smooth continuous motion simulation and provides an efficient mean for incorporating muscular feedback control based on horizon, center of mass, center of pressure and many other output variables. The control amplification constants must be chosen within a range limiting foot reactions to positive values and to realistic physical muscular forces. State variables such as leg and pelvis angles could be described by hyperbolic curves over that range. There appear to be a relationship between the feedback amplification constants based on horizon, com or cop control.

Acknowledgments

The authors are indebted to the NSERC and to the MRC MT-4343 for their financial support.

## References

[1] R.Newton, R.Neal "Three-dimensional quantification of human standing posture." Gait and Posture, V.2:205-212.1994.

[2] D.Winter,F.Prince,P.Stergiou,C.Powell."Medial-lateral and antero-posterior motor response associated with centre of pressure changes in quiet standing. Neuroscience Research Communication, 12, 1993, 141-148.

# MANIPULATION PRIMITIVES FOR 3D POSTURE SIMULATION WITH A GRAPHICAL MANIKIN

J.P. Verriest - INRETS/LESCO, 69 675 BRON Cedex - FRANCE - Email : jpv@inrets.fr

**Introduction** - The biomechanical analysis of posture of a worker or of an athlete requires the quantification of its body attitude. Optronic tracking devices are commonly used for this purpose in association with surface markers. This kind of method is not applicable if markers cannot be placed on the subject (in a competition for instance) and, in many situations, especially in a work environnment, objects can mask one or more markers during movements. An alternative method to determine the body attitude is to use multiple simultaneous calibrated video pictures and match a calculated human figure to the picture of the real subject. Such a human figure can be a geometrical human model projected onto the video picture with the same viewing parameters as in the real scene . This technique of matching can be interesting if the geometrical model is provided with a manipulation method enabling to build rapidly any posture from a standard one (i.e. standing, sitting, ..). The objective of this paper is to present such a manipulation method based on movement primitives.

**The geometrical model** - To represent the real subject, a manikin composed of a 55 rigid links skeleton and 1300 facets body contours enveloppe is used [Verriest et al., 1994a]. The links are grouped into functional segments organized in a tree-like chain whose root is the pelvis. These segments are : the torso, the head-neck assembly, the upper and the lower limbs. Each segment is provided with basic movements responding to geometrical constraints assigned to skeletal or contour points. In general, a segment has more than one joint and is kinematically redundant; i.e. the number of mobility axes is larger than the number of constraints. To overcome this problem, coupling is introduced between certain axes or an implicit optimization criterion of minimum angular velocity is applied to select a particular axis of motion when more than one are possible.

**Global strategy of manipulation** - Any manikin point Pi, either articular center or contour vertex, can be choosen to determine a movement. Constraint points Qi are the goals for the manikin movements. Building the desired posture is conducted step by step, from a starting posture by chaining segmental movements. In the first step, the entire body is moved as a whole. Then, each segment is moved until the manikin posture matches the subject's one. If the subject's dimensions are not known, before building the posture, the dimensions of the standard manikin are modified according to distances measured between subject surface points.

**Specification of segmental movements** - A minimum of three points are necessary to move the entire body. Preferably, torso points should be used, especially shoulder points and hip points to determine this global motion. The 3D translation and rotation resulting from the constraints are calculated (method of Veldpaus, 1988) and applied to the pelvis. Then, for the segments, one or more position constraints can be specified to determine the segment's orientation. The thorax is linked to the pelvis by three vertebral ball-and-socket joints located approximately at the level of T12/L1, L3/L4 and L5/S1. The angular displacement calculated from the constraints is equally distributed between the three joints. Since the position of the constrained points and the articular angular position are not linearly dependant the angular position is iteratively adjusted until the orientation constraints are satisfied. The case of the head-neck segment is similar to that of the torso segment except that there are only two joints considered : C7/T1 and C1/skull. Three basic movements are defined for the upper and lower limbs : flexion-extension, orientation and pivoting. Any attitude of the limbs can be reproduced with these basic movements depending on the number of constraints [Verriest et al.1994b].

The complete definition of a posture can be obtained with 14 constraints : 3 for the global body position and orientation, 2 for the torso/pelvis position, 1 for the head-neck assembly and 2 for each of the four limbs. The time needed to build the posture from a picture is the time to click the required points on the screen.

**References**

Veldpaus et al (1988)," A least-square algorithm for the equiform transformation from spatial marker coordinates", j. of biomechanics, vol. 21, no.1, pp 45-54

Verriest J.P., Wang X., Trasbot J., Tessier Y (1994a), "Application of a 3D human model to Computer aided ergonomic design of vehicles", in proc. of FISITA'94: "Engineering for Customers", pp53-61, Int. Academic Publishers, Beijing

Verriest, J.P., Rezgui M., Wang X. (1994b)," Experimental validation of arm reach movement simulation", in Proc.12th congress of Int. Ergonomics Ass., Toronto, pp342-344

# AN INTERACTIVE COMPUTER GRAPHICAL SYSTEM FOR THE STUDY OF UPPER EXTREMITY KINEMATICS USING UP TO 3 DEGREES OF FREEDOM AT EACH JOINT

Julie M. Pickard, William L. Buford, Jr., Karin W. Elder, Rita M. Patterson
Biomechanics Lab, Department of Orthopaedic Surgery and Rehabilitation
University of Texas Medical Branch, Galveston, Texas, U.S.A.

## INTRODUCTION

With advances in computer graphic workstations and graphics oriented software, kinematic model workstations and/or software have become outdated. Buford, et al. [1] developed an interactive kinematic simulation on Evans and Sutherland systems which are no longer available. Lin, et al. [4] developed a three dimensional kinematic simulation using PHIGS programming software. Since PHIGS (Programmer's Hierarchical Interactive Graphics System 1988 ISO Graphics Standard) is no longer the accepted standard for graphics software, this simulation is no longer portable.

Other biomechanics simulation packages are currently under development. The works of Delp, et at. [2] and Chao, et al. [3] are commercial systems which provide certain kinematic animation and anatomic visualization capability. Unlike the others, the UTMB simulation is a research tool with multi-degree of freedom joint axis adjustment capability written within a flexible programming environment.

## METHODS

Upon acquisition of a fresh (unembalmed) adult cadaver, the upper extremity was imaged using a G.E. Model 9800 Computerized Tomography scanner. Using the highest resolution (1mm thick slices) for the wrist, elbow, and shoulder joints, and a lower resolution (5mm thick slices) for remaining sections of the arm, 490 consecutive slices were produced. From these images, UTMB developed software was used to construct the three-dimensional bone structures, consisting of over 120,000 triangular polygons.

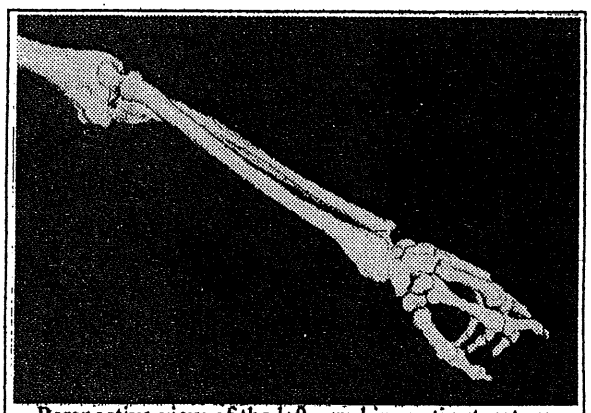
To begin software development, these three-dimensional polygonal images were transferred to a Sun UltraSparc 170 workstation with an Evans and Sutherland Freedom™ 3200 graphics accelerator. With the capability of processing 600,000 polygons per second, the Evans and Sutherland graphics accelerator allows for a real-time interactive simulation of the upper extremity. The program was written using C++, OpenGl, and Motif software within the UNIX X (Solaris) Window environment.

## RESULTS

The kinematic model allows for the arbitrary (6 degrees of freedom) adjustment of multiple axes of motion for each joint. The user interface provides scaling, translation, and rotation in any perspective. The figure is a photo of the complete upper extremity structure.

The present simulation provides for:

1. Interactive control and visualization of the upper extremity boney structure.
2. 3-D adjustment of the following axes:
   - A hinge axis for each of the distal and proximal interphalangeal joints of the fingers.
   - Two axes for each of the metacarpophalangeal joints of the fingers.
   - Three axes for the carpometacarpal joint of the thumb.
   - Three axes for the wrist and forearm.
   - One axis for the elbow joint.
   - Three axes for the shoulder joint.
3. Rotation of each segment about the previously defined axes under program control or using mouse/spaceball I/O.



Perspective view of the left arm kinematic structure (Gouraud shading) from a proximal/radial position.

## DISCUSSION

The flexibility required in UTMB Orthopaedic Biomechanics research dictates the development of a portable three-dimensional, computer graphic, kinematic simulation of the upper extremity. While this kinematic model is based upon the structure of a single cadaver specimen, variables are included in the simulation to allow for changes in geometry. By changing geometric transformations or data files, this system can be used to model different cases. Scaling of each bone segment, limits in joint range of motion, muscle size, muscle constraints, and other variables provide for the ability to model a given person based on radiographic or other clinical measurements.

## REFERENCES

1. Buford, W.L. and Thompson, D.E., *IEEE Trans. on Biomed. Eng.*, 34:6, 444-453, June, 1987.
2. Delp, S.L., Loan J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., and Rosen, J.M., *IEEE Trans. on Biomed. Eng.*, 37:8, 757-767, 1990.
3. Chao, E.Y.S., Lynch, J.D., Vanderploeg, M.J., *Trans. of the ASME*, 115, 562-568, November 1993.
4. Lin, Horng-Haur, Buford, Jr., W.L., Patterson, R.M., Elder, Karin, Proc., *XV Int. Soc. Biomech. Congress*, 84-85, June 1995.

## ACKNOWLEDGMENT

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# CENTER OF PRESSURE (COP) ENHANCEMENT IN PIEZOELECTRIC FORCE PLATES

R. Sommer, D. Kohler, C. Calame; Kistler Instrumente AG Winterthur
Winterthur, Switzerland

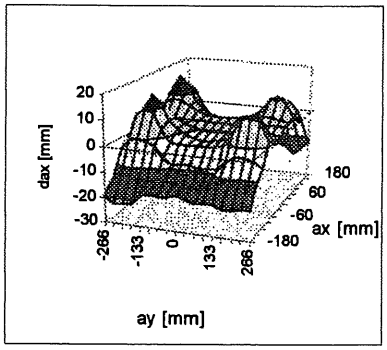



*Fig1: Typical COP error map*

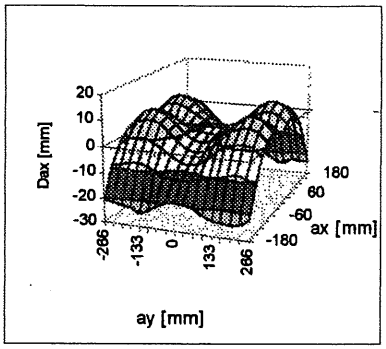



*Fig2: Polynomial error simulation*

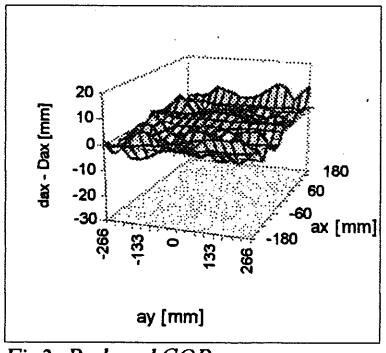



*Fig3: Reduced COP error*

## INTRODUCTION

The point of force application (COP, center of pressure) measured with force plates is an important parameter for the evaluation of biomechanical joint forces and moments together with motion analysis systems. The COP accuracy influences directly the accuracy of these evaluations [1,2].

COP errors are systematic and can be corrected for. This work describes an algorithm which significantly improves the COP accuracy.

## METHOD

COP error maps were established by applying forces on >200 points evenly distributed over the force plate surface.

For each point the difference between the true actual COP (ax, ay) and the measured COP ($ax_m$, $ay_m$) defines the COP error map (dax, day) *(Fig1)*. A symmetrical surface polynome of 4th order was developed to simulate these errors (Dax, Day) *(Fig2)*. The simulation is used for the compensation (*Fig3*).

$$Dax = ((P_{1x} \cdot ay_m^4 + P_{2x} \cdot ay_m^2 + P_{3x}) \cdot ax_m^3 + (P_{4x} \cdot ay_m^4 + P_{5x} \cdot ay_m^2 + P_{6x}) \cdot ax_m) \quad (1)$$

$$Day = ((P_{1y} \cdot ax_m^4 + P_{2y} \cdot ax_m^2 + P_{3y}) \cdot ay_m^3 + (P_{4y} \cdot ax_m^4 + P_{5y} \cdot ax_m^2 + P_{6y}) \cdot ay_m) \quad (2)$$

$$ay_{corrected} = ay_m - Day \quad (3)$$

$$ax_{corrected} = ax_m - Dax \quad (4)$$

where: $ax_m$, $ay_m$ = measured COP coordinates (5)

$P_{ix}$, $P_{iy}$ = coefficients (force plate model specific) (6)

The coefficients P were found by performing a polynomial interpolation (*Fig2*) on the measured COP errors (*Fig1*).

## RESULTS

The COP accuracy can be substantially improvement illustrated by the results for three different force plate models:

| Force plate model | 9281B11 | 9281C... | 9287B |
|---|---|---|---|
| Original COP error [mm] | 8.4 ± 4.4 | 11.6 ± 6.1 | 14.1 ± 9.1 |
| Improved COP error [mm] | 2.0 ± 0.7 | 3.2 ± 2.0 | 5.8 ± 3.7 |

To achieve these results, the force plate has to be mounted on a rigid base according to specifications. The coefficients P for different force plate models are available on request.

## DISCUSSION

The COP error was proven to be exactly proportional to the bending moments in the four „feet" of the force plate. In other words: force plate „feet" are either stiff (high frequency response, low COP accuracy) or they are relatively „soft" (high COP accuracy, low frequency response). With the polynomial correction, it is possible to achieve both, high frequency response and good COP accuracy on very stiff piezoelectric force plates. The algorithm can also be used on COP data acquired earlier. Input parameters are only the measured COP coordinates and force plate model specific coefficients.

## REFERENCES

[1] Bobbert M. F., Schamhardt H. C. (1990) Accuracy of Determining the Point of Force Application with Piezoelectric Force Plates, J. Biomechanics, Vol 23, No 7

[2] Katsumi Mita et. al. (1993) An Investigation of the Accuracy in Measuring the Body Center of Pressure in a Standing Posture with a Force Plate, Frontiers Med. Biol. Engng, Vol 5, No 3

# THE DETERMINATION OF TOTAL POWER DURING ON-WATER ROWING

V.V.Kleshnev
Research Institute of Physical Culture, Saint-Petersburg, Russia

## INTRODUCTION

The purpose of the study is to introduce correct way of calculation of power produced by a rower during on-water rowing. The power applied to a handgrip was defined in many studies for calculation of the rowing mechanical efficiency (1,2) as well as for evaluation of rowers' work capacity and skill. Usually, the handgrip power (Ph) was calculated as:

$$Ph\ (W) = \int Fh \cdot dVh \cdot dt \qquad (1)$$

, where Fh (N) - the handgrip force, Vh (m/s) - linear velocity of a handgrip. This equation is correct during rowing with stationary axis of oar rotation (in a pool, tank or ergometer), but it seems to be not correct during on-water rowing when the axis is mobile.

## METHODS

The measurements were done using a single scull, rowing pool and a computer-aided rowing simulator which allows to imitate as on-water rowing so and stationary rowing in a pool. The devices were equipped by gauges which was connected with computer though ADC. The patterns of each mechanical parameter during the stroke cycle were received (3). A group of male rowers (n = 16, 19.5±3.6 yr., 1.91±0.07 m, 89.7±6.9 kg, mean ±SD) volunteered to take part in the study. The subjects performed the test on the simulator consisted of three one minute trials with different stroke rate. The test was repeated twice on the simulator with the mobile rower's workplace position, that simulates on-water rowing and with the stationary workplace position, that simulates in-pool rowing. The four of the rowers performed the same test in the scull and in the rowing pool.

## RESULTS

While the hand grip power was calculated using the mentioned above way, the power applied to an oar blade (Pb) was calculated during rowing on the simulator as:

$$Pb\ (W) = \int Fb \cdot (Vb + Vs) \cdot dt \qquad (2)$$

, where Fb (N) - the force applied to the blade, Vb (m/s) - relative velocity of the blade, Vs (m/s) - relative velocity of the shell. The Fig.1 shows the differences in the patterns of the blade and handgrip power during tests on simulator that imitates on-water and pool rowing.

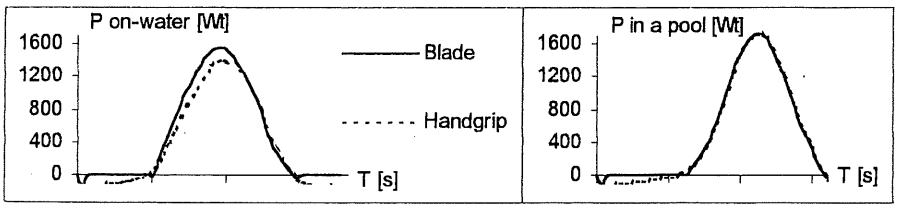



**Fig.1. Comparison of the blade and handgrip power patterns during on-water and in-pool rowing**

The average blade power was 11.2±0.4% higher than handgrip one (Table 1) during simulated on-water rowing. These two powers were practically equal during simulated in-pool rowing.

**Table 1. The average blade and handgrip power during simulated on-water and pool rowing**

| Power on: | Handgrip (W) | Blade (W) |
|---|---|---|
| On-water rowing | 421±38 | 467±45 |
| In-pool rowing | 441±42 | 443±41 |

The blade power was calculated during real on-water and in-pool rowing as:

$$Pb\ (W) = \int Fb \cdot (L \cdot (d\theta/dt) + Vs \cdot \cos\theta) \cdot dt \qquad (3)$$

, where L (m) - outboard length of the oar, θ - angle of the oar with perpendicular axis of boat shell. The difference between blade and handgrip powers was also determined during real on-water rowing, but it was smaller (7.1 - 8.5%).

## DISCUSSION

The reason of higher blade power could be explained by effect of the boat shell movement that increase the blade velocity. Therefore, the power applied to a handgrip represents only a part of the total power produced by a rower during on-water rowing. On contrary, during in-pool or ergometer rowing the power produced by a rower can be determined quite correct by the handgrip power.

In previous studies the maximal efficiency of in-pool rowing was determined as 27% (1) and on-water rowing as 24% (2). This difference could be quite explained by the shown above phenomenon.

## REFERENCES

1. Fukunaga T., Matsuo A., Yamamoto K., Asami T., Mechanical efficiency in rowing. European J. Appl.& Occupational Physiology: 1986: 55(5): 471-475
2. Nozaki D., Kawakami Y., Fukunaga T., Miyashita M., Mechanical efficiency of rowing a single scull; Scand.J.Med.Sci.Sport 1993: 3:251-255
3. Kleshnev V., The Application of Computer Technologies to Management of Special Training in Rhythmic Sports; Current Research in Sport Sciences: Plenum, 1996, 137-146

# THE 3D ARTICULAR EFFORTS IN ROWING

P. Pudlo, F. Barbier, J.C. Angué

Laboratory of Automatic and Industrial and Human Mechanics
URA CNRS 1775, University of Valenciennes. France

## INTRODUCTION :

The studies of rowing are numerous and diverse. The physiology of the rower and the mechanics and biomechanics of rowing are the preferred areas of study for researchers in this field. Such studies improve training and help in the selection of the crew. However, at present, no study appears to include articular efforts in training aids : computer data, which has often been used in the accurate study of gait. The purpose of this paper is to present the first results of the computation of articular efforts resulting from experiments carried out on a male rower competing at French regional level.

## METHODS :

The first stage consisted of developing an experimental device. The device is made up of a Model C Concept II ergometer completely instrumented (Pudlo. 1996a) and opto-electronic system SAGA3 (Pudlo. 1996b). The device allows the measurement of the 3D coordinates of the anatomic points, the measurement of the external efforts, as well as their application point. The second stage is the 3-dimensional modeling of the body : anthropometric characteristics (mass, position of the centre of gravity) come from (Dempster. 1955), inertial characteristics from (Hanavan. 1964). The computing code used is an adaptation of (Barbier. 1994).

## RESULTS :

Table 1 shows the maximal modulus of the articular forces, of the articular torques and the mean modulus of the articular torques for the wrist, the elbow, the shoulder, the hip, the knee and the ankle, on both right and left side. Figures 1 and 2 show the modulus of the articular torques for the upper limbs and lower limbs on the right side. These results are the mean of 3 strokes for a considerable stroke rate (40 strokes/mn).

| | wrist | elbow | shoulder | hip | knee | ankle |
|---|---|---|---|---|---|---|
| **Max. Modulus of the Force right side (N)** | 523.98 | 526.97 | 533.19 | 459.86 | 503.43 | 519.77 |
| **Max. Modulus of the Force left side (N)** | 453.99 | 456.27 | 463.45 | 572.39 | 595.38 | 611.32 |
| **Max. Modulus of the torque right side (N.m)** | 10.85 | 55.32 | 91.21 | 205.32 | 130.03 | 132.99 |
| **Max. Modulus of the torque left side (N.m)** | 8.85 | 36.49 | 102.24 | 276.11 | 143.00 | 163.43 |
| **Mean Modulus of the torque right side (N.m)** | 3.48 | 18.36 | 29.77 | 128.47 | 86.37 | 80.04 |
| **Mean Modulus of the torque left side (N.m)** | 2.61 | 13.96 | 32.25 | 163.92 | 90.93 | 93.86 |

Table 1 : Maximal modulus of the articular forces, maximal modulus of the articular torques, mean modulus of the articular torques right side and left side for : the wrist, the elbow, the shoulder, the hip, the knee, the ankle, 39.22 strokes/mn.

Modulus of the articular force, articular torque: $\sqrt{(Fx_i^2 + Fy_i^2 + Fz_i^2)}$, $\sqrt{(Cx_i^2 + Cy_i^2 + Cz_i^2)}$ for the i joint

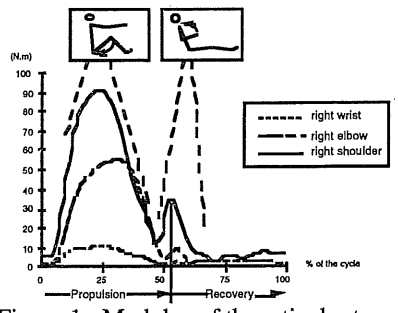



Figure 1 : Modulus of the articular torques of the upper limbs (39.22 strokes/mn)

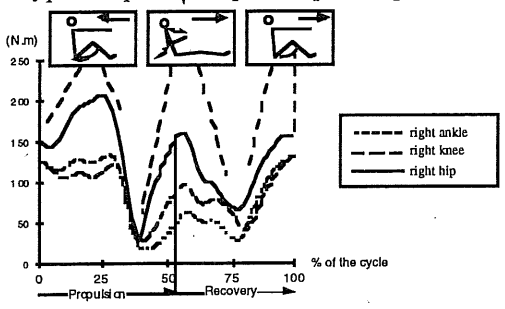



Figure 2 : Modulus of the articular torques of the lower limbs (39.22 strokes/mn)

## DISCUSSION :

The results show that: (1) for the upper limbs, the maximal modulus of the articular forces, the maximal modulus of the articular torques and the mean modulus of the articular torques are greatest at the level of the shoulders. (2) for the lower limbs, the modulus of the articular forces are greatest at the level of the ankles, (3) the mean modulus and the maximal modulus of the articular torques are greatest at the level of the hips, (4) a dissymetry between right side and left side. Moreover. the curves 1 and 2 emphasize the fact that : - 2 peaks for the upper limbs (the first corresponds to the push of the lower limbs, the second is the activation of the upper limbs). - 3 peaks for the lower limbs (the first corresponds to the push of the lower limbs during the propulsion, the second is the muscular activation needed to maintain the trunk and the upper limbs during the end of the propulsion, the third is the recovery).

## REFERENCES

Pudlo P., Barbier F., Angué J.C., (1996a). Instrumentation of the Concept II ergometer for optimization of the gesture. The Engineering of Sport. Editor : Steve Haake. ISBN: 90-5410-822-3. pp 137-140.

Pudlo P., Barbier F., Angué J.C., (1996b). Kinematic and Dynamic analysis of the rower's gesture on concept II ergometer. ISBS96. pp 317-320

# LOWER LIMB MUSCLE ACITIVITY PATTERNS IN FEMALES DURING MAXIMAL ERGOMATER ROWING: DIFFERENCES DUE TO LEVEL OF COMPETITION

R. Smith, Faculty of Health Sciences The University of Sydney, AUSTRALIA
P. Milburn, School of Physiotherapy, The University of Otago, NEW ZEALAND

## INTRODUCTION

During the traditional rowing race, rowers must sustain a high power output for a period of approximately six to eight minutes. This is a sufficiently long time for the efficiency of the movement to have a large bearing on the outcome of the competition. Investigation of the ways in which the athlete can achieve economies in the movement are important in the effort to improve performance. One method by which economies can be achieved is through the utilisation of two joint muscles. Two such muscles were investigated in this project to see if there were any differences in the deployment of these muscles between rowers performing at different competition levels: national and club.

## METHODS

Ten female rowers (club n=5, national n=5) were asked to row maximally for six minutes at a stroke rate of 29 strokes per minute. During each minute the position of the major joints on one side of the body, the vertical and horizontal forces and their point of application on the foot, were recorded at 60 Hz. Simultaneously the electromyograms of gluteus maximus, rectus femoris, vastus lateralis, and biceps femoris were sampled at 900 Hz. Equations of motion were derived for each segment to enable the calculation of hip, knee and ankle joint moments. Electromyograms were amplitude normalised to percent of maximal effort and time normalised and ensemble averaged for each group.

## RESULTS AND DISCUSSION

The evidence of the electromyograms (Figure 1) suggested that the national level rowers used the single joint muscles gluteus maximus and vastus lateralis nearer their maximum force than the two joint muscles, rectus femoris and biceps femoris. The situation was reversed in the case of the club level rowers. Power savings would occur for the hip and knee joints if the powers of the joint moments were of opposite sign (Prilutskii et al., 1992). This is the case for both levels of rowers from 22% to 35% of the stroke. However the club level rowers, by deploying the two joint muscles at greater tensions inhibit the transmission of power from the knee joint to the hip joint by keeping the knee joint at a high level of stiffness. This may be one mechanism whereby the national level rowers were able to achieve a higher level of mechanical effectiveness (1.32 J per joule of external work) compared with the club level rowers (1.44 $J/J_{extwk}$).

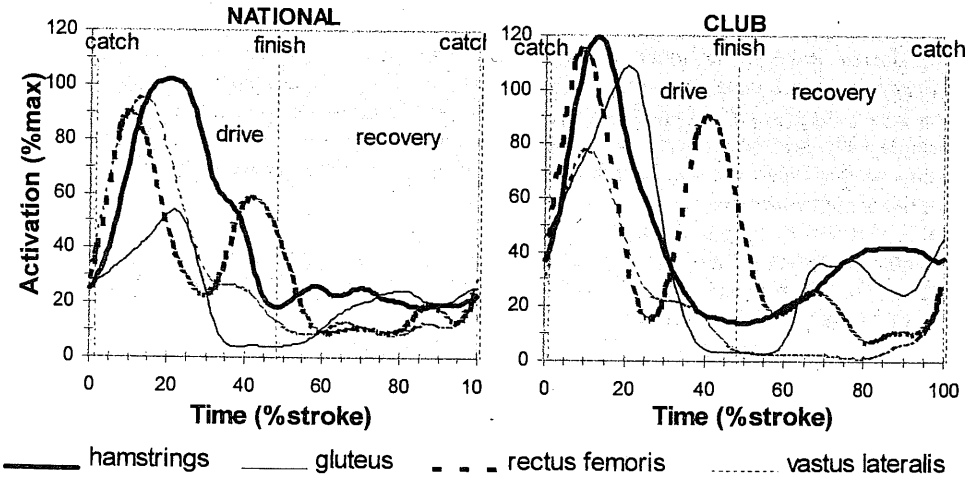



Figure 1 Ensemble average activation of the large muscle groups of the lower limb during maximal rowing for national and club level rowers

## REFERENCE

1. Prilutsky BI, Zatsiorsky VM, Petrova LN, (1992a) Mechanical energy expenditure on human movement and the anthropomorphic mechanism. Biofizika. **37,** (6), 1101-5.

# POWER PRODUCTION DURING SWIM STARTING

D.G.E. Robertson, V.L. Stewart
*School of Human Kinetics, University of Ottawa, Ottawa, ON, Canada*

## INTRODUCTION

The start is one important element of the sprint races in swimming. Although it occurs in less than a second from when the gun goes until the swimmer leaves the blocks, it affects the swimmer's flight from the blocks to the water and more importantly the entry and first few stokes in the water. Little research has been done to determine which muscle groups contribute to the starting thrust or how these muscles are sequenced. Some coaching manuals state that during a thrusting motion there is a sequential recruitment of the strongest muscle groups, followed by progressively weaker groups finishing with the weakest (Norman, 1975). Robertson and Fleming (1987), however, in their study of the vertical jump and the standing broad jump discovered that this was not the case. In light of this inconsistency, this study will determine the mechanical work patterns of the moments of force at the ankle, knee, hip, shoulder and elbow joints during the grab start of swimming.

## METHODS

Eight female varsity swimmers served as subjects after giving informed consent. Each subject performed five trials from a starting platform instrumented with a force platform (Kistler). Forces were sampled at 200 Hz while the subjects were filmed from a sagittal view at 50 fps by a cine camera (LOCAM). Inverse dynamics methods were used to calculate forces, moments and powers at each of the six joints. It was assumed that the subjects performed the starts with bilateral symmetry so only one side of the body was digitized. Takeoff parameters, such as, takeoff angle, velocity and flight time were determined from the marker kinematics.

## RESULTS AND DISCUSSION

The external work done and the peak positive powers by the five moments of force averaged across the eight subjects are shown in Table 1. Figure 1 shows the ensemble average moment power histories produced by the five joints. Clearly the hip moment of force was the most important contributor to the work done during the start. It was also the first of the leg moments to be recruited; an observation in accordance with established coaching principles. Surprisingly, the knee and ankle moments were recruited almost simultaneously rather than sequentially. This result agrees with the research of Robertson and Fleming (1987). They showed that during vertical jumping all three leg joints were recruited simultaneously. Another interesting result concerns the negative work done by the hip moments immediately prior to takeoff. This work occurred to prevent the hip and trunk from hyperextending and to keep the swimmer in a shallow trajectory. As reported in other swimming literature, takeoff angles (projection angles) were very low averaging only 1.2 degrees above the horizontal. It seems that swimmers do not try to obtain maximal flight time despite the reduced drag from air as compared to water but prefer a shallow trajectory to obtain a low angle of entry into the water. In conclusion, the hip moment was the most important contributor to the swim start followed closely by the knee moment. Notably, the elbow and to a lesser extent the shoulder moments contributed significant amounts of work, particularly when compared to their contributions for running or sprinting. Interestingly, the ankle moment of force was not a major contributor to the external work done even though it contributed a large burst of power in the final 30% of the thrust. This occurred because it contributed negative work for the first 70% to control the rate of lowering of the body by dorsiflexors.

**Table 1: Work done and peak moment powers.**

| | Work (J) | Work/Total (%) | Peak power (W) |
|---|---|---|---|
| Ankle | 43 | 10.1 | 650 |
| Knee | 122 | 28.4 | 1100 |
| Hip | 158 | 36.5 | 1200 |
| Elbow | 68 | 15.8 | 290 |
| Shoulder | 37 | 8.6 | 490 |
| Total | 428 | 100.0 | n.a. |

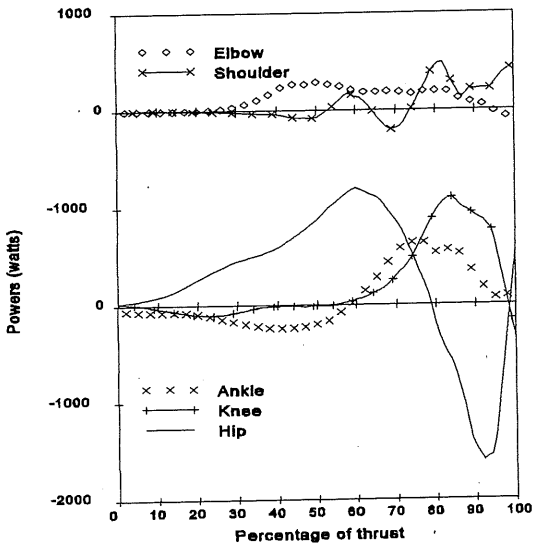



**Figure 1.** Ensemble averaged moment powers

## REFERENCES

Norman, RW (1975) *How to be an Effective Coach*, pp. 92-111.

# THE EFFECTS OF TETHERED SWIMMING ON FREESTYLE STROKE TECHNIQUES

T.Takahashi and B.D.Wilson
School of Physical Education, University of Otago, New Zealand

## INTRODUCTION

Competitive swimmers will achieve superior performances if effective stroke techniques are maintained throughout races. Resistance training may improve his/her ability to maintain an effective stroke technique and may also develop an individual's tolerance to fatigue. Methods of resistance training in use internationally include the Swim Bench, oversized swim paddles and tethered swimming. Both the Swim Bench and paddle produce stroke patterns unlike those of natural swimming (Sharp et al., 1982; Payton and Lauder, 1995). Previous work on tethered swim training using the Sparta resistance Trainer has shown the loads at which the greatest mechanical power output was achieved.

The purpose of this study was to examine if changes in stroke technique occurred during non-tethered (NT) and tethered (T) freestyle swimming with the Sparta resistance Trainer using the optimal load for power output.

## METHOD

Three male and five female competitive swimmers were used as subjects. All followed their normal pool training sessions with sixteen 25 m repetitions (400 m) of maximal effort NT swimming (4 subjects) or sixteen 25 m repetitions (400 m) of T swimming (4 subjects). The conditions were reversed for a second test occasion after one week. The swimmers were videotaped during the corresponding NT and T laps. The videos were digitized and analyzed with a PEAK 3-D system using three-dimensional techniques. Values for each subject's stroke length, stroke rate, velocity, maximum stroke hand depth, and width during each of the NT and T trials were calculated. Heart rate and finger prick blood samples were taken before the 1st lap (resting level) and after the 4th, 8th, 12th, and 16th laps of each condition. Multiple Analysis of Variance (MANOVA) $p<0.05$ and paired sample t-tests $p<0.01$ were used to determine whether any significant difference existed between NT and T conditions. Blood lactate concentration was correlated with each stroke characteristic.

## RESULTS

Stroke length decreased from means of 1.89 to 1.81 m (NT) to 1.45 to 1.43 m (T). Stroke rate decreased from means of 49.1 to 48.5 stroke·min$^{-1}$ (NT) to 44.0 to 46.7 stroke·min$^{-1}$ (T). Velocity decreased from means of 1.53 to 1.46 m·s$^{-1}$ (NT) to 1.01 to 1.09 m·s$^{-1}$ (T). Within-subject comparisons of NT and T results were significant for stroke length (4th, 8th and 16th trials), stroke rate (4th, 8th and 12th trials) and velocity (all trials). The within-subject difference effect sizes were large for all four trials of stroke length and velocity, large for the 4th and 8th trials of stroke rate, and medium for the 12th and 16th trials of stroke rate. The interaction effects were significant for velocity. There were large differences between subjects for all variables ($p<0.05$). Mean blood lactate concentration increased during both the NT (from 1.08 to 4.19 mmol·l$^{-1}$) and T (from 1.17 to 5.24 mmol·l$^{-1}$) tests but there were no significant differences between conditions ($p<0.05$). There were no differences in the stroke hand depth or the stroke width between NT and T conditions. Correlations of each stroke parameter against blood lactate concentration were significant ($p<0.05$) for NT-stroke length ($r=0.41$), NT and T stroke rate ($r=0.39$, $r=0.39$), T velocity ($r=0.43$) and NT and T heart rate ($r=0.52$, $r=0.62$).

## DISCUSSION

The resisted swimming significantly affected stroke length, stroke rate and velocity but did not alter stroke hand depth and stroke width. Therefore changes in velocity due to the resistance were caused by changes in stroke length and stroke rate rather than changes in stroke geometry. Lactate levels remained low (<5.25 mmol·l$^{-1}$) throughout the experiment possibly indicating low carbohydrate levels and limited lactate production with progressive anaerobic exercise. Maximum heart rate during exercise trials ranged from 158 to 192 beat·min$^{-1}$ suggesting that none of the subjects swam at maximum level. This is perhaps consistent with the need to complete 16 repetitions of each 25 m swim. Keskinen and Komi (1988) showed that blood lactate and stroke rate were significantly correlated in low to high intensity free swimming. This result was not confirmed by the present experiment, possibly because of the small sample size. However, during the T trials the swimmers displayed a significant, positive correlation between blood lactate and stroke rate and velocity. With increasing lactate levels observed at higher trial numbers for NT or T swimming the swimmers increased stroke rates and decreased stroke lengths to maintain swimming velocity.

## REFERENCES

Keskinen, K. L., & Komi, P. V. (1988). Interaction between aerobic/anaerobic loading and biomechanical performance in freestyle swimming. In E. U. Bodo, W. Kurt, & R. Klaus (Eds.), Swimming science V, (pp. 285-293). Champaign: Human Kinetics Publishers, Inc.

Payton, C. J., & Lauder, M. A. (1995). The influence of hand paddles on the kinematics of front crawl swimming. J. Human Movement Studies, 28(4), 175-192.

Sharp, R. L., Troup, J. P., & Costill, D. L. (1982). Relationship between power and sprint freestyle swimming. Med. Sci. in Sports and Exercise, 14(1), 53-56.

# SWIMMING TECHNIQUE AND RACE PERFORMANCE IN 100M INTERNATIONAL FREESTYLE AND BACKSTROKE EVENTS

B. D. WILSON
University of Otago, Dunedin, New Zealand

## INTRODUCTION

Swimming technique characterised by stroke length and frequency has been reviewed by Keskinen,1993. There are large individual difference in technique. Additionally, stroke length has been described as decreasing over successive lengths of swimming events and longer stroke lengths are associated with better performers in front crawl (freestyle). These relationships are equivocal for backstroke. The start, turn, stroking and finish times, have relative importance depending on the event distance (Hay, 1987), and a significance to performance dependent on stroke and event distance (Arellano et al 1996).

The purpose of this study was to identify and compare the swimming technique and race performance of swimmers in 100m freestyle and backstroke international competition events. We hypothesised that the technique and race performance strategy adopted by males and females for the "opposition strokes" of front crawl and back crawl (backstroke) would be similar for the 100m events.

## METHOD

The analysis was based on timing data of the finalists in the 100m backstroke and freestyle events at six international competitions from 1992 to 1996 including both Olympics. The timing data was determined from race video analysis according to the Haljand model and provided information on stroke characteristics in each lap as well as start, turn and finish times for each swimmer. Data collection was supported by FINA and IOC with meet data reports being provided to New Zealand Swimming.

The timing data was entered into an EXCEL spreadsheet and transformations were made on some of the data to ensure that comparisons could be made across the competitions. An example of a data transformation for backstroke is the calculation of the 15m turn time: Timing data for the turn provided over 20m for the 96 Olympics was transformed to a time for 15m by subtracting the times taken to swim 2.5m (in) and 2.5m (out). These times were calculated as "distance/speed", using the free swim speed "in" and "out" of the turn. Data were deleted if there was an outlier due to an error which could not be resolved by reference to the original data reports. Scattergrams were plotted to identify data outliers or data distributions which might invalidate a linear analysis model. SPSS bivariate Pearson correlation and multiple regression were then used to assist in data interpretation.

## RESULTS AND DISCUSSION

The mean 100m freestyle time was 50.1s for males and 55.9s for females. The components of race time; start over 10m (8-7%), free swimming (67-68%), turn over 15m (14%) and finish over 10m (11%) were very similar for male and female events respectively. Good freestyle race performance involved a fast start, a high swimming speed characterised by a long stroke length, and a fast turn, second 50m and finish as evidenced by significant correlations ($p<0.05$) between these variables and race result time. The best predictor of result race time was turn time for males which explained 47% of the variance in race result times. For females the best predictor of result race time was free swim velocity in the second 50m (69%). The second best predictor of result race time was free swim velocity in the second 50m for males (20%) and turn time for females (16%).

The mean backstroke time was 55.9 for males and 62.5 for females. The components of race time; start over 15m (13%), free swimming (62%), turn over 15m (14%) and finish over 10m (11%) were the same for the two events. Good 100m backstroke race performance involved a fast start, a high swimming speed characterised by a high stroke rate, and a fast turn. The best predictor of result race time was turn time for the males which explained 57% of the variance in race result times. For females the best predictor of result race time was free swim velocity in the second 50m (60%). The second best predictor of result race time was free swim velocity in the second 50m for males (21%) and turn time for females (24%).

Thus while swimming times are dissimilar, the components of race time are essentially the same for males and females and for the two strokes. The technique and importance of the race components to race performance differs for the events and for males and females. Long stroke lengths are associated with shorter race times in freestyle and higher stroke rates are associated with shorter race times in backstroke, while the turn time for males and free swim velocity in the second 50m for females appear to be the race components which best predict performance.

## ACKNOWLEDGMENTS

Grant support from Sport Science New Zealand and the Hillary Commission, the support of New Zealand Swimming, and the preparation of the data base by Joanne Swanson are gratefully acknowledged.

## REFERENCES

Arellano, R., Brown P., Cappaert, J., and R. C. Nelson (1996). Application of regression equations in the analysis of 50 and 100m swimming races of the 1992 Olympic Games. Proceedings of the XIV International Symposium on the Biomechanics of Sports. Funchai, Portugal.

Hay J. G. (1987). Swimming Biomechanics: A brief review. Swimming technique 23(3),15-21, 1987.

Keskinen, K. L. (1993). Stroking characteristics of front crawl swimming. Studies in Sport, Physical Education and Health #31, University of Jyvaskyla, Jyvaskyla.

# ANALYSIS OF CERTAIN INVERSE PROBLEMS IN BIOMECHANICS USING NONLINEAR OPTIMIZATION AND BOUNDARY ELEMENT METHOD

Hamid R. Katoozian

Biomedical Engineering Dept., Shahid Beheshti University of Medical Sciences, Tehran, Iran.

**Introduction:** The inverse problem can be defined as the problem where one should estimate the cause from the effect. In structural mechanics, the inverse problem can be formulated into a parameter identification problem in which the set of parameters corresponding to loads and material properties will be estimated by minimizing a suitable cost function [1]. In this report two types of inverse problems in biomechanics of bone are investigated by the boundary element method. It is assumed that the values of *i*)mechanical properties (elastic modulii and Poisson's ratio) or *ii*)applied loads, are not known and hence should be identified by using the given displacement data. For the inverse problem, this paper presents an analysis procedure using the *h*-adaptive quadratic boundary element method for elastic problems combined with the standard nonlinear optimization algorithm. An objective function relating the least-squares difference of model-predicted and known displacements is minimized with respect to unknown material parameters or unknown tractions using method of modified feasible directions (MFD) algorithm [2].

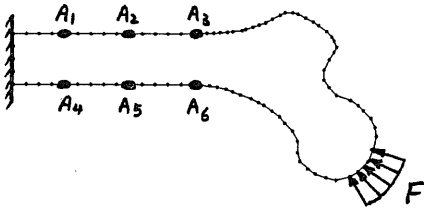
**Method:** A modularized system of stress analysis and displacement calculation using *h*-adaptive quadratic boundary element (BEM) was adopted. The system consists of BEM code, constraint and objective function calculation procedure in connection with optimization module. Variation of nodal displacement to be evaluated by BEM can be considered as a function of the mechanical properties and/or loading conditions, $u(E, \nu)$. The measured nodal displacement $\bar{u}(E, \nu)$ can be used to find the local error. Finally, error estimator function (objective function to be minimized) was taken as the sum of the error over the entire boundary;

$$e_{total} = \sum \left| u(E, \nu) - \bar{u}(E, \nu) \right|^p$$

The exponent $p$ can be set to *2* to emphasize the local peak error and to eliminate the singularities of the absolute values. Solution of the inverse problem begins with incorporating some initial values for the design variables ($E, \nu$, or *traction*), using MFD technique, optimization search ends with final values of design variables that minimize the error function.

**Results:** To verify the efficiency and usefulness of this strategy two numerical experiments are carried out. In these problems, using known values of displacement at six specific sites over the shaft of the femur ($A_1$, $A_2$, $A_3$, $A_4$, $A_5$, $A_6$), elastic characteristics of the femur or traction at the femoral head were computed. Figure shows the two dimensional boundary element mesh of the femur which has been used to demonstrate the validation of the proposed technique. To pass the local minimums initial values were found using parametric studies. Table indicates the results of two runs for traction computation and material properties identification.

| initial values | optimum results | # of iter. | % error reduction |
|---|---|---|---|
| $E$ , $\nu$<br>1.5E+5 , 0.28 | $E$ , $\nu$<br>2.1E+5 , 0.3 | 7 | %95 |
| $F_x$ , $F_y$<br>-18.5 , -4.5 | $F_x$ , $F_y$<br>-21.5 , -6.2 | 5 | %97 |




**Discussion:** The study reported here was an initial attempt to use inverse mechanics method for load and material characteristics identification in biomechanics. The results are generally encouraging in that fairly simplified models were implemented on a PC-level computer at acceptable computational cost. The main goal was to examine the feasibility of the approach. Although it appears the *p*- and *hp*-adaptive BEM techniques are capable of giving fairly good results in terms of error reduction, the *h*-version clearly allows sufficient versatility in the optimal outcomes. In terms of reduction in error function as well as the fastness of convergence, results are encouraging. The efficiency of the proposed technique and its reliability has been proved by the numerical experiments. Further studies will be conducted for three-dimensional model with more realistic material properties.

**References:**

1) Link, M., 1992, "Requirements for the structure of analytical models used for parameter identification", IUTAM Symposium on: Inverse Problems in Engineering Mechanics, Tokyo.

2) Katoozian, H., Davy, D.T., 1993. "Three-dimensional shape optimization of femoral components of total hip prostheses", Bioeng.Conf., ASME, BED-Vol.24, pp.552-555.

# MOTION ANALYSIS IN SPACE : THE EUROMIR '95 EXPERIENCE

G. Ferrigno, G. Baroni, G. Andreoni and A. Pedotti
Dipartimento di Bioingegneria - Politecnico di Milano and Centro di Bioingegneria - Fondazione Pro Juventute Don Carlo Gnocchi and Politecnico di Milano, Milano, Italy

## INTRODUCTION

During ESA 180-days EUROMIR '95 space mission, two in-flight experiments have foreseen 3-D human movement analysis in μgravity. Flight HW requirements and critical working conditions pushed for the accuracy and flexibility of opto-electronic motion analysers (OMA). These take measures within a suitable calibrate working volume, without any contact with the subject nor constraining movements. Rather than any "subject calibrated" device, OMA turn out to be more reliable as permanent facility on board of orbital modules. For this aim, a space qualified OMA, based on passive markers, has been installed on board of the MIR Space Station. The system has been exploited for the acquisition of in-flight kinematics data. A set of sessions has been performed on two subjects, in order to evaluate their motor and postural behaviour, along the whole mission. This work deals with the implementation of the space-qualified motion analysis technology and describes newly conceived methods for its exploitation on board.

## METHOD

The experiments performed during the mission (T4 : "Human Posture in Microgravity" and 38-D : "Vestibular research") required the use of four TV cameras for the motion recording. T4 included 8 experimental sessions. As technological experiment, main goal of the activity was to confirm the chance of successful exploitation of motion analysis technology on board of an orbital module in weightlessness. Up to 50 passive markers have been used and acquisitions as long as 320 s. at 50 Hz were done. ELITE motion analyser [1] has been the HW platform, from which the space-qualified Elite-S has been developed. Such a choice has been justified by the use of passive markers in unlimited number and by on-line marker shape recognition allowing successful operation in noisy environments and reduced marker dimension. Motion analyser space qualification has essentially concerned constructive criteria and system layout. Elite-s has been interfaced with the general purpose Crew Support Computer (IBM ThinkPad 750C). Flight HW was completed by the system calibration form. TV cameras optics (8.5 mm for Tv 1 and 2; 6 mm for Tv 3 and 4) lead to a field of view of 2430x1810 mm, which guaranteed subject's entire body visibility. A specific flight software has been developed in MS-DOS environment. Session performance has been organised in a sequence of SW guided procedures, enriched by the presentation of protocol related drawings and instructions (in English and Russian language). The setting of specific acquisition parameters (marker number, sample rate, acquisition duration, file denomination) has been fully automated (with manual override possibility), decreasing session overhead. The implementation of analytical acquisition statistics, based on the comparison between number of acquired markers vs. nominal number, allowed the operator to verify immediately the quality of the acquisition. Automatic data saving after each experimental session has assured a reliable primary back-up on PCMCIA disks, as well as the selection of data to be down-linked. Flight SW was completed by further tools as electronic notepad and automatic session log generation. System three-dimensional calibration [1] has been modified. Actual calibration parameter computation, based on the calibration frame acquisition performed on orbit, has been included in the data on-ground post-processing. Due to time and extra HW required, no optical lens distortion correction procedure could be foreseen. A shrinkage correction has been included in the on-ground post-processing. Since reference frame was solid with the calibration form, the question of expressing markers 3D co-ordinates with respect to the MIR Core Module reference frame had to be faced on ground exploiting a set of geometrical constraints on Tv camera positions within the MIR Core Module. The implementation of a constrained minimisation algorithm has allowed to compute the coefficients of a rotation matrix, which aligned at each session the form mounted reference frame to the MIR mounted one [2].

## RESULTS

Averaging the camera parameters of the 8 sessions after alignment, a standard deviation not greater than .4 mm has been found for the positions of the cameras (up to 48 mm before correction).
System accuracy in 3-D marker localisation was evaluated by distance on rigid bodies. In experiments using only camera 1 and 2, SD of errors was always below 1.8 mm (0.064%). With camera 3 and 4 (worst for optics and position) the error SD was less than 2.97 mm (0.095%). After experiences in short-term missions and on parabolic flights, this work represents the first permanent installation and repeated use of an automatic OMA in a long term space mission.

## REFERENCES

[1]Ferrigno, G., Borghese N.A. and Pedotti, A. (1990) Pattern recognition in 3-D automatic human motion analysis. ISPRS Journal of Photogrammetry. Remote Sensing 45, 227-246.
[2]Baroni, G., Ferrigno, G., Andreoni, G., Bracciaferri, F. and Pedotti, A. (1996) Accuracy assessment in μgravity human movement analysis, In Proc. Of VIth European Symposium on Life Sciences Research in Space, ESA SP-390 129-134.

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# A BIOMECHANICAL METHOD FOR THE CONSTRUCTION OF A "STANDARD MOTION" AND THE IDENTIFICATION OF ESSENTIAL MOTIONS BY MOTION VARIABILITY

M. Ae* , N.Fujii*, and J.Takamatsu**

* Institute of Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan

** Doctoral Program in Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan

## INTRODUCTION

One of the most difficult steps in optimization process of sports techniques will be identification of technical problems and limiting factors. Because there exist few model techniques for comparison and few methods for the construction of model technique. The purpose of this study was to present a method for the construction of "Standard Motion" as a motion pattern model trough biomechanical data, and a method for the identification of essential viewpoints of sports techniques on the basis of motion variability.

## METHODS

The procedure to construct Standard Motions could be divided into four steps; Step 1: By means of cinematography and/or videography, data collection of skilled and/or excellent athletes was done during official competitions and experimental set-up. Step 2: Normalization of coordinate data and angular data by motion time and anthropometric variables, such as body height of athletes. Step 3: Averaging the normalized coordinate and angular data for Standard Motion. As a measure to quantify motion variability, coefficients of variation were computed on some biomechanical variables, such as joint and segment angles.

## RESULTS AND DISCUSSION

Figure 1 shows an example of Standard Motion for sprint running. We all know that there exist some variability even in the firmly stereotyped human movements and some individual differences among skilled athletes. Figure 2 shows changes in coefficients of variation(CV) for joint and segment angles of the support leg during the support phase. CVs of the thigh, shank, and knee angles of the support leg were small over the whole support phase, but that of the foot was greater in the second half, acceleration phase, than the first half, deceleration phase. The joints and segments with small CV can be referred to as essential joints and segments in sprint because small CV implies that the excellent sprinters showed similar motion patterns. On the other hand, large CV of the foot in the second half implies that foot motion varied sprinter by sprinter. Large CV may be twofold; the first one is that individual differences may be allowed , and the second is that the segments with large CV may have a possibility to be limiting segments which determine the performance.

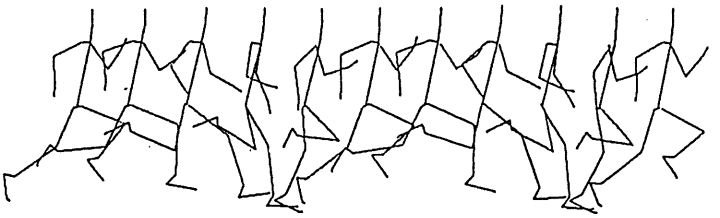


Figure 1 Standard Motion of sprint running

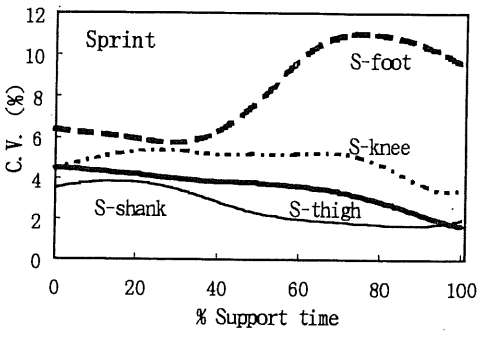



Figure 2 Coefficients of variation of the joint and segment angles during support phase in sprint running.

## 3-D BACK SURFACES AS AN ERGONOMIC TOOL

G.Andreoni[1,3], M.Rabuffetti[3], A.Tarzia[2] and A.Pedotti[1,3]
1 - Politecnico di Milano, Dept. of Bioengineering, Milan, Italy
2 - FIAT Auto, D.T. - I.S.D. Ergonomic Department, Turin, Italy
3 - Centro di Bioingegneria, Fnd. Pro Juventute "Don C.Gnocchi" - Politecnico di Milano, Milan, Italy

### INTRODUCTION

Human environments are designed with respect to anthropometry. Therefore, in seat design back surface represents an important parameter both for manufacturing and for comfort evaluation. For this reason we developed an experimental protocol for the acquisition and for the elaboration of 3-D back surface which can be used as an ergonomic tool.

### METHODS

3-D co-ordinates of surface and anatomical points are measured by means of ELITE system. The experimental protocol foresaw the positioning on the subject of 7 technical markers used to define the anatomical reference systems for upper trunk and pelvis. The surface was scanned through a manually moved infra-red laser beam, which produced on the skin a little spot recognised by ELITE as a marker [Aliverti et al. 1993].

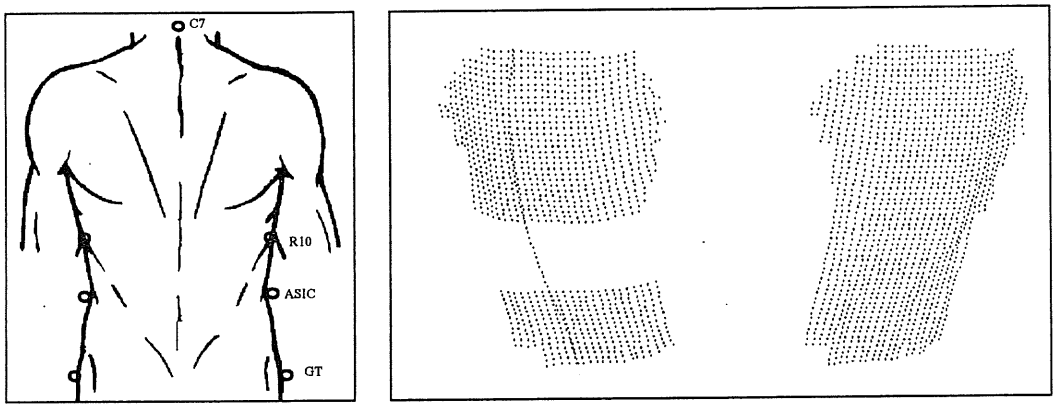



**Fig. 1: The marker positioning for the acquisition of the back surface (right) and the 3-D co-ordinates of the back points (upper trunk, pelvis and spine) and the reconstruction of the lumbar region (left).**

The surfaces of the upper trunk and pelvis surface regions are supposed to be rigid bodies and therefore minimally deformable. This hypothesis was confirmed through a dedicated experimental session.
The two partial surfaces were separately placed in the position identified by the technical markers which corresponds to the sitting posture adopted by the subject. Three different sitting postures were analysed: a normal position with a slight lumbar flexion; a erect sitting posture, where the subject behaved as in the standing position; a slumped sitting posture, characterised by a pronounced kyphosis due to a relaxation of the subject. The lumbar zone was then reconstructed through fitting spline function.

### RESULTS & DISCUSSION

A validation of the rigid body hypothesis were performed: we found that in the central zones both of upper trunk and of pelvis, the difference between the acquired sitting back surface and the repositioned one is about 2.5 mm. This value increased in the lateral regions where border errors can be present. But these lateral zones can be excluded because they don't take part in the man-seat interface.
Different functions for reconstructing the lumbar region were tested comparing the obtained data with the data directly acquired by the ELITE system. The best reconstructing method was a fitting spline function which showed a low error (mean error was 3 mm considering all the lumbar surface) in the normal and in the erect sitting; greater errors were found in the slumped one.
This method wants to offer the opportunity to use anatomical surfaces as ergonomic parameter in helping the designers to evaluate different products or prototypes. A first in progress application of the developed experimental protocol is in automobile ergonomics for improving the comfort and the safety of car seat.

### REFERENCES

Aliverti, A., Ferrigno G., Pedotti A. (1993), Surface analysis by laser beam scanning and stereophotogrammetry. in: Videometrics II, vol. 2067, S.F. El-Hakim (Ed.), 209-219.

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# BIOMECHANICAL MODELLING OF THE HUMAN SPINE

## – Optimization of the Thrust Line in the Human Spine Modelled as an Arch

D.C.Xiao[1], B.S.Acar[2], K.Case[1] and J.M.Porter[3]

[1] Department of Manufacturing Engineering, [2] Department of Computer Studies

[3] Department of Design and Technology

Loughborough University, Loughborough, Leicestershire LE11 3TU, UK

**INTRODUCTION** Lumbar back pain and disorder may be related to spine curvature and disc pressure. The human spine is a statically indeterminate structure. Many attempts have been made to model the spine as levers, cantilevers, beams, or columns using elastic analysis and as an arch using plastic analysis with the criterion that stability requires that a thrust line should be located within the "core" of the arch[2]. However the criterion for the failure of the arch spine and determination of the best–fitting thrust line in the spine are still problems.

**METHODS** (1) Plastic theory and "Middle third rule" [2] are used to generate a criterion for the failure of the arch spine. If the best–fitting thrust line amongst all thrust lines in an arch spine can be found, but not located within the "core"[2] of the spine, then hinges form and spinal failure or disorders can occur. (2) The methods used to calculate the best–fitting thrust line of the arch from force polygons are described as is the computer optimisation whose objective function is that the thrust line is as close as possible to the centre line or reference line of the arch spine. Four objective functions are described as follows:

a) $f_1$ = Minimise [ max $d_i$ ] i=1, n

b) $f_2$ = Minimise [ $\sum d_i^2$ ] i=1, n

c) $f_3$ = Minimise [ $\sum w_i \times d_i$ ] i=1, n

d) $f_4$ = Minimise [ $w_1 \times d_n + w_2 \times \max d_i$ ] i=1, n–1

where d is the distance between current calculated thrust line and the spine reference (or centre) line and w is a weighting factor for the optimization calculation. The best result of several locally optimised values of the above four objective functions was chosen as the final optimisation result.

**RESULTS** The forward bending, heavy load problem [1] is taken as a calculation example. A person in a stooped position carrying a 900 N load in hands with an intra–abdominal pressure of 70 N or 14 N acting perpendicularly to each of the vertebrae L1 to L5. The body weight, intra–abdominal pressure and additional load carried are treated as the external loads. The ligament and muscle forces are treated as internal forces. Optimization calculations show that (1) the best fitting thrust line optimised is closer than the one [1] to the centre line of the spine. A better predictor is obtained by optimization techniques; (2) If the best–fitting thrust line reaches to the surface of the "core" of the arch spine, hinges might be generated and spinal failure or disorders may occur; (3) if the intra–abdominal pressure is reduced from 70 N to 14 N, the best–fitting thrust line would move forward to the centre line of the spine and close it; (4) if a lordosis is introduced and the centre line of the spine is located at the best–fitting thrust line, then the second calculated thrust line has almost the same position as the first one.

**DISCUSSION** (1) the best–fitting thrust line and a better predictor can be obtained by optimization techniques; (2) the spine disorders may occur as the best–fitting thrust line is not located within the "core" of the spine; (3) the greater the intra–abdominal pressure, the further away the thrust line from the centre line of the spine; (4) a lordosis is introduced and locates the spine centre line at the thrust line, the thrust line optimised under this condition would approach to the centre line. Further work is concerned with integration with the SAMMIE computer aided ergonomics design system.

**ACKNOWLEDGEMENT** The research is funded by EPSRC (Engineering and Physical Sciences Research Council) of the UK under Grant No. GR/K58241, 'Spine Modelling: A Tool for Safe to Use Equipment Design'.

REFERENCES [1] Aspden R.M.,The Spine as an Arch: A New Mathematical Model, Spine, 1989, 14,pp266–274.

[2] Heyman, J., *The Masonry Arch*, Ellis Horwood Series in Engineering Science, Ellis Horwood Ltd, Publishers, Chichester, New York, Brisbane, 1982.

## SPINE CURVATURE ANALYSIS OF 97 SUBJECTS

C. Duque[1], R. Brenzikofer[2] and R.Barros[1]
[1] Faculty of Physical Education - State University of Campinas/SP-Brasil
[2] Institute of Physics "Gleb Wataghin" - State University of Campinas/SP-Brasil

INTRODUCTION

The aim of this presentation is to show preliminary results about the spine curve geometrical characterization of 97 volunteers subjects (52 females and 45 males), all of them bank clerks. The principal variable used in this work is the conventional geometric curvature which is measured by the inverse of the radius of the circle that fits locally the analyzed curve in the 3D space.

METHODOLOGY

The back of the analyzed subjects where marked with contrasting adhesives disks placed along the spinous processes of the vertebras. The anatomic points were marked with the subject in the static erect position, ready to be photographed. A pair of stereoscopic photographs of the subjects was taken in front of a calibrated reference system. A precise 3D reconstruction of the marks position was undertaken using the methodology described by Brenzikofer (1993). The coordinates of the reconstructed anatomic points were fitted by polynomial functions and the local curvature of the 3D curve calculated as function of the vertical position.

RESULTS

The results for each of the 97 subject includes projections of the 3D spinal curve in the sagital and frontal planes and the geometric curvature. We show them for one subject in graphical form as functions of the vertical coordinate.

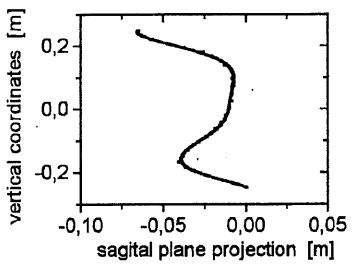



**Fig. 1. Sagital plane projection**

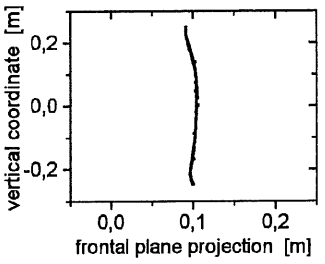



**Fig. 2. Frontal plane projection**

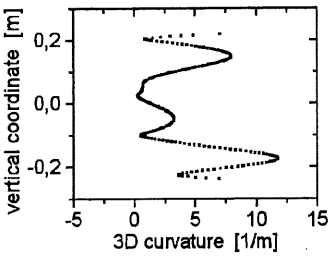



**Fig. 3. 3D Curvature**

DISCUSSION

In the example of figures 1,2 and 3 the thoracic region appears at vertical coordinates -0.01m < z < +0.20m. Figure 1 shows qualitatively a rectificated section of the column between two accentuated curvatures in the sagital plane. In frontal plane (figure 2) the shown lateral deviations are of low amplitude. Figure 3 permits a precise localization and quantification of the characteristics observed in figure 1. The major curvature ($C = 8.0\ m^{-1}$) is localized at z = 0.15m, near the cervic-thoracic transition. A smaller maximum in curvature ($C = 2{,}8 m^{-1}$) is present at z = -0.05m. The rectified section has low $C < 1.0 m^{-1}$ for -0.22m < z < -0.10m.

The lumbar region has vertical coordinates between -0.22m < z < -0.10m and shows a strong curvature of $C = 11.7\ m^{-1}$.

REFERENCE

Brenzikofer, R. Three-Dimensional Reconstruction for Classroom and Laboratory; Proceedings of the XIV I.S.B. Congress, pp. 216-217, Paris, France, 4-8 July, 1993.

Supported by: CAPES, FAPESP, CNPQ, FAEP.

# BIOMECHANICAL EVALUATION OF A SPINAL INSTRUMENTATION USING A NEW TEST APPARATUS - EXPERIMENTAL AND FINITE-ELEMENT INVESTIGATIONS

H. -H. Chen, C.-K. Cheng, H. -Y. Shao, C. -L. Liu*
Institute of Biomedical Engineering, National Yang-Ming University, Taipei, Taiwan, R. O. C.
* Department of Orthopaedic Surgery, Veterans General Hospital - Taipei, Taiwan, R. O. C.

## INTRODUCTION

Standardized test protocol using artificial spine models was developed to characterize the spinal instrumentation, including their likely failure modes (Cunningham et al., 1993; Goel et al., 1994). Such studies have proved useful method for manufacturers to obtain Food and Drug Administration (510K) approval for the devices(Geiger and Melkerson, 1995). A new anterior spinal instrumentation was proposed to support the end-plate in order to provide rigidity and enhance bony union for the spinal fusion, especially in the osteoporotic spine(Liu et al., 1996). To evaluate the applicability of the finite-element method, the biomechanical behaviors of this new device were studied by using the new experimental and finite element methods..

## METHODS

To simulate the vertebral bodies in a corpectomy model, two Ultra High Molecular Weight Polyethylene (UHMWPE) blocks were used. Two blocks were then connected by two kinds of spinal instrumentation, the Kaneda device and a new designed spinal instrumentation, and kept a 76 mm gap between them to simulate the corpectomy model. Besides the mechanical test, a three-dimensional corpectomy model, consisted of 1646 elements and 1915 nodes, was constructed and analyzed with the finite element software ANSYS 5.0. The test procedures included compression, flexion and torsion.

## RESULTS

The compression stiffness, the torsion stiffness and the bending displacement of the new device was 1381 N/mm, 17.4 N-m/deg., and 1.36 mm, respectively. The Kaneda device had the stiffnesses of 1238 N/mm, 17.9 N-m/deg., and 1.21 mm, respectively. The stabilization effect of the new device was similar to that of the Kaneda device. The errors between both methods were 0.55 % in compression, 16 % in torsion, and 46 % in bending (Table 1). The good accordance between two methods was revealed, especially in compression test.

## DISCUSSION

The present study showed that the new anterior device had the similar stiffnesses to the Kaneda device in the UHMWPE model. If the fixation is confirmed, the new device can stabilize the spinal segment as the Kaneda device does. The advantages of the new test apparatus for the spinal instrumentation are: (1)easy preparation (2) easy mounting for multiple loading, (3) reproducible specimen. The present study showed that an equivalent results of mechanical behavior of the experimental method could be done by a finite element model. The substitution of finite element model for the experimental measurement was suggested for the evaluation of the spinal instrumentation.

**Table 1. Comparison of mechanical measurements between experimental and finite element methods.**

| Measurement | Experimental Method | Finite Element Method | Error |
|---|---|---|---|
| compression (mm) | 0.192 | 0.191 | 0.55 % |
| torsion (deg.) | 0.35 | 0.41 | 16 % |
| bending (mm) | 1.36 | 1.99 | 46 % |

## REFERENCES

Cunningham, B. W., Sefter, J. C., Shono, Y. and McAfee, P. C. (1993) Static and cyclic biomechanical analysis of pedicle screw spinal constructs. Spine 18, 1677-88.

Geiger, J. M. and Melkerson, M. N. (1995) Static and fatigue test methods for spinal implants constructs in a corpectomy models - Draft C. ASTM F-04.25.04.01.

Goel, V. K., Winterbottom, J. M. and Weinstein, J. N. (1994) A method for the fatigue testing of pedicle screw fixation devices. J Biomech 27, 1383-8.

Liu, C.-L., Kao, H.-C., Cheng, C-K, Wu, C.-M., Jao W.-T., Wong, T.-H. , Shih, C.-M. and Lo W.-H. (1996) Biomechanical evaluation of a new anterior spinal instrumentation with end-plate fixation. J Orthop Surg ROC 13, 11-15.

# EXPERIMENTAL PROCEDURE FOR STUDYING SCOLIOSIS DURING SURGERY

B.Watier*, C. Lecire*, I. Ghanem**, J. Dubousset**, W. Skalli*, F. Lavaste*

*: Laboratoire de Biomécanique , ENSAM, Paris, France

**: Hôpital St Vincent de Paul, Paris, France

## INTRODUCTION

The scoliosis is characterized by a three dimensional deformity of the spine leading to an instability of this structure. The surgical treatment of this pathology involves a complex movement of the vertebrae whose knowledge is essential for understanding correction mechanisms[1]. The present study aims to analyze the surgical correction through the three dimensional displacement of scoliotic patient's vertebrae during the COTREL-DUBOUSSET instrumentation.

## METHODS

VICON 370, included 3 cameras was placed in the surgery room in front of the patient installed on a COTREL table. Two markers were placed on the skin at C7 and S1 level and permit to analyze global displacement of spinal axis during correction. A « pen » including 8 markers was developed and used to calculate three-dimensional position of particular points on each vertebra, before and after correction (fig.1). Moreover, « quadripodes » including 4 markers ware inserted in some vertebrae spinous process to visualize three-dimensional kinematics of correction. These « quadripodes » were implemented on the apical vertebra, on the extreme vertebrae of the instrumented segment (lower and upper) and on the adjacent ones (non instrumented), and one of them was positioned in the sacrum to verify if there is a movement in this part during correction (fig. 2).

Three cadavers were used for control. Eight patients were followed using this procedure.

With the three dimensional coordinates obtained on three point of each vertebras with the « pen », a coordinate system was reconstructed and we can localize the three-dimensional positions of the vertebrae at the initial and the final position.

Three-dimensional translations and rotations (Bryant angles) were calculated for each vertebra instrumented with « quadripodes » and results are expressed in a global coordinate system of the spine (defined by the axis C7-S1 and pelvis). As large rotations in space occur, the sequence of rotation matters[2]: we chose X, Y', Z". So, we have defined our rotations of vertebrae during correction as: flexion-extension, lateral bending and axial rotation.

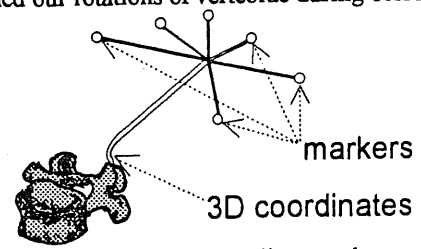



fig. 1: The « pen » used to localize vertebrae

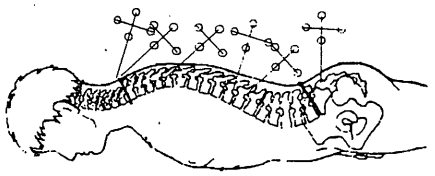


fig. 2: Patient with the six « quadripodes »

## RESULTS

The preliminary results of this study showed us some difficulties, such as important noise (fig. 3) and loss of markers. However we have calculated that the axial rotation of apical vertebrae increased in thoracic curvatures (concave rod rotated) and decreased in thoracolumbar and lumbar curvatures (convex rod rotated). Y axis translation (lateral) was the major displacement observed.

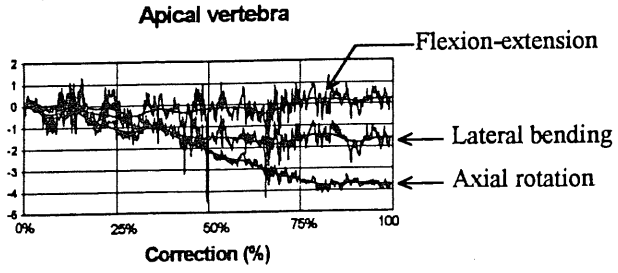



fig. 3 : 3D behavior of apical vertebra during correction by CD instrumentation

## DISCUSSION

Published reports indicates that CD instrumentation has been more effective in producing spinal derotation than vertebral axial derotation, but none of them analyzed intra-operatively the effects on the vertebrae produced solely by rotation of the rod. This study showed that rotation of the rod produces rotational and translational changes along each axis. Axial rotation of apical vertebra worsens in thoracic scoliosis, and improves in thoracolumbar and lumbar scoliosis.

## RÉFÉRENCE

1. Dubousset J, Graf H, Miladi L, Cotrel Y (1986) : Spinal and thoracic derotation with CD instrumentation. Orthop. Trans. **10**, 36.
2. Skalli W, Lavaste F, Descrimes JL (1995) : Quantification of 3D vertebral rotations in scoliosis : what are the true values ? Spine **20**, 546-553.

# ELIMINATION OF TORQUE ARTIFACT IN DYNAMOMETERS

K.S. Olree*, J.R. Engsberg*^, S.A. Ross*
*Human Performance Laboratory, Barnes-Jewish and St. Louis Children's Hospitals, ^Washington University School of Medicine, St. Louis, MO, USA.

## INTRODUCTION

Isokinetic dynamometer torque measures generally include: 1) the net torque about the joint (ie the torques caused by the bony geometry and all active and passive structures crossing it), 2) a gravitational component due to the weight of the limb, and 3) an inertial component due to the acceleration/deceleration of the limb-dynamometer system. Usually, only the net torque is of interest. The gravitational component can be subtracted out by weighing the limb or estimating its mass. With respect to removing the inertial component (or torque artifact), a commonly used strategy is to provide an initial starting resistance to control acceleration during active muscular contractions (Gransberg et al., 1983). However, this method delays the angle at which final velocity is reached and will not work for passive torques (ie torques measured with the muscle relaxed as the dynamometer moves the limb). The purpose of this investigation was to develop a more robust method for eliminating torque artifact.

## METHODS

Passive torques were recorded from the knees of 17 children with cerebral palsy on a KinCom dynamometer as the limb was moved from flexion to the end of extension at speeds of 10, 30 , 60 and 90° /s (Engsberg et al., 1996). The gravity component was calculated and subtracted from the recording in software. The inertial component was modeled as the response of an underdamped second-order system responding to a step input (Kuo, 1987). This was done by first estimating the time ($t_n$) and magnitude ($c(t_n)$) of the overshoots and undershoots from the gravity corrected curve (Fig 1, max overshoot at n=1). $c(t_n)$ was then used to calculate the damping ratio, $\zeta_n$ , using eqn 1. Using $\zeta_n$ and $t_n$, the natural frequency, $\omega_n$, was found from eqn 2. The inertial torque artifact was then modeled by eqn 3 (Fig. 2). This inertial curve was subsequently subtracted from the gravity corrected curve to yield the net joint torque (Fig. 3). If the torque artifact was not successfully removed, the initial estimates of $c(t_n)$ and $t_n$ were revised and the procedure repeated until no overshoot or oscillations were present in the resulting data.

$$\zeta_n = \sqrt{\frac{\left(\frac{\ln\frac{c(t_n)-1}{(-1)^{n-1}}}{n\pi}\right)^2}{1+\left(\frac{\ln\frac{c(t_n)-1}{(-1)^{n-1}}}{n\pi}\right)^2}} \quad (1)$$

$$\omega_n = \frac{n\pi}{t_n\sqrt{1-\zeta_n^2}} \quad (2)$$

$$c_i(t) = \frac{e^{-\zeta_n\omega_n t}}{\sqrt{1-\zeta_n^2}}\sin\left[\omega_n\sqrt{1-\zeta_n^2}\,t - \arctan\frac{\sqrt{1-\zeta_n^2}}{-\zeta_n}\right] \quad (3)$$

## RESULTS

The process removed the inertial torque artifact (compare Figures 1 & 3). Similar results were obtained for all 17 children at the four different speeds.

## DISCUSSION

This versatile method reduces the inertial torque artifact associated with dynamometers. However, as the final velocity, mass of limb, or point of torque measure increase, it becomes more difficult to find valid estimates of $c(t_n)$ and $t_n$. In active torque measures, the method could replace or be used in conjunction with an initial starting resistance to eliminate residual artifact. The method also allows passive testing protocols which can be useful for quantifying muscular disorders such as spasticity, a velocity dependent resistance to passive stretch.

## REFERENCES

1. Engsberg JR et al. 1996. Quantitative clinical measure of spasticity in children with cerebral palsy. *Arch Phys Med Rehabil* 77:594-99.
2. Gransberg L et al. 1983. Determination of dynamic muscle strength in man with acceleration controlled isokinetic movements. *Acta Phys Scand* 119:317-20.
3. Kuo BC. 1987. *Automatic Control Systems*. Prentice-Hall Inc., Englewood Cliffs, NJ. 313-26.

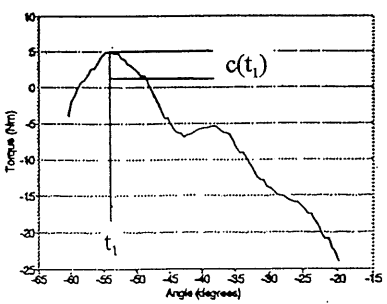



Figure 1. Passive torque-angle data for 1 subject at 90°/s before removing inertial torque artifact.

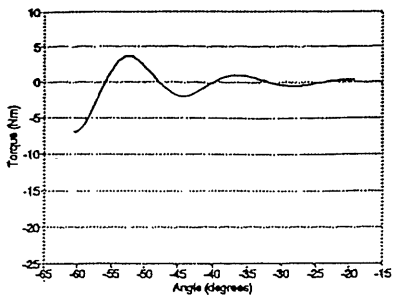



Figure 2. Modeled inertial torque artifact from eqn 3.

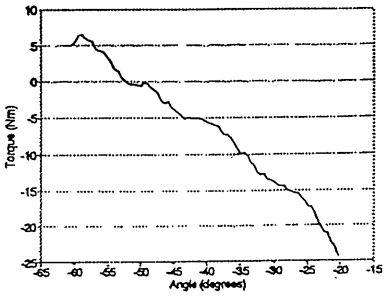



Figure 3. Torque-angle data after removing inertial torque artifact.

# THE EFFECT OF MUSCLE LENGTH REDUCTION ON GASTROCNEMIUS MOTONEURON EXCITABILITY

Andrew G. Cresswell, Wolfgang N. Löscher and Alf Thorstensson
Departments of Neuroscience and Human Biology, Karolinska Institute and University College of Physical Education and Sport, Stockholm, Sweden.

## INTRODUCTION

An earlier investigation on electromyographic activity (EMG) and muscle length (Cresswell *et al.*, 1995) showed a continuous reduction of EMG in the gastrocnemius muscle as its length and force producing capability was gradually reduced via knee flexion. This reduction in gastrocnemius EMG occurred despite the fact that all contractions were of maximal effort. It was concluded that the EMG reduction was at least partially due to a decrease in the number of muscle fibres within the EMG electrode recording volume. However, it could not be excluded that the EMG reduction was also due to increased inhibition and/or reduced facilitation of the motoneuron pool. The aim of this study was to determine whether gastrocnemius motoneuron pool excitability is reduced upon its shortening via knee flexion.

## METHODS

Subjects lay prone with their left foot secured to a force instrumented foot-plate at an internal ankle angle of 90°. Adjustments made under the thigh and trunk created internal knee angles of either 180 or 60°. Voluntarily produced plantar flexor torque and a torque ramp of 1.8 $Nm \cdot s^{-1}$ were displayed simultaneously on an oscilloscope to the subject. Starting from rest, the task for the subjects was to match as closely as possible their voluntary plantar flexor torque to that of the template torque ramp for a period of 10 seconds. Intramuscular interference pattern EMG was recorded from the soleus muscle (SOL) using fine-wire bipolar electrodes with a potential sensitive length of 4mm and separated approximately 10mm. Single motor unit activity from the medial gastrocnemius muscle (MG) was recorded using a concentric needle electrode with a potential sensitive area of .07mm. Twenty voluntary ramps and recordings were made in a randomised order at knee angles of 180 and 60°. For each measurement the MG recording electrode was advanced to a new site within MG. Care was taken to ensure that the same region of MG was used for recordings in the 180 and 60° knee positions. If no MG motor unit activity was observed (or heard) during the 10 second ramp, the subject was instructed to continue the torque ramp until a unit volley was clearly discriminated. Torque, SOL EMG and MG EMG were sampled at frequencies of 100, 2,000 and 12,000 kHz, respectively, then stored on hard disk for post processing. SOL EMG was RMS processed using a bin width of 200ms. Onset of MG activity was determined visually as the time when at least 5 compound motor unit action potentials from the same unit, clearly distinguishable from the baseline noise, were recorded with an inter-spike interval not greater than 300ms.

## RESULTS

To produce a plantar flexor torque of 18 Nm the central drive to SOL, as measured by SOL RMS, was significantly greater (2.7 times) in the 60° as compared to 180° knee position. The level of SOL RMS corresponding to the time of onset of MG single unit activity was significantly greater in the 60° knee position (short MG) as compared to the 180° (long MG) position. At 180° of knee flexion all subjects recruited an MG unit prior to 4Nm of plantar flexor torque, while in the 60° knee position there were many occurrences when an MG unit could not be recorded until well after 10s of ramp contraction (at torque levels greater than 18 Nm). No significant difference was seen for MG compound motor unit action potential size, and/or duration of the first negative phase, in either the 180° or 60° knee positions.

## DISCUSSION

The main finding of the study was that for a given voluntary submaximal level of central drive to the plantar flexor muscles, gastrocnemius motoneuron excitability was reduced when its overall muscle length was shortened. This was intimated by the commencement of MG motor units in the shortened muscle at levels of SOL drive that far exceeded those when the muscle was lengthened. In the shortened muscle, the effect of a decrease in the number of muscle fibres within the EMG electrode recording volume was, in all likelihood, small as MG potentials were of similar magnitude and shape in the two MG length conditions. It therefore seems plausible that in the shortened condition, when many MG fibres are incapable of producing force (Cresswell *et al.*, 1995; Herzog *et al.*, 1991; Rack and Westbury, 1969), peripheral feedback to the MG motoneuron pool increases the recruitment threshold of those units not capable of significant force production. Such a mechanism would indicate a sophisticated feedback control of MG motoneurons based upon their respective muscle fibre length and thus force production.

## REFERENCES

Cresswell, A.G., Löscher, W.N. and Thorstensson, A. (1995) Influence of gastrocnemius muscle length on triceps surae torque development and electromyographic activity in man. *Exp. Brain Res.* **105**, 283-290.

Herzog, W., Read, L.J. and Keyrs, H.E.dJ. (1991) Experimental determination of force-length relations in intact human gastrocnemius muscle. *Clin. Biomech.* **6**, 230-238.

Rack, P.M.H. and Westbury, D.R. (1969) The effects of length and stimulation rate on tension in the isometric cat soleus muscle. *J. Physiol. (Lond.)* **204**, 443-460.

# LONG-LASTING OR TRANSIENT FORCE DEPRESSIONS IN SKELETAL MUSCLES FOLLOWING SHORTENING?

W. Herzog, T.R. Leonard

The University of Calgary, Calgary, Alberta, Canada

## INTRODUCTION

It has been known for half a century (1) that the isometric force following skeletal muscle shortening is depressed compared to the corresponding force obtained for a purely isometric contraction. Several mechanisms have been proposed to explain this phenomenon: sarcomere non-uniformities (2), accumulations of inorganic phosphates and protons (3), and inhibition of cross-bridge attachments (4). The first two of these proposed mechanisms have been shown untenable by experimental evidence (5), however, the third appears to agree with experimental observations. If force depressions following shortening were caused by a reduced affinity of $Ca^{2+}$ to actin binding sites one would expect the force depression to be transitory and disappear within about 1 s. Recently, we have proposed a new mechanism to explain the force depressions (5) which is based on stress-induced cross-bridge inhibitions. This mechanism requires that the force depressions are long-lasting. The purpose of this study was to test whether force depressions following shortening were long-lasting or transitory, therefore providing insight into the mechanism of the observed force depressions.

## METHODS

Force depressions following shortening protocols were measured in cat soleus muscle (n=11) at 37° C. Isokinetic shortening was varied in speed (0-50 mm/s), amount (0.25-8 mm), activation (no activation-supramaximal activation), and was performed at up to 17 different muscle lengths. Force depressions were compared to the corresponding isometric contractions for at least 2 s following the end of the shortening phase. All contractions were performed *in situ* using a muscle puller (MTS) which measured the force and displacement of the soleus for all contractions.

## RESULTS

Force depressions following shortening were most pronounced when the speed of shortening was slow (Fig. 1) and the distance of shortening was large (Fig. 2). For these conditions, force depressions were still present at 2 s post the shortening contractions suggesting that the most pronounced depressions were long-lasting.

## DISCUSSION

Most recently, it was suggested that force depressions following shortening were related to reduced $Ca^{2+}$ affinity on the actin binding sites or a mechanical, stress-induced cross-bridge inhibition. The former mechanism predicts that the force depressions are transient (4), the latter requires a long-lasting force depression (5). The findings of this study support the idea that force depressions following shortening are caused by a stress-induced inhibition of cross-bridge attachments in the new overlap of actin-myosin filaments which has formed during shortening. The force depressions appear to be associated with a mechanical rather than a biochemical cause, and therefore, are long-lasting.

## ACKNOWLEDGEMENTS

NSERC of Canada

## REFERENCES

1) Abbott, B.C. et al. (1952) *J Physiol* 117:77-86, 2) Edman, K.A.P. et al. (1993) *J Physiol* 466:535-552, 3) Granzier, H.L.M. et al. (1989) *J Physiol* 415:499-327, 4) Edman, K.A.P. et al. (1996) *Acta Physiol Scand* 156:183-192, 5) Herzog, W. et al. (submitted *J Biomech*).

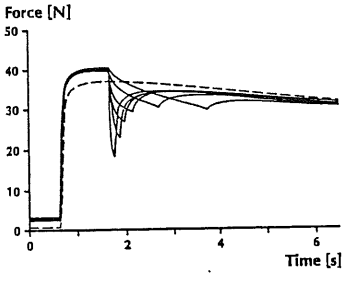



Figure 1. Force depressions as a function of speed of shortening (shortening distance constant). The dashed line represents the isometric reference force.

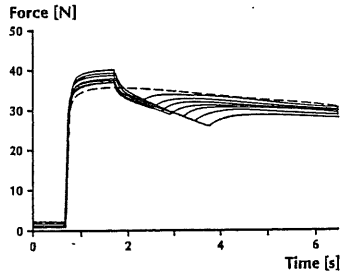



Figure 2. Force depressions as a function of distance shortened (speed constant). The dashed line represents the isometric reference force.

# MODELLING OF POST-SHORTENING FORCE DEFICIT AND POST-LENGTHENING FORCE ENHANCEMENT WITH A HILL TYPE MUSCLE MODEL.

**K. Meijer*, H.J. Grootenboer*, H.F.J.M. Koopman*, B.J.J.J. van der Linden* and P.A. Huijing*§**
***Institute for Biomedical Technology, University of Twente, Enschede, The Netherlands**
**§Institute for Fundamental and Clinical Movement Sciences, Vrije Universiteit, Amsterdam, The Netherlands**

## INTRODUCTION

It is well known that isometric muscle force is depressed after concentric contractions and that it is enhanced after eccentric contractions (e.g. Deleze, 1961). Although, in most muscle models these effects are neglected, they may be functionally relevant. For example, it has been shown that stretch induced enhancement of positive work is at least partially the result of lengthening potentiation (Ettema et al., 1990). The purpose of the present study was to develop a muscle model that accounts for the effects of movement history.

## METHODS.

The basic model is a Hill type muscle model. All parameter values were derived from experimental data of rat medial gastrocnemius muscle (GM). The effects of movement history were incorporated in the contractile element of the Hill model by means of two functions (Fig. 1a & 1b). These functions were derived from GM post-shortening length-force characteristics (Meijer et al., 1996). It was assumed that these functions were also valid for eccentric conditions. With the model simulations of isokinetic shortening and lengthening (v= ±10 mm/s, Δl= 2 mm) of maximally activated GM were performed. The target length, 1 (mm) over optimum muscle length, was the same for all contractions. The results were compared to experimental data of rat GM.

## RESULTS

The main result is that the model can simulate post-shortening force deficit as well as post-lengthening force enhancement (Fig. 1c). Furthermore, the model can account for the length as well as the velocity dependency of these effects. Incorporation of the descriptive functions for movement history into the model reduced the error in the force predictions by approximately 50 % compared to a standard Hill type muscle model (results not shown).

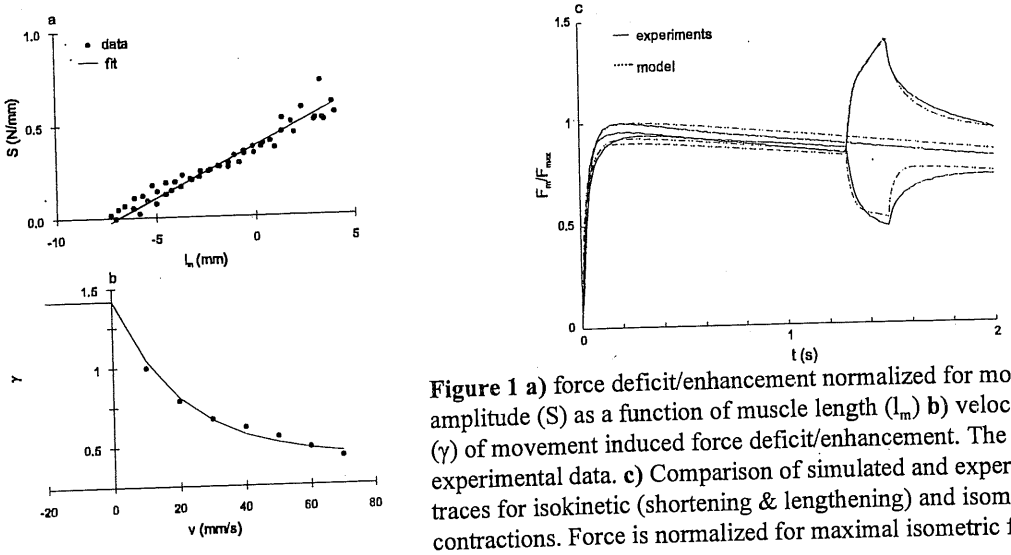



**Figure 1 a)** force deficit/enhancement normalized for movement amplitude (S) as a function of muscle length ($l_m$) **b)** velocity dependence (γ) of movement induced force deficit/enhancement. The dots represent experimental data. **c)** Comparison of simulated and experimental force traces for isokinetic (shortening & lengthening) and isometric contractions. Force is normalized for maximal isometric force ($F_m/F_{max}$)

## DISCUSSION

Incorporation of movement history improves the model predictions considerably. This may be particularly important when muscle models are used to study cyclic movements. Surprisingly, the functions that were derived from concentric contractions also apply for eccentric contractions. This indicates that the mechanisms responsible for post-shortening force deficit and post-lengthening force enhancement are the same. Morgan (1990) showed that distributed behavior of sarcomere lengths in series could be candidate for the mechanism involved. However, based on the results of stiffness measurements performed by Sugi & Tsuchiya (1988), different sources for force deficit and force enhancement are expected. Irrespective of the mechanism, further research will be necessary to evaluate the applicability of the model for more physiological movement patterns (e.g. stretch-shortening cycles).

## REFERENCES

Deleze JB (1961). *J Physiol.* 158: 154-164
Ettema GJC, AJ van Soest and PA Huijing (1990) *J. Exp. Biol.* 154: 121-136.
Meijer K, HJ Grootenboer, HFJM Koopman and PA Huijing (1996).: *Proceedings of the EMBS conference, Amsterdam , The Netherlands; Eds: HBK Boom & WLC Rutten.* p 491.
Morgan DL (1990). *Biophys J.* 57: 209-221
Sugi and Tsuchiya (1988) *J. Physiol. Lond.* 407: 215-229.

# MECHANICAL EXPLANATION OF MUSCLE FIBER SLACK-LENGTH

B.J.J.J. van der Linden[#], H.F.J.M. Koopman[#], P.A. Huijing[#,$] and H.J. Grootenboer[#]
[#]Institute for Biomedical Technology, University of Twente, Enschede, The Netherlands.
[$]Institute for Fundamental and Clinical Movement Sciences, Vrije Universiteit, Amsterdam, The Netherlands.

## INTRODUCTION

A large amount of research is devoted to functioning of skeletal muscle. In many studies muscle functioning is expressed in the force-length relationship. Two characteristic points in this relationship are the fiber active optimum length ($l_{fao}$) and active slack length ($l_{fas}$), where maximal force and no force are exerted at the ends of the fiber respectively. According to the sliding filament theory the position of $l_{fao}$ is related to the point of maximal overlap of the myofilaments, i.e. actin and myosin. The exerted active force above this length decreases proportional with the filament overlap. The mechanism, which generates active force below $l_{fao}$, is not fully understood. It is assumed that at slack length there are no attached cross-bridges, since at this length no force is exerted at the fiber ends.

The goal of this study is to evaluate the mechanism at fiber active slack length mechanically by simulating this with a Finite Element (FE) model. A comparison is made between intact and skinned fibers.

## METHODS

The mechanical analysis of fiber behavior is done with a Finite Element model of muscle (van der Linden et al., 1995). The FE-method approximates mechanical equilibrium of a continuum, in this case the skeletal muscle fiber. The model fiber consists of a finite number of elements. This two-dimensional model is based on a muscle element which has anisotropic passive and active properties. Active stress is generated in fiber direction only. The element obeys the demand of constant volume by keeping the area constant. This muscle property is taken care of by a penalty function. The deviation from the initial element area can be set with parameters of the penalty function. The Poisson ratio for incompressible materials is 0.5 and is approximated by the penalty function. This allows us to study skinned fibers, which do not obey the demand of constant volume according to Maughan and Godt (1979). They experimentally showed a Poisson's ratio of 0.35 for skinned fibers;.

Muscle fiber force-length relationships of intact as well as skinned fibers are constructed. At every fiber length the geometry and the internal stresses are evaluated. This model has the ability to study the active fiber stress separately from the passive stress in fiber direction.

## RESULTS

The simulation results for intact fiber show that no total fiber force is measured at fiber active slack length ($l_{fas}$), which is at 60% of $l_{fao}$. The active fiber stress is still about 40% of the maximum stress in case the area of the model is preserved, deviation is less than 2%. Simulations of skinned fiber result in a shift of active slack length to 45% of $l_{fao}$. The fiber stress is in this case 20% of maximal active fiber stress. The mean area of the elements is reduced to 65% of the initial area.

## DISCUSSION

The results show that the active slack length depends on the demand to conserve muscle volume. This can be explained with the passive muscle properties perpendicular to the fiber direction. Active force, generated in fiber direction, is needed to deform the passive tissue in the direction perpendicular to the fiber. Experimental observations of Brenner and Yu (1991) showed that radial forces are present in $Ca^{2+}$-activated skinned fibers. The authors suggest that attached cross-bridges have a radial elasticity component, which results in radial forces that are proportional to stiffness. Their study also showed that there was no linear coupling between the active axial and radial forces, suggesting that besides cross-bridges there are more structures that contribute to radial forces. Strumpf et al. (1993) showed on canine diaphragm that stiffness perpendicular to the fiber (x-fiber) is highly non-linear and becoming increasingly stiffer as stretch increases. Under passive conditions the x-fiber stiffness is significantly higher than in fiber direction. When the muscle membrane was disrupted, the x-fiber stiffness dramatically decreased, suggesting that the membranes bear most of the passive load and are anisotropic themselves.

The final conclusion is that active slack length is a point where there is equilibrium between the internal forces in a fiber. This mechanism can also explain the observations that for sub-maximally stimulated fiber active slack length shifts to longer lengths (Roszek et al., 1994). The consequence of this mechanism is that at active slack length an active force is generated by the cross-bridges and therefore the fiber does consume energy at the point where no external force is delivered.

## REFERENCES

Brenner, B ,LC Yu (1991), J Physiol, 441:703-718.
Huxley (1957), Prog Biophys Biophys Chem, 7: 255-318.
Maughan, DW, RE Godt (1979), Biophys J, 28: 391-402.
Roszek, B, GC Baan, PA Huijing (1994), J Appl Physiol, 77: 2115-2124.
Strumpf, RK, JD Humphrey, FCP Yin (1993), Am J Physiol, 265: H469-475.
Van der Linden, BJJJ, PA Huijing, HFJM Koopman, K Meijer, M Kuiper, HJ Grootenboer (1995), Proc XV[th] ISB Congress, Jyväskylä, Finland, 956-957.

## CHANGES IN MUSCLE-TENDON UNIT DURING PASSIVE STRETCHING IN HUMANS

Yasuo Kawakami, Yoshiho Ichinose, and Tetsuo Fukunaga
*Department of Life Sci. (Sports Sci.), The University of Tokyo, Komaba, Meguro, Japan*

### INTRODUCTION

Passive stretching is known to increase the joint range of motion (ROM) acutely and chronically. It is conceivable that the increase in the ROM occurs due to elongation of fascicles and/or tendons, and that the degree of elongation depends on the visco-elastic properties of the muscle-tendon unit (2). However, changes in the muscle as well as in the tendon by stretching have not been clarified in humans. In the present study, passive stretching was induced to human muscles, and time-dependent changes in passive torque and fascicle / tendon lengths were determined.

### METHODS

The subjects were six men and one woman aged 22 ~ 31 yrs. The subject lay prone on a test bench, and his/her right foot was attached to a footplate which was connected to an electrical dynamometer (Myoret, Asics, Japan). Passive stretching was induced to the plantarflexor muscles by passively dorsiflexing the ankle to the maximal ROM, and keeping it in position for 10 min. Three sessions were carried out with the same procedure, with a brief rest in between at the resting position. Fascicle length and pennation angles of the medial (MG) and lateral (LG) gastrocnemii and soleus (Sol) muscles were determined by ultrasonography (1). The aponeurosis length of MG (from the distal end to the Achilles tendon) was also measured.

### RESULTS

The ROM (maximal dorsiflexion angle) significantly increased through the first (32 ± 4°, mean ± SD) to third (38 ± 3°) sessions, which was accompanied by a tendency for the fascicle and aponeurosis length to increase, and for the pennation angle to decrease. Elongation of fascicles from the resting length was the most pronounced in Sol (Fig. 1). The passive plantarflexion torque during each step showed an initial peak, after which torque decayed to a plateau (Fig. 2). Surface electromyogram obtained from the triceps surae muscles of two subjects showed a minimal activity during each session.

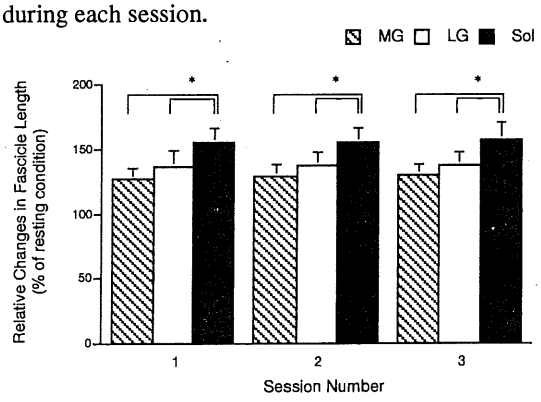



**Fig. 1**

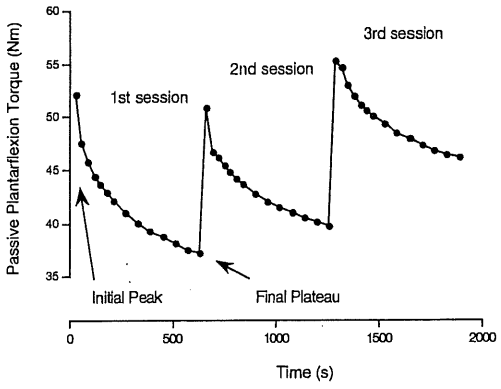



**Fig. 2**

### DISCUSSION

The results would suggest that the increase in passive dorsiflexion angle would be due to elongation of fascicles and possibly tendons (although the magnitude is comparatively small) of the plantar flexor muscles, not from any change of muscle activation. The change in passive plantarflexion torque might be due to the stress relaxation of the viscous element(s) in the muscle-tendon unit (3), which could have changed the extensibility of fascicles and/or tendon and contributed to the increased ROM.

### REFERENCES

1. Fukunaga, T., Ichinose, Y., Ito, M., Kawakami, Y. and Fukashiro, S. (1997) *J. Appl. Physiol.* **82**: 354-358.
2. Taylor, D. C., Dalton, J. D., Seaber, A.V., and Garrett, W. E. (1990) *Am. J. Sports Med.* **18**: 300-309.
3. Toft, E., Sinkjaer, T., Kalund, S. and Espersen, G. T. (1989) *J. Biomech.* **22**: 1129-1132.

# FORCES OF INDIVIDUAL CAT ANKLE EXTENSOR MUSCLES DURING LOCOMOTION PREDICTED USING STATIC OPTIMIZATION

B. I. Prilutsky, W. Herzog and T. L. Allinger
Faculty of Kinesiology, The University of Calgary, Alberta, Canada

## INTRODUCTION

Force patterns of soleus (SO), gastrocnemius (GA), and plantaris (PL) during cat locomotion are stereotypical and similar among cats for a given movement (Herzog et al., 1993). It is not understood why the central nervous system (CNS) chooses these particular force patterns from an infinite number of possible force patterns which could provide similar movements at the ankle. It seems appealing to use mathematical optimization to search for the "optimal" physiological principles that might explain the mechanisms and the functional significance of the observed force-sharing among synergistic muscles. The purpose of this study was to predict and validate forces of cat SO, GA, and PL during walking and trotting using optimization criteria proposed in the literature.

## METHODS

Forces of SO, GA, and PL were measured using 'E'-shaped, stainless steel tendon force transducers in three cats freely walking and trotting on a motor-driven treadmill at speeds of 0.4, 0.8, 1.2, 1.5, and 1.8 m/s. Video records of one camera (60 Hz, Motion Analysis, VP310) and reflective skin markers were used for calculating ankle and knee joint angles and muscle moment arms at the joints. The forces of SO, GA, and PL during cat walking and trotting were also calculated by solving the static optimization problem formulated as follows: for each instant in time, minimize an objective function $\Phi = f(F_1, F_2, F_3)$ subject to the equality constraints $M_j - \Sigma_i d_{ij} F_i = 0$ ($j$=1, 2; $i$=1, 2, 3) and the inequality constraints $F_i \geq 0$ ($i$=1, 2, 3), where $F_1$, $F_2$, and $F_3$ are the unknown forces of SO, GA, and PL, respectively; $M_1$ is the total moment produced by SO, GA, and PL at the ankle joint; $M_2$ is the total moment produced by GA and PL at the knee joint; $d_{ij}$ is the moment arm of the $i$-th muscle relative to the $j$-th joint. Moments $M_1$ and $M_2$ were calculated as the sum of the products of the measured forces and moment arms of the muscles crossing the corresponding joints. Six optimization criteria proposed in the literature were tested. Muscle forces were calculated for each percent (or video frame) of the step cycle and each optimization criterion, using the routine *constr* of the optimization toolbox of MATLAB (The MathWorks Inc.).

## RESULTS AND DISCUSSION

The criterion that minimizes muscle fatigue and accounts for the percentage of slow-twitch fibers in the muscles (Dul et al., 1984) demonstrated the best performance: an average normalized error of force predictions calculated for the stance phase of locomotion of the three cats varied from 26 to 52% for this criterion (assumed percentages of slow-twitch fibers of SO, GA, and PL were 100%, 21%, and 26%, respectively) compared to 34 to 322% for the other examined criteria. The criterion by Dul et al. predicted a loop-like relationship (force-sharing) between SO and PL, and SO and GA forces. However, the shapes of the predicted loops differed substantially from the actual shapes. The muscle forces predicted using the criterion of Dul et al. (1984) for each video frame over seven consecutive step cycles (Fig. 1) showed a high correlation ($r$>0.96) with the corresponding measured forces. The forces predicted using this criterion and different values of the percentage of slow-twitch fibers in GA and PL (within an approximate range reported in the literature) showed that the difference in the average error of force predictions between different muscle fiber composition values was up to two-fold. Out of six examined optimization criteria, only the criterion of Dul et al. (1984) demonstrated the potential to predict all the main features of the actual force-sharing among cat SO, GA, and PL during walking and trotting: the synergistic action, the loop-like force-sharing behavior, and the higher peak stresses in SO than in GA and PL. The results of this study suggest that during cat walking and slow trotting, the CNS preferentially activates muscles with a high percentage of slow-twitch fibers in order to minimize muscle fatigue.

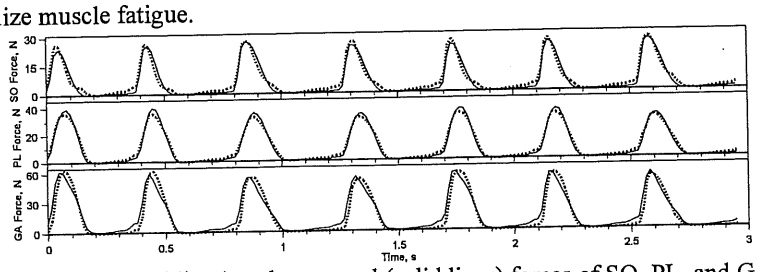



Figure 1. Predicted (dotted lines) and measured (solid lines) forces of SO, PL, and GA during seven consecutive step cycles. The assumed percentages of slow-twitch fibers of SO, GA, and PL were 100%, 20%, and 35%, respectively. Average normalized error of force predictions was 62%. Speed 1.5 m/s.

## REFERENCES

Dul, J. et al. (1984) *J. Biomechanics* **17**, 675-684.
Herzog, W. et al. (1993) *J. Biomechanics* **26**, 945-953.

# AN EVALUATION AND VALIDATION OF PARKINSONIAN GAIT

Nachiappan Chockalingam*, S Yogaraj#, C U Velmurugendran#

* Gait Analysis Laboratory, Central Leather Research Institute, Adyar, Chennai 600 020
# Institute of Neurology, General Hospital, Chennai 600 003

INTRODUCTION:

Parkinsonism is a common disorder with men affected more than women. The beginning of this disorder is around the age of 60. The occurrence of this disorder increases with age. Early symptoms are often interpreted and attributed to aging and other musculo - skeletal problems. Not only recognizing early parkinsonism but also separating from other degenerating diseases poses a diagnostic challenge. Since, parkinsonism is a progressive chronic disorder the current therapeutic modalities do not reverse gait abnormality and postural instability. A detailed gait analysis of parkinsonian patients can give an insight into the postural disturbances and would result soothing the patients with right diagnosis and administering proper physio therapy exercises. The reported work aimed at quantifying the postural disturbances and the main gait parameters of parkinsonian patients. A few studies have been carried out earlier show that these type of studies could give a valuable information. The current work uses a more sophisticated system than the previous studies to validate the findings and to check for new avenues. An emphasis is given to the kinetic parameters of gait that gives a good picture on the postural imbalance.

METHODS:

A VICON movement analysis system with three cameras and two AMTI force platform was employed in the reported study. The software AMASS was used to record and analyze the movement data. The force platform was embedded in the 10 m walk way. Five subjects of an average age of 56, height of 174 cms and a weight of 59 kgs took part in the trials. These subjects were treated with common anti parkinsonian drugs. Retro reflective markers were kept on different anatomical landmarks on the lower limbs for the movement data collection. The subjects were allowed to walk several times to get themselves accustomed to the laboratory environment before the actual data collection. The markers kept at the lateral malleoli and the heads of the fifth metatarsal were given more emphasis. These markers were of main use in the quantifying the postural stability. The direction of progression was marked X , the perpendicular direction as Y and the vertical direction as Z. The force platforms and the software used were capable of recording the path of centre of pressure . The software was also capable of giving the X,Y,Z coordinates of the required markers at a particular instant of time.

The X coordinate of the path of centre of pressure (PX) was graphed along with the X coordinate of the malleoli (RMX) and fifth metatarsal (RFX) marker in the ordinate axis. The time was graphed along the abscissa. These graphs for the time to the PX,RMX,RFX would give an insight into the postural stability of the subjects. The time of line PX that is contained within the lines of RMX and RFX was measured **(A)** . Also the time of line PX outside the RMX and RFX lines **(B)** and the total time taken for the line PX **(C)** were also noted. The ratio of the time of line PX contained within the other two lines is calculated **(R1)**. Similarly, the ratio of the line PX outside the lines RMX and RFX is also calculated **(R2)**.

The points P1, P2, P3 indicated the starting of line PX , the point in which it crosses the line RFX and the end point respectively. The ratio of the distance in millimeters along the ordinate axes for these points were also calculated. The cadence, velocity and stride length values were also recorded and analyzed.

RESULTS:

The observations indicate that the subjects have reduced stride length when compared to normal subjects of similar physical parameters. However, the time taken by the parkinsonian subjects during the first stride is considerably higher than the consecutive strides made at different walk velocities. It was also observed that the cadence and the stride length were dependent on the walk velocity. However, on an average the subjects exhibited a higher double limb support time and reduced stride length.

The ratio R1 was approximately 60.1 % and R2 was 39.9 % for the parkinsonian subjects. The ratios calculated for the normal subjects showed the R1 as 45.2 % and R2 as 54.8 %. However, the total time of PX observed in parkinsonian gait was higher than that of the normals. Also, the ratio indicating the distance between P1 to P2 and P2 to P3 showed a consistent value of 60 % and 40 %. This value of distance ratio is similar between the parkinsonian and the normal subjects. These ratio values indicate progression of path of centre of pressure and the stability. The findings have to be satistically proved with more subjects.

References:

1) M E Moris et al., "The pathogenesis of gait hypokinesia in Parkinson's disease" , Brain (1994) 117, 1169-1181
2) M Vildailhet et al., " The bereitschaftspotential preceding simple foot movement and initiation of gait in Parkinson's disease", Neurology (1993) 43;1784-1788

# STRUCTURAL AND FUNCTIONAL FACTORS THAT DETERMINE PEAK PLANTAR PRESSURE UNDER THE FOOT DURING WALKING

Erez Morag and Peter R. Cavanagh
The Center for Locomotion Studies, Penn State University, University Park, PA, USA

## INTRODUCTION

The relationship between structure and function of the human foothas attracted many investigators over the years, and has led to thedevelopment of many theories. In most cases, these theories have not beenquantitatively verified due, in part, to the complex structure and thevariable function of the human foot. Although there have been a number ofprevious attempts to predict plantar pressure (Stokes 1975, Rodgers 1985,Cavanagh et al. 1997) there is, at present, no model that can fully explainwhy load is distributed in a certain pattern in one person, and indifferent pattern in another. The present model was designed to explorehow a combination of structural and functional factors influence loaddistribution underneath the human foot during walking.

## METHODS

A number of structural and functional outcome measures (potentialindependent variables) were identified from a total of five structure andthree functional data sets as follows: (1) physical characteristics (age, height and weight), (2) anthropometric data, (3) passive range of motion atselected foot joints, (4) standardized weight bearing plain radiographs(lateral views and dorsi-plantar views of both sides individually) of thefoot and ankle, and (5) functional stiffness characteristics of the softtissue under the heel and the first metatarsal head (mth1). The functiondata were obtained from five walking trials in each subject at an averagespeed of 0.78 statures per second. Each walking trial was recorded bythree Mac Reflex cameras, and gait analysis was performed leading to (6)stride foot kinematics, (7) 3D joint kinematics, and (8) EMG. Plantarpressures in the right foot only were recorded simultaneously andsubsequent analysis provided peak pressure values under the rearfoot,midfoot and mth1. Selected potential predictors from each dataset wereidentified, based on their anatomical relevance and their relationship withpeak pressure. Regression analysis was performed to explore what portionof the variance in plantar pressure under the rearfoot, midfoot, andforefoot during walking could be explained by both structural andfunctional characteristics of the foot and lower extremity.

## RESULTS

The final regression models consisted of five to seven variables,introducing a unique combination of structural and functional variables.In this abstract only the heel model is presented. A (+) sign representsdirect association between pressure and a given variable, while a (-) signrepresents inverse association between the two. Heel peak pressure wasfound to be a function of: the inclination of the calcaneus (+), the amountof unloaded soft tissue thickness under the heel (-), contact time duringstance (-), the horizontal velocity of the foot at heel strike (+), thevertical velocity of the foot at one frame after heel strike (+), and age(-). The coefficient of multiple correlation was 0.75 (an explainedvariance of 56.6%, $r^2$ adjusted = 49.2%).

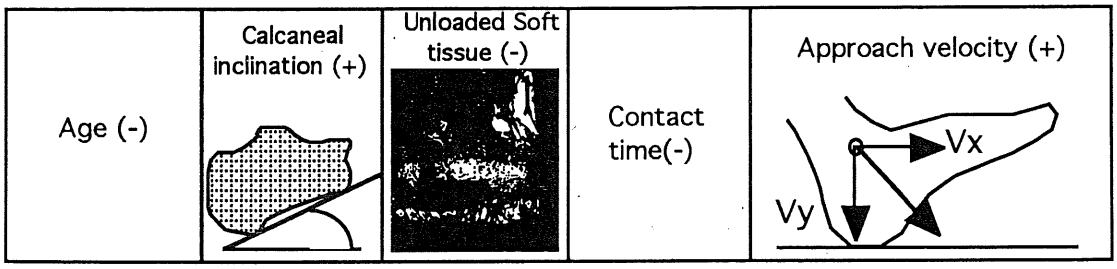



Figure 1. A schematic representation of predictors of peak pressure under the heel

## DISCUSSION

It was concluded that foot structure and function predictapproximately 55% of the variance in peak plantar pressure. Predictiveability for the three regions was similar, but the relative contribution ofstructure and function for the models was region dependent. Structureproved to be dominant in predicting pressure under the midfoot and mth1,while function accounted for the majority of the variance in peak pressurefor the heel region. While it is well known clinically that mechanicalfactors contribute to the development of foot disorders, the resultspresented here provide quantitative data in the form of a multifactorialpredictive model that provides insight into the potential etiologicalfactors that are associated with elevated plantar pressure.

## REFERENCES

Cavanagh PR, Morag E, Boulton AJM, Young MJ, Deffner KT, and Pammer SE(1997). The relationship of static foot structure to dynamic foot function,J. Biomech. 30 (3), 243-250.

Rodgers MM (1985). Plantar pressure distribution measurement duringbarefoot walking: normal values and predictive equations. Doctoraldissertation, The Pennsylvania State University.

Stokes IAF (1975). An analysis of the forces on normal and pathologicalhuman feet. Ph.D. Thesis, The Polytechnic of Central London.

# INTRINSIC MUSCLE PROPERTIES FACILITATE (LOCO-) MOTOR CONTROL

Karin G.M. Gerritsen † ††, Anton J. van den Bogert †† and Manuel Hulliger †††
†Dept. of Medical Sciences, ††Human Performance Laboratory and †††Dept. of Clinical Neurosciences,
The University of Calgary, Alberta, Canada

## INTRODUCTION

Engineering-inspired theories of motor control imply that to compute control solutions (muscle activation inputs), the central nervous system (CNS) requires precise and explicit information about the musculoskeletal system and its environment[1]. On this view, the highly nonlinear length-velocity-tension properties of muscle pose enormous computational demands on the CNS. On the other hand, these spring- and damper-like muscle properties have the potential of contributing to movement stabilization as shown for an explosive task (e.g. maximal vertical jumping[2]).

The goal was to investigate, theoretically, the role of intrinsic muscle properties in the control of locomotion, i.e. during submaximal activation. Specifically, the walking performance of a model actuated by muscles ('biological') was compared with that of a model driven by torque motors (robot-like), under different environmental conditions.

## METHODS

The biological model[3] consisted of a two-dimensional skeletal model (9 DOF) driven by models (Hill-type) of 16 lower extremity muscles, and interacted with the ground through a soft-constraint reaction force model. Modelled nonlinear muscle properties included (whole-muscle) force-length, force-velocity and activation dynamics. Inputs to the model were muscle activation patterns as a function of time. The robot-like model was created by replacing the muscles by joint motors, which produced torque profiles that were identical to those generated by the muscles in the biological model (Fig. 1).

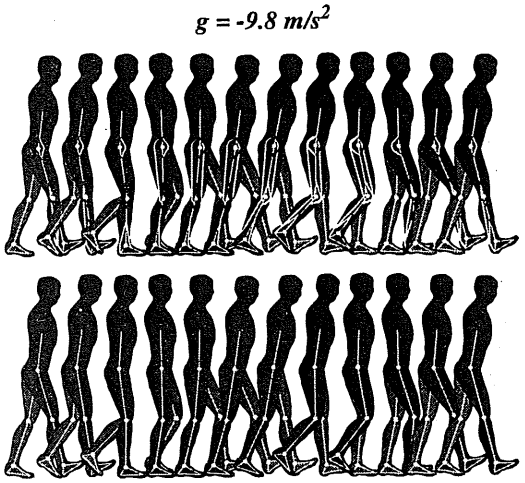



Fig. 1: sequence of stick figures ($\Delta t = 0.12$ s) of a simulated gait cycle for the biological model (top) and for the torque driven 'robot' model (bottom). Net joint torques are identical.

*Address for correspondence:*
kggerrit@acs.ucalgary.ca; http://www.ucalgary.ca/~kggerrit/

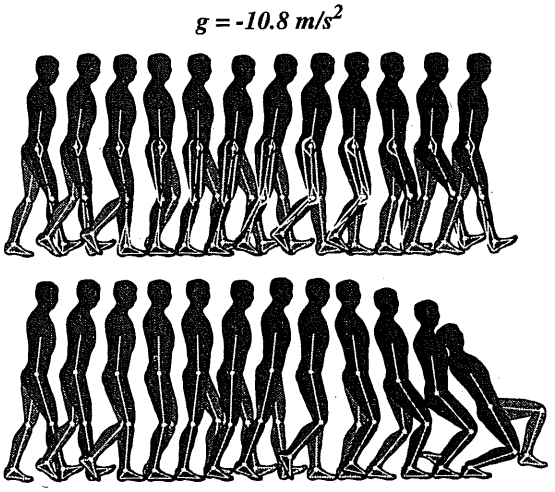



Fig. 2: sequence of stick figures ($\Delta t = 0.12$ s) of the simulated gait cycle during increased gravity for the biological model (top) and the torque driven 'robot' model (bottom).

External conditions were altered by increasing the gravitational acceleration ($g$) by 10%. Inputs to both models (muscle activation patterns and joint torque profiles, respectively) were kept constant.

## RESULTS

The biological model was not noticeably affected by the increase in gravity. Muscle activation patterns that produced normal gait kinematics during standard gravity conditions (Fig. 1) were equally effective during the altered conditions (Fig. 2). In fact, it was not until gravity was increased by 25% that muscle activity needed to be adjusted to avoid an irrecoverable collapse. In contrast, for the torque driven model, the 10% increase in gravity caused substantial deviations from normal gait kinematics (Fig. 2).

## DISCUSSION

The remarkable resistance of the biological model to the imposed alterations in gravity suggests that the nonlinear intrinsic properties of muscle do not further complicate but actually *facilitate* locomotor control, by increasing stability. The role of muscles seems to extend beyond mere actuation. As previously shown for maximal jumping[2], muscles also provide effective immediate feedback during submaximal activation. To some extent this mechanism can be controlled by the CNS by varying co-contraction.

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## REFERENCES

[1]Alexander GE et al. *Behav Brain Sci* **15**:656-665, 1992
[2]Soest AJ van & MF Bobbert *Biol Cybern* **69**:195-204, 1993
[3]Gerritsen KGM et al. *In: SPIE* **V7218**:481-491, 1996

# 3D KINEMATICS OF THE ANKLE JOINT AND MEDIAL ARCH OF THE FOOT AND ACTIVITY OF SELECTED EXTRINSIC MUSCLES DURING THE STANCE PHASE OF WALKING

A.E. Hunt and R.M. Smith
Faculty of Health Sciences, The University of Sydney, Australia

## INTRODUCTION

Despite the fact that muscles are the structures most able to adapt to the varying demands on the foot during walking, very little is understood about their role. The aim of this study was to investigate the relationship between the EMG activity of peroneus longus (PL), tibialis anterior (TA) and soleus (S), and the kinematics of the ankle joint complex (rearfoot) and the medial longitudinal arch of the foot (MLA) during the stance phase of walking.

## METHODS

There were several steps in the data collection protocol, two of which relate to the determination in 3D of the neutral (reference) position of the rearfoot described elsewhere (Moseley et al, 1996). The first of these steps followed the collection of anthropometric and subject history data; the midline of the subjects rearfoot and shank were determined. Surface EMG electrodes were then applied over PL, TA and S, and connected to a telemetry unit. Nine retro-relective markers were attached to the right foot and shank to identify the lower leg segment, the rearfoot, and the MLA. The foot was videorecorded in the reference position, and then the subject walked along the walkway in which a force plate was imbedded; 5 trials were recorded.

Following filtering of the data, the 3D angular displacements of the rearfoot, and the linear changes in arch height and arch length during stance phase were calculated and graphed. EMG data were graphed as linear envelopes. Pearson's bivariate correlations ($p<.05$) were performed in SPSS to confirm (or refute) the relationships that were expected between the dependent variables. Cross-correlations were used to determine the degree of concurrence of patterns for each of the rearfoot, the MLA and the three muscles.

## RESULTS

There were 11 subjects; six had a history of ankle or foot injury, but none had current pain. The kinematics of the the rearfoot and MLA were reasonably consistant across the subjects. For the rearfoot, plantar/dorsiflexion (PF/DF) had a range of 12deg, with three peaks, the first of PF at 12% of stance (prior to foot flat), the second of DF at 76% (after heel rise) and the third at toe off, where the PF was maximum; eversion/inversion had a range of 9deg, eversion increasing from heel contact gradually to 60% (before heel rise) and maximum inversion occurring at toe off; adduction/abduction had a range of 6deg, abduction gradually increasing to a maximum at 61% stance, and then adduction to a maximum at toe off. The MLA data was also fairly consistant across subjects: the pattern for MLA length was of a small, quick rise of approx 3mm, length remaining constant between 5% and 88%, where there was a decrease of approx 5mm; arch height varied by only 4mm over stance, with the minimum occurring at 72%, and the maximum height at toe off. In contrast, EMG patterns were variable between subjects; the most common: for PL a single peak between 60-80%, for TA a major peak at 20% and minor peak between 65-85% with activity increasing at toe off, for S a single major peak of activity between 65-85%. Although there were no significant bivariate correlations between the selected measures, cross-correlation revealed several interesting relationships between the data.

## DISCUSSION

For most of the subjects EMG activity followed a typical pattern, and though not evident from cross-correlation, inspection of the patterns of EMG activity revealed the possibility of biomechanical associations, eg typically that of activity in soleus and plantarflexion at heel rise; but also that of peroneus activity and inversion and plantarflexion at heel rise, consistant with its role both in controlling inversion as well as assisting plantarflexion. However, there were several subjects in this non-homogeneous group whose EMG patterns were atypical; eg for PL there were 2 cases in which there was notable early activity, their rearfoot patterns showed inversion at heel contact instead of eversion; one of these subjects reported that the ankle had suboptimal stability.

Although results derived from this study are interesting, it is evident that our understanding of normal and 'abnormal' dynamic control of the foot requires comprehensive analyses incorporating 3D moments of force of the individual joints, together with anatomical details of the muscles themselves, taking account of individual variation. These analyses are currently being undertaken by the authors.

## REFERENCE

Moseley L., Smith R., Hunt A. & Gant R. (1996) Three-dimensional kinematics of the rearfoot during the stance phase of walking in normal young adult males. Clin. Biomech. **11**, 39-45.

# ACHILLES TENDON FORCE DURING WALKING: MEASUREMENTS WITH THE OPTIC FIBER TECHNIQUE

**T. Laitinen, P.V. Komi and J. Lukkariniemi**
University of Jyväskylä, Department of Biology of Physical Activity, Jyväskylä, Finland

## INTRODUCTION

Accurate assessment of forces produced by individual muscles is important for the study of human movement. In addition to mathematical modelling of muscle forces, in vivo direct force measurements have been performed both with the buckle (Komi 1990) and optic fiber transducers (Komi et al 1995). The latter technique is based on the modulation of the light when the optic fiber is being compressed/bent inside the tendon. This technique was then utilized for the study of Achilles tendon forces (ATF) during walking.

## METHODS

In vivo AT loading during slow (1,1 m/s±0,01), normal (1,5 m/s±0,1) and fast (1,8 m/s±0,2) walking was measured from eight volunteers (5 males, 3 females). The sterilized optic fibre (P.M.M.A Ø 0,5 mm) was passed through the AT with a 19 gauge needle. By removing the needle the optic fiber remained in situ. The optic fiber unit included also HFBR-1414 light source and HFBR-2414 reciever. The calibration of the transducer was performed on an ankle ergometer where subject was asked to plantarflex his ankle with 10, 20, 30 and 40% of MVC. The fiber output demonstrated in general a good linear fit with the external force. After calibration the subjects walked over a 10 m force platform at specified target speeds. EMG was recorded from the tibialis anterior (TA), medial gastrocnemius (GM) and soleus (SOL) muscles.

## RESULTS

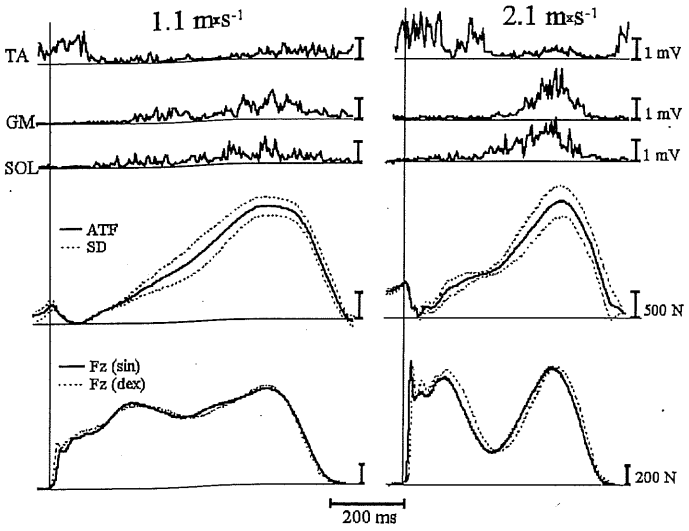



Fig 1. EMG, ATF and vertical ground reaction force of one subject (JJ) during slow (left) and fast (right) walking. The vertical line indicates heel contact. Each curve represents average of four ipsilateral contacts.

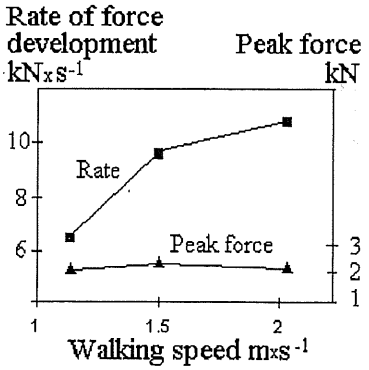



Fig 2. Rate of force development increased but peak force remained constant with increasing speed. Data taken from figure 1.

## DISCUSSION

The recorded highest peak ATFs showed great intersubject variation (1292 N±605). This was partly due to differences in the size of the tendon, but also to possible differences in the site and angular position during fiber insertion. However, in agreement with an earlier study with the buckle technique (Komi et al 1992) the peak ATFs were quite similar at all recorded speeds. Correspondingly, the rate of ATF increased with walking speed (fig 2). The responses were, however, quite individual, but the general trend was that of figure 2. Similarily, the magnitudes of the sudden release of force, which is very typical at heel contact in running (Komi 1990) was also very individual and propably dependent on the release of TA activation.

## REFERENCES

Komi P.V. (1990) Relevance of in vivo force measurements to human biomechanics. J Biomech 23,1:23-34 suppl.

Komi P.V., Fukashiro S., Järvinen M. (1992) Biomechanical loading of Achilles Tendon during normal locomotion. Clinics in Sports Medicine, 11, 521-531.

Komi P.V., Belli A., Huttunen V., Partio E. (1995) Optic fibre as a transducer for direct in vivo measurements of human tendomuscular forces. XVth Congress of the International Society of Biomechanics, Book of Abstracts, Jyväskylä, Finland.

# FREEZING AND FREEING DEGREES OF FREEDOM IN A MODEL NEURO-MUSCULO-SKELETAL SYSTEM FOR DEVELOPMENT OF LOCOMOTION

G. Taga
Department of Pure and Applied Sciences, University of Tokyo, Tokyo, Japan

## INTRODUCTION

Computational models can be used to gain insight into the complex interactions which exist between the neural and biomechanical components of the human motor control system. In earlier studies, combined models that link neural and musculo-skeletal dynamics revealed that the stable and flexible locomotion emerges from the attractor dynamics of the entire system composed of a large number of degrees of freedom (Taga et al. 1991; Taga 1995). It is still open that the dynamical systems approach that was used to study the real-time control of movements can be extended to understand the issue of the motor development (Thelen and Smith 1994), during which numbers of degrees of freedom of the system drastically change. A specific question addressed in this paper is what mechanisms produce a relatively invariant sequence of motor pattern milestones such as the newborn stepping immediately after birth, the disappearance of the newborn-stepping and the acquisition of the independent stepping. I demonstrate a model that reproduces development of posture and locomotion by freezing and freeing degrees of freedom of the neuro-musculo-skeletal system.

## METHODS

A computational model was constructed as follows. The musculo-skeletal system is composed of seven body segments in the saggital plane and attached muscles around the joints. The parameters are determined based on a literature. The neural system is composed of a rhythm generator (RG) with six neural oscillators for each joint and a posture controller (PC) with two neural units for each limb. The RG generates reciprocal activation of the antagonistic muscles, while the PC produces coactivation signals for all the muscles. The muscle torques are given by the weighted sum of the output of the RG and the PC. Sensory information on the segment displacements and ground reaction forces is sent to the RG and the PC. It is examined which way of a sequence of establishment of the aforementioned sub-systems and adjustments of parameters reproduce the sequence of the development of posture and locomotion. The neural system and the musculo-skeletal system are expressed as differential equations and numerically solved by a workstation.

## RESULTS

The qualitative changes in motor patterns during development were reproduced by changes in the structure and parameters of the model as follows. (1) When only the RG was established and the body was mechanically supported, the model generated a stepping movement which is similar to the newborn stepping. The rhythmic activities of three neural oscillators on the ipsilateral side of the RG were synchronized by the excitatory connections. "Dynamic freezing" of the neural degrees of freedom produced tightly synchronized movements of the joints. (2) When the PC was recruited and its parameters were adjusted, static posture was maintained. On the other hand, "static freezing" of the degrees of freedom of joints resulted in the disappearance of the stepping. (3) When the inhibitory connections between the RG and the PC were established, independent stepping appeared. This movement was lacking in ability to progress forward. We referred to this mechanism as "static freeing," since frozen degrees of freedom of the musculo-skeletal system by the PC were freed by the RG. (4) By increasing the input of the sensory information on the segment displacements to the RG and decreasing the output of the PC, forward walking was gradually stabilized. The simply synchronized pattern of neural activity in the RG changed to a complex pattern with each neural oscillator generating rhythmic activity asynchronously with respect to one another. By this mechanism which was referred to as "dynamic freeing," gait patterns became more similar to those of adults.

## DISCUSSION

Infants, on one hand, stabilize motor patterns that are originally unstable and variable during development of posture and locomotion, and, on the other hand, acquire dexterity of movements by increasing the complexity of motor patterns. From a dynamical point of view, these two problems that infants seem to solve are formalized as a reduction and an increase in the numbers of degrees of freedom respectively. The present model suggests that these apparently contradicting problems can be solved through the sequence of dynamic freezing, the static freezing, the static freeing and the dynamic freeing of degrees of freedom of the neuro-musculo-skeletal system. The future study is to examine whether these mechanisms can be applied to the development of other types of movements. It also remains to be elucidated how the developing neural system drives the freezing and freeing degrees of freedom by itself.

## REFERENCES

Taga G, Yamaguchi Y, Shimizu H (1991) Self-organized control of bipedal locomotion by neural oscillators in unpredictable environment. Biol Cybern 65: 147-159

Taga G (1995) A model of the neuro-musculo-skeletal system for human locomotion. I. Emergence of basic gait. Biol Cybern 73: 97-111

Thelen E, Smith LB (1994) A Dynamic Systems Approaches to the Development of Cognition and Action, MIT Press

# KINEMATCS AND DYNAMICS OF ANKLE JOINT IN NORMAL AND PATHOLOGIC GAIT

M. Galli*•, M. Crivellini*, F. Motta•, M. Tarter•, A. Vallatta*
* Dipartimento D.S.T.M., Politecnico di Milano, Milano, ITALY
* Laboratorio di Bioingegneria della Riabilitazione, Dipartimento di Bioingegneria., Politecnico di Milano, Milano, ITALY
• Ospedale Bassini, Divisione di Ortopedia Pediatrica, Cinisello Balsamo, ITALY

## INTRODUCTION

The ankle joint has a fundamental role in gait. Particularly the kinematic and dynamic analysis allows to have important informations about the mechanical behaviour of the muscles and other soft tissues associated with this joint. In subjects with Celebral Palsy (CP) the lower limb musculature works in a way completely different than in normal subjects. Rose[1] evaluated the effect of surgical lengthening of the gastrocnemius fascia on ankle joint kinematics and dynamics during gait in patients with CP. Davis[2] analysed the ankle dynamic joint stiffness ( the slope of joint-angle moment plot) and demonstrated its potential clinical utility in pathological gait.
The purpose of this paper is to document the kinematic and dynamic behaviour of ankle joint in normal and pathological gait in order to evidence some indexes that should be useful in diagnosing and in evaluating the effects of orthopaedic surgery in CP subjects.

## METHODS

Four young healthy subjects (from 20 to 26 years, mean 23.5± 2.5 years), eight healthy children (from 8 to 13 years, mean 10.2± 2.2 years) and eight CP children (from 6 to 14 years, mean 10.1± 3.4 years participated to this study. Clinical measurements of joint range of motion (ROM) and strength assessment were performed by the laboratory orthopaedist. A motion measurement system (ELITE, bts@, ITALY) provided the three-dimensional trajectories of reflective markers, a force platform (AMTI, Newton, MA) the ground reaction and the muscular activity was registered by a 6 channel telemetric EMG system. Twenty reflective markers were placed on each subject and aligned with respect to anatomic landmarks on the trunk, pelvis, thigh, shank, and foot segments. The EMG signals of lower limb musculature (mainly quadriceps, hamstrings and gastrocnemius/soleus muscle groups) were acquired using surface electrodes. The subjects were asked to walk at the self-selected speed. A minimum of four trials were collected for each subject to confirm the trial-to-trial consistency.
Joint kinematics and joint dynamics were computed based on Euler Angles and Newtonian mechanics, respectively. To evaluate the energy cost of ankle joint associated with normal and pathological gait, mechanical work indexes (mid stance work, $W_{ms}$ , and pre swing work, $W_{ps}$) were computed from the ankle joint power (Figure 1a) during the stance phase.

## RESULTS

In Figure 1b the work indexes of normals (young and children) and CP children are shown. In particular it is possible to observe that the CP children take place in a plane characterised by a pre swing work ($W_{ps}$ ) lower than that one of normals (both children and adults).

## DISCUSSION

The results concerning the work at the ankle joint during the stance phase demonstrated that children with CP tend to have diminished ability to generate energy for push-off. The relevance of these results is that the evaluation of work indexes, previously defined, are useful to 1) assess the ankle energy cost of the normal gait, 2) evaluate the different energy cost of CP subjects 3) plane the orthopaedic surgical procedure (i.e. the muscles may be in a better mechanical position relative to the ankle to generate energy) and 4) evaluate with a post-operative analysis the results of surgery itself.

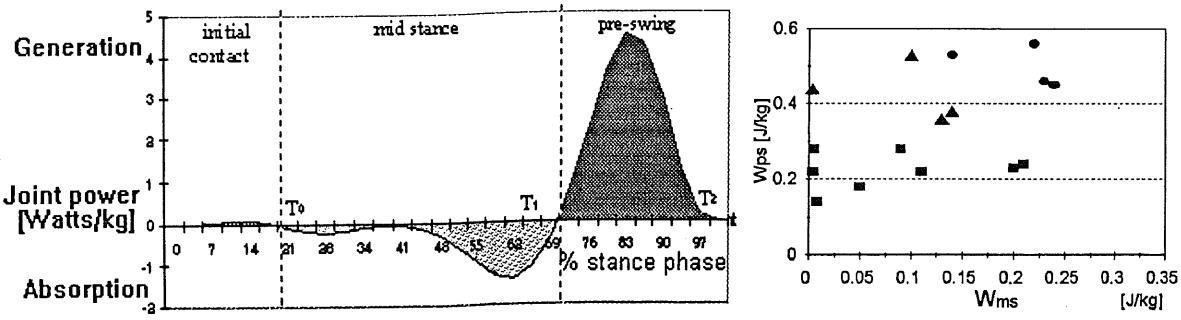



Figure 1: a) Mid-stance work ( $W_{ms}$ ) Pre-swing work ($W_{ps}$ ) b)● young healthy subjects, ▲healthy children, ■ CP children.

## REFERENCES

1. Rose S. A., DeLuca P. A. Davis R.B., Ounpuu S., Gage J.R., Kinematic and kinetic evaluation of the ankle after lengthening of the gastrocnemius fascia in children with Celebral Palsy. *Journal of Pediatric Orthopedics*, 1993; 13: 727-732.
2. Davis R.B., De Luca P A Gait characterization via dynamic joint stiffness. *Gait and posture* 1996; 4: 224-231.

# UNRESTRAINED MONITORING OF STRIDE LENGTH AND WALKING VELOCITY USING A PIEZOELECTRIC GYROSCOPE

Shinji Miyazaki
Institute for Medical and Dental Engineering, Tokyo Medical and Dental University, Tokyo, JAPAN

## INTRODUCTION

Long-term monitoring of stride length and walking velocity is considered to provide useful information for making decisions on treatment of patients with gait disabilities. The purpose of this study is to develop a device with the following design criteria: Lightweight, easy attachment, little hindrance to the natural gait pattern, sufficient memory to record for one day, and practicality in clinical use.

## METHODS

The prototype consists of a piezoelectric gyroscope and a microprocessor-based data logger. The gyroscope (ENV-05S; Murata Manufacturing Co. Ltd; 24x24x58mm, 45g) was attached to the frontal aspect of the lower part of the thigh of the unaffected limb with a rubber band to detect the angular velocity of the thigh in the sagittal plane. The angular velocity was converted to the angle with a leaky integrator. The angle was compared to a reference voltage and duration of each gait cycle (c) was converted to a digital expression. The maximum and minimum of the angle in one gait cycle ($\theta max$, $\theta min$) were detected by analog circuits and converted to digital expressions. These data were sequentially stored in the data logger which could accomodate up to 8,000 gait cycles. The whole electronic circuit was contained in a box measuring 140x80x 40mm and the total mass was 175g including two 9v alkaline batteries. Once the all data were stored, they were transferred to a personal computer via an RS232C communication line for later processing.

A simple gait model which assumed symmetric gait with two extended limbs were employed to roughly estimate the stride length (s) of one gait cycle by a trigonometric equation:

$$s = 2 \times 2 \times L \times \sin((\theta max - \theta min)/2) \quad (1)$$

where L is the limb length. The instantaneous walking velocity (v) is then calculated as:

$$v = s / c \quad (2)$$

A normal male subject walked on a straight level walkway at various velocities. The mean walking velocity (V*) was calculated as the distance of the walkway divided by the total time necessary to pass this distance. Then, the dependency of s on V* was expressed by a linear regression line. A closer estimate of stride length and instantaneous walking velocity were obtained using this regression line by solving 2nd-order equation with respect to V*. The same subject also simulated a hemiplegic gait at various velocities. Another linear regression line for hemiplegic gait was thus obtained.

In order to assess the measurement accuracy of the stride length and thus the instantaneous walking velocity, data were collected from 7 patients with A/K prostheses and 10 hemiplegic patients. They were asked to walk once at slow, medium, and fast velocities. The linear regression line of simulated hemiplegic gait was used to calculate the closer estimate of walking velocity when the mean walking velocity was below 2km/h in the case of the hemiplegic patients. Otherwise, the linear regression line of normal gait was used.

## RESULTS

Figure 1 shows the results for the A/K patients and Fig. 2 those for the hemiplegic patients. The horizontal axis represents the mean real velocity and the vertical axis the mean velocity as calculated from the data of the device.

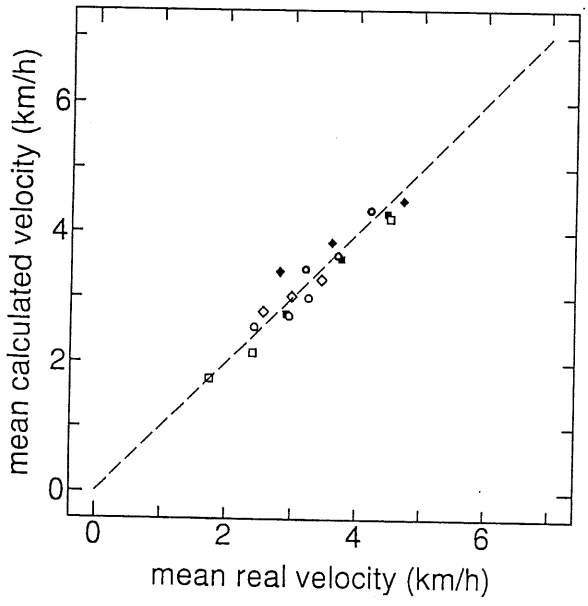



Fig. 1 Results of patients with A/K prostheses

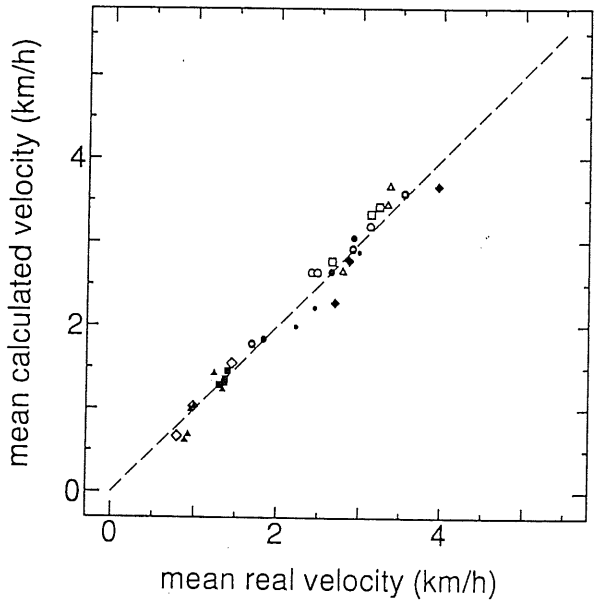



Fig. 2 Results of hemiplegic patients

# PLANTAR PRESSURE DISTRIBUTION DURING GAIT AFTER RECONSTRUCTION OF SEVERE TIBIAL FRACTURES AND LARGE SOFT-TISSUE DAMAGE

**J. Perttunen[1], H. Nieminen[2], E. Tukiainen[2], S. Asko-Seljavaara[2] and P.V. Komi[1]**
[1]Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland
[2]Department of Plastic Surgery, University Hospital of Helsinki, Helsinki, Finland

## INTRODUCTION

The advantage of microvascular free tissue transfer has changed the management of severe lower extremity trauma. The early microvascular free flap reconstruction of the leg after tibial fracture offers good chance of saving legs. The fractures with soft-tissue damages induces often impaired calf muscle function which may be associated with poor gait ability (Nieminen et al 1995). A little information is available how the patients walk after surgical interventions. Therefore, the aim of the present study was to investigate the gait symmetry after surgery and to evaluate the efficacy of treatment.

## METHODS

A total of 18 subjects with severe tibial fracture participated in this study. The most frequently used free flap in series was a latissimus dorsi flap. The delay between fracture and free flap changed from 1 week to over 4 years, and the measurements were performed 9 months - 14 years after reconstruction.

All the patients walked twice at their preferred speed along a 30 m floor. In each trial at least five contacts were collected from both feet. The walking speed was measured with photocells, and it varied between 0.56. $m*s^{-1}$and 1.52 $m*s^{-1}$. A portable data-acquisition system (Paromed-System®) was used for measuring the pressure distribution under the feet and EMG-activity from the leg muscles. The insoles with 16 pressure sensors were placed in both shoes. The surface electrodes were applied to record bilateral EMG-activities from the tibialis anterior (TA), gastocnemius lateralis (GL) and vastus medialis (VM) muscles. These EMG -results will not be reported in this abstract.

## RESULTS AND DISCUSSION

The patients walked slowly as expected. During gait the bilateral comparison showed weight bearing asymmetries in almost every patient. Figure 1 is an example of the patient who has undergone surgery and had 4 cm length difference between legs. In bilateral comparison the pressure models were quite different under the heel area. The pressures under the heel area were clearly shorter in duration and smaller in amplitude in the operated side implying that the walking had not returned back to normal. The pressure patterns of the non-operated side demonstrated often great step variability. The measurements emphasize that despite long adaptation periods after the reconstruction, most of the patients showed incomplete return to the normal pattern as shown in an example in figure 2.

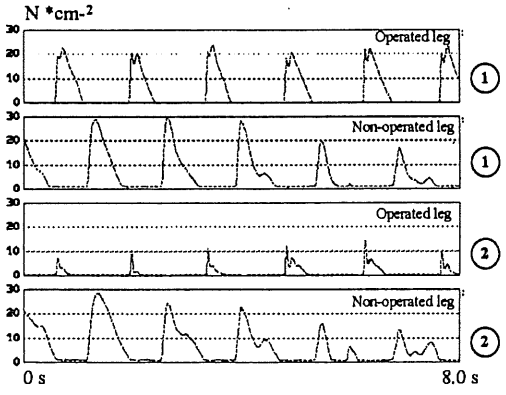



*Figure 1. Example of the pressures measured from the sensors 1 and 2 placed under the the same heel area on the operated and the non-operated legs.*

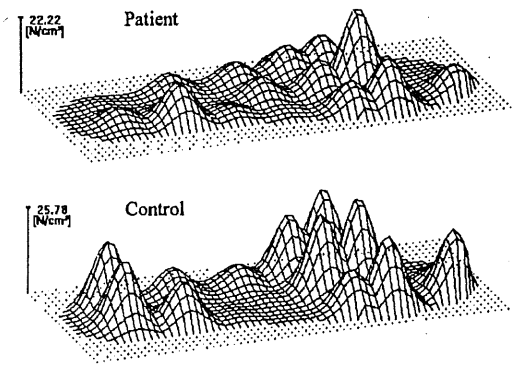



*Figure 2. Example of the foot pressure contours in a patient (after 9 months of reconstruction) and in a normal healthy control subject.*

## REFERENCE

Nieminen, H. et al: Free Flap Reconstructions of Tibial Fractures Complicated after Internal Fixation. J. Trauma. 38: 660-664, 1995.

# TRANSTIBIAL AMPUTEE GAIT ADAPTATIONS AS A RESULT OF PROSTHETIC INERTIAL MANIPULATION

Hillery, SC and Wallace, ES*

Brunel University College, Middlesex, England

*Northern Ireland Bioengineering Centre, University of Ulster, Jordanstown, N. Ireland

## INTRODUCTION

The biomechanics of transtibial (TT) amputee gait has been shown to differ from that of the normal population with the primary source of compensation for the shortfall of energy from the muscles below the knee has been shown to be at the hip (Winter and Sienko, 1988). The large hip power generation at terminal stance and preswing serves to pull the limb into swing. Dynamic Elastic Response (DER) Prostheses have been developed which, in theory, return some energy that has been absorbed elsewhere in the gait cycle. A prototype DER prosthesis was developed which would allow for the assessment of mass manipulation on the gait of amputees. This has been shown to be effective in the gait of above knee amputees, both through simulation and in practice (Boda et, 1993; Beck and Czerniecki, 1994), however, literature relating to the effect on TT amputees is scant.

## METHODS

One male TT amputee was assessed using a prototype prosthesis. Three different masses were assessed: Mass 1: 118g; Mass 2: 530g; Mass 3: 1460g. The moment of inertia was determined by pendular suspension.
A full three dimensional biomechanical analysis was carried out using a five camera VICON motion analysis system linked to an Advanced Mechanical Technologies Inc. (AMTI) force plate. The force plate is placed in the middle of a 12m walkway. Results were processed using the AMASS software and the VICON VX Clinical Manager. Five self-selected natural walking trials on the affected limb were analysed for Mass 1 and Mass 2. The subject could only complete 4 trials on Mass 3.

## RESULTS

Cadence proved to be consistent for each mass condition however it was marginally affected by the mass condition as was stance as a percentage of the gait cycle. Using the key hip moments identified by Winter (1983) the magnitude of the power generated at preswing for each of the conditions are given in Table 1 along with the temporal characteristics.

| | Cadence (steps/min) | | Stance/Swing (%cycle) | | H3 (watts/kg) | |
|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | Mean | SD |
| **Mass1** | **85** | **2** | **62.33/37.67** | **0.10** | **2.65** | **0.24** |
| **Mass2** | **87** | **5** | **60.45/39.55** | **0.06** | **2.80** | **0.30** |
| **Mass3** | **84** | **2** | **59.5/40.5** | **0.13** | **3.43** | **0.19** |

*Table 1. Effect of mass manipulation on specific temporal and kinetic variables*

## DISCUSSION

The results indicate that the cadence is minimally affected by altering the mass of the prosthesis and the amputee is able to maintain his natural walking cadence over a limited distance and number of trials by adapting his walking pattern in other aspects. As the mass increases on the prosthes is the stance time as a percentage of the gait cycle decreases. This indicates that the heavier mass, although swinging faster due to its increased inertia, is more difficult to accelerate into swing and decelerate for the next stance phase therefore the amputee needs more time to complete swing. The results confirmed the findings of others that the primary compensations occur at the hip. Evidence exists that shows the mass condition has an effect on the amount of "pull-off" power generated at the hip with more power required to lift the prosthesis into swing. These results indicate that moving the mass of the prosthesis distally effects the gait of the amputee most notably in stance time and in the hip energetics.

## References

Beck and Czerniecki (1994) A method for optimisation of above-knee prosthetic shank-foot inertial characteristics. *Gait and Posture*, 2, 75-84.

Boda,W., Findley, T., Tapp,W., Reisman, S., Lacheeer, M., Chaudry, H., Sisto, S., Cordero, D. and Winter, D.A. and Sienko, S (1988) Biomechanics of below-knee amputee gait *J. Biomechanics* **21**, 361-367.

Winter, D.A. (1983) Biomechanical motor patterns in normal human walking. *Journal of Motor Behavior*, 15, 302-330

# TRANS-TIBIAL AMPUTEE KNEE AND HIP KINEMATICS AND KINETICS FOR NATURAL AND FAST WALKING USING A CUSTOM-MADE DER PROSTHESIS

Hillery, SC^ and Wallace, ES*

^Brunel University College, Middlesex, England

*Northern Ireland Bioengineering Centre, University of Ulster, Jordanstown, N. Ireland

## INTRODUCTION

The biomechanics of trans-tibial (TT) amputee natural cadence gait was been well established for different prostheses. The adaptations which are made for fast walking have not been well documented, though it is clear from nonpathological walking that adaptations are inherent with different walking cadences (Winter, 1987). A custom made Dynamic Elastic Response (DER) prosthesis was manufactured for an active amputee to assess the affect that altering the cadence would have on the knee and hip of the subject over the self-selected natural and fast cadences.

## METHOD

A biomechanical analysis was carried out using a five camera VICON motion analysis system linked to an Advanced Mechanical Technologies Inc. (AMTI) force plate. The force plate was placed in the middle of a 12m walkway. Results were processed using the AMASS software and VICON VX Clinical Manager. Ten self-selected natural and fast walking cadences for the affected limb were collected and analysed. Cadence proved to be consistent for both the natural and fast trials.

## RESULTS

The results in the plane of progression are reported. The magnitude of the knee and hip kinematics were altered by increasing the velocity and the knee and hip joint moments and powers were affected in magnitude but not in pattern for the different cadences. The patterns at the knee were different to the normal patterns reported in the literature (Winter 1987), though similar to those of Colborne et al (1992) for TT amputees. The knee kinematic and kinetics are shown in Fig 1 and the hip kinematics and kinetics shown in Fig 2.

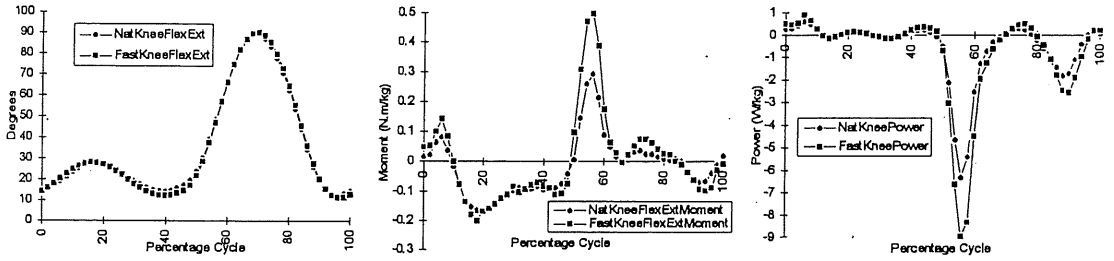



Fig 1 Knee Kinematics and Kinetics

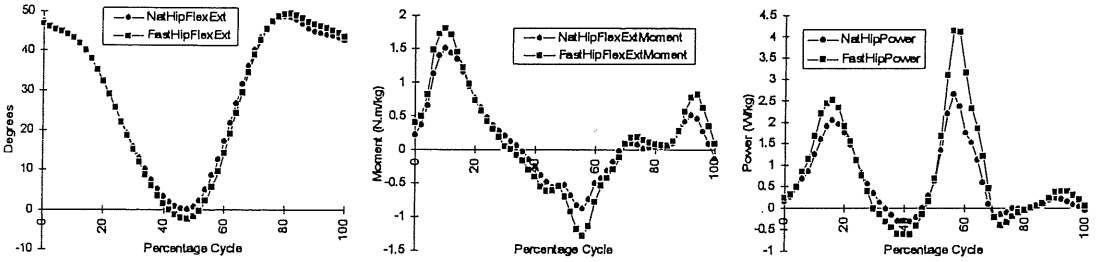



Fig 2 Hip Kinematics and Kinetics

## DISCUSSION

The increase in magnitude, though not in pattern, of the joint moments and powers which relates to the increase in velocity over the forceplate is evidence that the motor pattern adopted by the amputee is not affected by altering the speed. The primary function of the knee at both cadences appears to be that of a stabiliser, while the hip appears to compensate for the loss of function at the ankle in two ways, firstly by ensuring that adequate progression occurs and secondly that the limb is effectively accelerated into swing to increase both stride frequency and stride length. As the cadence increased, then so too do the knee and hip kinetics to ensure the cadence is maintained, though the functions of the joints were not altered..

## REFERENCES

Colborne, G.R., Naumann, S, Longmuir, P.E. and Berbrayer, D. (1992) Analysis of mechanical and metabolic factors in the gait of congenital below knee amputees: a comparison of the SACH and Seattle feet. *Am. J. Phys. Med. Rehabil.* **71**, 272-278

Winter, DA (1987) Biomechanics and Motor Control of Human Gait, University of Waterloo Press: Canada

## PAIN PREVENTIVE GAIT SYNERGY HYPOTHESIS IN LEG AMPUTEES

M. R. Pitkin
Tufts University School of Medicine, Boston, USA

### INTRODUCTION

We hypothesize that in the amputee gait, the normal gait synergy is substituted by one, where prevention of stump pain plays a dominant role and determines deviations from the norm. The pain is resulted from excessive pressures on an amputee's residuum from the prosthetic socket, and methods to decrease such pressures are analyzed. The goal of current study was to minimize stresses on transtibial amputee's residuum by selecting an appropriate prosthetic foot design. A Rolling Joint Foot (RJF) (Pitkin, 1995) was found to be adequate to that request, since as biomechanical study showed, a prosthetic gait synergy has been transferred to a more normal one.

### MODEL

During gait, an amputee tries to avoid additional stresses from the resistive moment $M_a$ in prosthetic ankle, as well as overloading the stump in axial direction. To decrease a vertical load, an amputee decreases the speed of walking and shorten the step of involved leg. To minimize the effect of resistive moment $M_a$, amputee should try to avoid bending the "rigid" ankle. He/she would rather try to position the shank vertically at the time of maximal load acceptance without natural rear-fore deflection in such a prosthetic ankle. Both components of the prosthetic gait strategy, which we call "pain preventive," are well documented (Winter & Sienko, 1988). A model of the "residuum-socket" interface has been developed to predict normal and shear stresses and to link such stresses with design of prosthesis. The model is an open chain of six weightless rods: two feet, two shanks, and two thighs. A"residuum-socket" connection is simulated by a cylinder-plunger pair with a variable shear constant. Existing anatomical joints are modeled by higher pairs with normal moment vs. angle characteristics. Prosthetic joints are simulated by their submodels which depend on specific design. Analysis of the model of the "residuum-socket" interface indicated that the Rolling Joint Foot showed less stresses on residuum compared to SACH foot, Flex-foot, and Uniaxial foot. The model was used to calculate knee angle at the moment of maximal load acceptance for norm (**Fig. 1,a**), and for involved (**b**) and uninvolved (**c**) leg in prosthetic (unilateral, BK) gait. Step lengths $L_1$, $L_2$, $L_3$, and angles $\gamma_{n1}$, were averaged from the biomechanical data available. In prosthetic step we put $\gamma_{n2}=0^0$ which resulted in $15^0$ of the knee angle (**b**). For uninvolved step (**c**), we calculated the shank angle, since the knee angle is known to be a normal-like. It resulted in advanced orientation of the uninvolved shank, in contrast with vertical position seen both in norm and in involved leg's step.

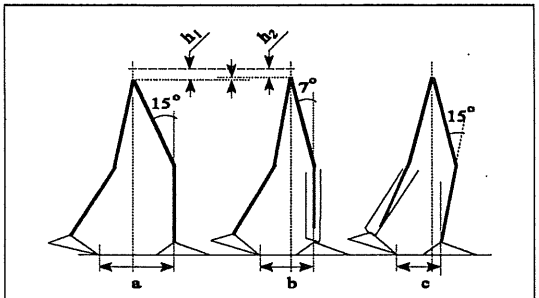



**Figure 1.** Normal (**a**) and prosthetic gait synergies.

### RESULTS

1. Theoretical analysis of the model showed that higher (normal-like) compliance of RJF, in combination with vertical position of the amputee's shank, minimizes the moment from the anterior-posterior forces of socket reactions applied to the stump.

2. A confirmation of the pain preventive synergy hypothesis can be seen in a recent gait study of four unilateral below-knee amputees, when a change in a number of correlated characteristics of the gait synergy, such as knee ROM, step and stride length and speed was found, while wearing the RJF.

### REFERENCES

Pitkin MR (1996). Synthesis of a cycloidal mechanism of the prosthetic ankle. *Prosthetics and Orthotics International*, 20:159-171.

### ACKNOWLEDGEMENTS

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

# THE PASSIVE ELASTIC MOMENT OF ACL DEFICIENT KNEE JOINT WITH AND WITHOUT A FUNCTIONAL KNEE BRACE

M. Lamontagne, P. Johnson, B. Diallo, and J.P. Desrochers
School of Human Kinetics and Dept. of Anatomy and Neurobiology, University of Ottawa, Ottawa, On. Canada K1N 6N5

## INTRODUCTION

The concept of joint stiffness and passive moments of force was mainly introduced in the early sixties in connection with another research problem; the quantification of joint stiffness in rheumatoid arthritis and other related pathologies. The elastic moments of force are the moments of force that resist joint movement due to the deformation of all tissues such as the joint capsule, tendons, and ligaments that cross the joint when the muscles are inactive [1-3].

The purpose of the study is to measure the net passive elastic moment of force of both the normal and the ACL deficient knee with and without a functional knee brace (FKB) (Legend Brace: Smith & Nephew Donjoy Inc.).

## METHODS

Seven ACL deficient male and ten males with normal knee joint volunteered for this study conducted at the University of Ottawa. The passive elastic moment function was measured by moving the lower leg in a passive mode in both supine and prone positions[4]. This passive motion of the lower leg was performed on an isokinetic device (KIN-COM 500H) at a fixed velocity of 5 °/ s from full extension to 105 degree of knee flexion. This procedure was performed with and without the FKB on both injured knee and normal subjects. For the normal knee subjects, five subjects were tested on the left leg and five with the right leg under both braced and unbraced conditions. The knee joint stiffness was calculated from both the change in the passive elastic moment of the knee joint and the viscous damping values.

## RESULTS AND DISCUSSION

For both groups of subjects the elastic moments of force were increased by about 20% at 10 degrees of flexion with the use of the functional knee brace. However, at 105 degrees of flexion, the change in elastic moment of force with the braced condition was more marked. The increase was 27.3% for the normal subjects and 33% for the ACL deficient subjects respectively. An ANOVA was used to test for significance of the elastic moment of force ($p<0.05$) between the normal and ACL-deficient groups. No significant difference was detected. However there was significant difference ($p<0.05$) between the braced and unbraced conditions. Similarly the joint stiffness (Nm/rad) values were not significantly different between groups but highly statistically different between conditions ($p<0.0001$).

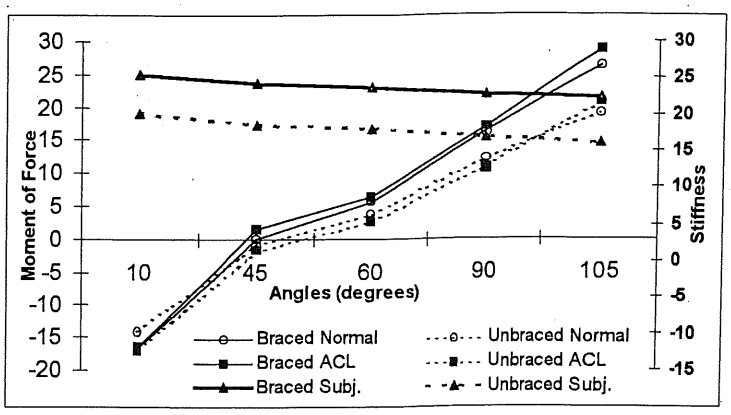



**Figure 1. Elastic Passive Moment of Force (Nm) and Joint Stiffness (Nm/rad) as Function of Knee Joint Angle.**

The functional knee brace significantly increases the passive moments of force at the knee joint. That was a greater passive moment of force at the more extreme angles (10 and 105°). Thus the functional knee brace seems to have a direct mechanical effect on the knee joint moment of force, irrespective of the degree of injury to the knee joint.

## REFERENCES

1. Yoon, Y. and Mansour, J. (1982). The Passive Elastic Moment at the Hip. *J. of Biomech.*, **15**: 905-910.
2. Mansour, J. and Audu, M. (1986) *J. of Biomech*, **19**(5):369-373.
3. McFaull, S. and M. Lamontagne. (1993) The XVth ISB Congress, Paris, p.848-849.
4. Smith, JW (1957) J. of Anatomy, **91**: 236-259

## ACKNOWLEDGEMENTS

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# EFFECTS OF USING A SEMITENDINOSUS/GRACILIS TENDON AUTOGRAFT FOR ACL RECONSTRUCTION ON MUSCLE ACTIVATION PATTERNS DURING A LANDING TASK

J. M. Manning*, J. R. Steele*, A. J. Clingeleffer$^{\Psi}$ and L. A. Pincwezski$^{\Psi}$
* University of Wollongong, Wollongong, Australia
$^{\Psi}$ Australian Institute of Musculo-skeletal Research, Crows Nest, Australia

## INTRODUCTION

Once the anterior cruciate ligament (ACL) is ruptured, reconstruction of the ligament is imperative in order to maintain an active lifestyle. A method of ACL reconstruction currently favoured involves harvesting 22 cm of the semitendinosus (ST) and gracilis (Gr) tendons, doubling them and then fixing the graft to the origin and insertion sites of the native ACL. Although proven a successful surgical procedure in terms of stabilising the knee (Pinczewski, Clingeleffer, Otto, 1995), effects of the procedure on hamstring muscle function, particularly during dynamic tasks, are unknown. Therefore, the purpose of this study was to determine the effects of harvesting the ST and Gr tendons on activation patterns of the hamstring muscles during a landing task.

## METHODS

Thirty unilateral, isolated, subacute ACL deficient patients (mean age = 24.7 ± 4.9 yrs) from the North Sydney Orthopaedic and Sports Medicine Centre were randomly assigned to an experimental and a control group. Experimental subjects had the ST-Gr tendon graft to reconstruct their ACL harvested from their uninjured limb (test limb = uninjured limb) whereas control subjects had the ST-Gr tendon graft harvested from the ACL injured limb (test limb = uninjured limb). EMG data were sampled (1000 Hz; 15-250 Hz bandwidth) from rectus femoris (RF), biceps femoris (BF) and ST during a landing task performed from a 10 cm high platform which the subjects completed pre-operatively and at 6 and 12 weeks post-operatively. Temporal characteristics of the linear envelope representing each muscle burst, relative to initial foot-ground contact in the landing task, were then analysed.

## RESULTS

Two Way Repeated Measures ANOVA results indicated a significant main effect of test week for BF duration ($F_{2,45}$ = 3.599, p = 0.036) where the mean BF duration pooled across subject groups was significantly shorter at 12 weeks post-surgery (377.7 ± 35.6 ms ($SE_M$)) compared to pre-surgery (462.7 ± 34.2 ms) and 6 weeks post-surgery (481.6 ± 36.3 ms). A similar but non-significant trend was demonstrated for ST duration.

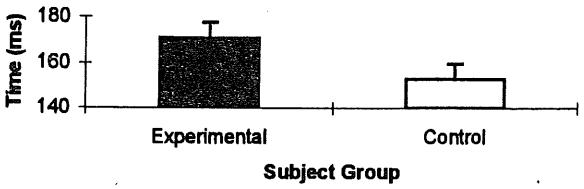



**Figure 1**: Time of onset of ST relative to foot-ground contact.

A strong trend was also evident for a main effect of subject group on timing of ST onset relative to initial foot-ground contact at landing ($F_{1,28}$ = 3.499, p = 0.073) where the experimental subjects' ST was activated earlier relative .to foot-ground contact (mean = -170.8 ± 6.9 ms) compared to the control subjects (mean = -152.7 ± 6.8 ms) (see Figure 1.). No other significant week or group effect was noted for the timing or synchronisation of the hamstring muscles relative to foot-ground contact or the quadriceps muscles, respectively.

## DISCUSSION

The reduction in BF and ST duration at 12 weeks post-surgery may reflect more efficient co-ordination of the hamstring muscles compared to pre-surgery and immediately post-surgery. As this reduction in hamstring duration was displayed by both subject groups, the change was attributed to the subjects' rehabilitation program with any negative effects of the graft harvesting procedure on hamstring duration being overcome by the rehabilitation. However, the trend for earlier activation of ST prior to foot-ground contact in landing by the experimental subjects compared to the control subjects may have been to compensate for structural changes to the muscle caused by graft harvest. It was concluded harvesting the ST and Gr tendons may influence activation patterns of the ST muscle at foot-ground in landing, changes which necessitate consideration during exercise selection for rehabilitation programs prescribed to patients who have undergone an a semitendinosus/gracilis tendon autograft for reconstruction of the ACL.

## REFERENCES

Pinczewski, L., Clingeleffer, A., Otto, D. (1995). Endoscopic ACL reconstruction using Patellar tendon versus Hamstring tendon autograft - results of pilot study. Australian Conference of Science and Medicine in Sport 1995, Hobart, Tas., 17-20 October.

# EFFECTS OF CHRONIC ACL DEFICIENCY ON MUSCLE ACTIVATION PATTERNS DURING AN ABRUPT DECELERATION TASK

J. R. Steele and J. M. M. Brown
University of Wollongong, Wollongong, Australia

## INTRODUCTION

As a primary structure controlling knee motion, rupture of the ACL can cause significant alterations in knee joint kinematics, resulting in a potentially crippling condition[1]. However, some chronic ACL deficient patients can perform dynamic locomotor tasks following complete ACL rupture without undergoing surgical repair of the ligament. Adaptations made by such functional ACL deficient patients are thought to be produced by subconscious protective mechanisms acting to avoid excessive anterior tibial displacement. Analysis of these adaptations can provide information which may be used in rehabilitating the acutely ACL injured knee. Therefore, the purpose of this study was to examine the influence of chronic ACL deficiency on muscle activation patterns displayed during an abrupt deceleration task.

## METHODS

Eleven functional chronic unilateral ACL deficient patients (mean age = 31.6 ± 7.6 yrs) and 11 control subjects (mean age = 30.4 ± 8.3 yrs) matched to the patients for age, gender, sports experience, and anthropometry and with no history of disease or trauma in either knee participated in the study. Electromyographical (EMG) and ground reaction force data were sampled (1000 Hz) as the subjects landed in single-limb stance on a KISTLER Multichannel force platform after receiving a ball thrown at chest level and then decelerating abruptly. EMG data were collected for rectus femoris, vastus lateralis, vastus medialis, gastrocnemius, biceps femoris (BF) and semimembranosus (SM) during the deceleration task for five successful trials for the subject's right and left lower limbs. Temporal characteristics of the linear envelope representing each muscle burst relative to initial foot-ground contact (IC) were then analysed for each trial. For analyses, the lower limbs of the ACL deficient and control subjects were matched for limb dominance.

## RESULTS

Two-way ANOVA results indicated no main effect for either subject group or test limb for any muscle durations or for any temporal characteristic displayed by the subjects' quadriceps or gastrocnemius muscles. However, there was a significant group x limb interaction ($F_{1,40} = 4.42$, $p = 0.042$) for the time from onset of BF activity to IC, normalised for time to peak resultant force (see Figure 1). A similar non-significant ($F_{1,40} = 3.68$, $p = 0.062$) trend was demonstrated for SM. A significant group x limb interaction ($F_{1,40} = 4.09$, $p = 0.049$) for the normalised time from peak BF activity to IC was also found (see Figure 2).

Figure 1: Time From Onset of BF to IC.

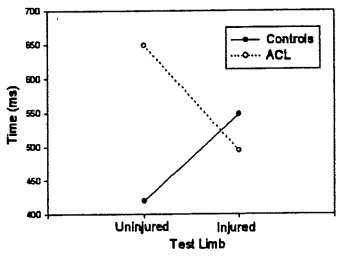
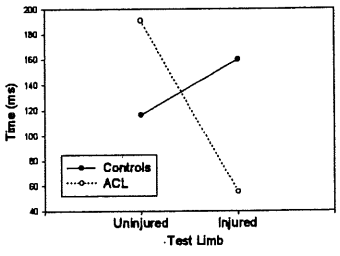



Figure 2: Time From Peak BF Activity to IC.




## DISCUSSION

Kålund *et al.*[2] and Lass *et al.*[3] claimed that ACL deficient patients activated their hamstring muscles earlier relative to foot-ground contact during inclined treadmill walking as a protective mechanism to brace the knee in readiness for the impact forces generated at heel strike. In contrast, the onset and peak hamstring muscle activity displayed by the functional chronic ACL deficient subjects in the present study were closer to the time of IC on their injured compared to their uninjured limb, a trend opposite to that demonstrated by the control subjects. Similar delays in peak hamstring muscle activity was noted for ACL deficient subjects during single-limb stance in an open pivot maneuver[1]. It was concluded this delay in hamstring muscle activation was a functional adaptation developed to enable peak muscle activity to better coincide with IC, thereby stabilising the injured limb to avoid excessive anterior tibial displacement in the absence of the ACL.

## REFERENCES

1. Shiavi, R. *et al.* (1991) Electromyography profiles of knee joint musculature during pivoting: Changes induced by anterior cruciate ligament deficiency. *J. Electromyography Kinesiology* **1**, 49-57.
2. Kålund, S. *et al.* (1990) Altered timing of hamstring muscle action in anterior cruciate ligament deficient patients. *Am. J. Sports Med.* **18**(3), 245-248.
3. Lass, P. *et al.* (1991) Muscle coordination following rupture of the anterior cruciate ligament. Electromyographic studies of 14 patients. *Acta Orthop. Scand.* **62**(1), 9-14.

# THE FEASIBILITY OF USING VIDEO STEREOGRAPHY TO PREDICT THE DISPLACEMENT OF ANTERIOR CRUCIATE LIGAMENT - CADAVERIC STUDY

[1]William W.N. Tsang, [2]K.M. Chan, [3]Arthur F.T. Mak
[1]Department of Rehabilitation Sciences, Hong Kong Polytechnic University; [2]Department of Orthopaedics and Traumatology, Chinese University of Hong Kong; [3]Rehabilitation Engineering Center, Hong Kong Polytechnic University, Hong Kong

## INTRODUCTION

The prevalence of anterior cruciate ligament (ACL) injury is increasing as more people participate in sport activities. Intensive rehabilitation is important before and after ACL reconstruction. Many controversial issues still exist regarding the types of exercise and the timing of progression in rehabilitation program (Fu et al 1992). Strain and displacement measurements have been used to study the effect of rehabilitation exercises in the ACL. However, data are still lacking on fast exercises and sport training (Beynnon et al 1995).

## AIM

This project investigated the feasibility of using video stereography (PEAK system) to predict the displacement of ACL. Cadaveric knees with intact, partial and full sectioned ACL were employed.

## METHODS

The investigation began with an accuracy test for the PEAK system. The measurement of PEAK was then compared with a roentgen stereophotogrammetric analysis (RSA) system (Friden et al 1992) both in capturing the external markers position and for calculation of displacement of ACL. The internal markers used in RSA were tantalum beads 0.8 mm in diameter, and were inserted into both the proximal and distal ends of the ACL arthroscopically to minimize skin disturbance. The external markers captured by PEAK were plate markers (Cappozzo et al 1995) and skeletal markers.

## RESULTS

The root mean square for accuracy test was 0.3 mm, which was within the accuracy required by this project. The intraclass correlation coefficient for the capturing of coordinates of the external markers by the two systems was 0.999. Twenty three conditions of knees were studied with force application of 0 N and 100 N. The correlation coefficients of the displacement, as calculated by the video stereography and RSA, were 0.93 and 0.81 (both $p<0.001$) for the plate and skeletal makers respectively.

## DISCUSSION

The video stereography commonly used to study three-dimensional and high speed movement is feasible to predict the displacement of ACL in a cadaveric model. The computer programs developed can used in clinical setting. Plate markers can provide satisfactory non-invasive prediction in cadaveric study. Future studies will be pursued in in vivo rehabilitative exercises.

## REFERENCE

BEYNNON,B.D., FLEMING,B.C., JOHNSON,R.J., NICHOLS,C.E., RENSTROM,P.A., POPE,M.H. (1995) Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo. Am J Sports Med;23(1);pp24-34.

CAPPOZZO,A., CAPPELLO,A., LEO,T., PAUL,J.P. (1995) 3-D reconstruction of human motion: Theoretical and practical aspects. Notes of a tutorial held in Jyvaskyla(Finland) prior to the XV Congress of the International Society of Biomechanics, July 2.

FRIDEN,T., RYD,L., LINDSTRAND,A. (1992) Laxity and graft fixation after reconstruction of the anterior cruciate ligament: A roentgen stereophotogrammetric analysis of 11 patients. Acta Orthop Scand;63 (1);pp80-84.

FU,F.H., WOO,S.L-Y., IRRGANG,J.J. (1992) Current concepts for rehabilitation following anterior cruciate ligament reconstruction. J Orthop Sports Phys Ther;15(6);pp270-278.

# COMPUTERIZED PENDULUM TEST FOR SPASTICITY EVALUATION IN PARAPLEGIC PATIENTS: MEASURE REPEATABILITY AND DAILY VARIABILITY

M. Ferrarin, L. Brunati, V. Samarani, R. Spadone and A. Pedotti
Centro di Bioingegneria, Fondazione Pro Juventute Don Gnocchi - Politecnico di Milano, Milano, Italy
Laboratorio di Neurofisiologia, Fondazione Pro Juventute Don Gnocchi, Milano, Italy

## INTRODUCTION

The possibility to measure objectively the spasticity level in spastic patient is of paramount importance for the evaluation of the efficacy of any antispastic treatment. In clinical environment the currently used tests (Ashworth, Pedersen) are based on scale concerning the muscle resistance to stretch movements performed manually by the physical therapist. These approaches present the disadvantages of measure's subjectivity and low repeatability. In order to overcome these problems a computerized approach to perform a Pendulum (or Wartemberg) Test for spasticity assessment in clinical context has been recently developed [1]. It makes use of a TV based motion analysis system with passive markers for leg movement detection and an electromyograph with surface electrodes for EMG recording from the main knee extensor muscles. A study on the applicability and repeatability of this technique on hemiplegic patients was previously performed [2].
In the present work the same technique was applied to paraplegic subjects with the goal to study the repeatability and daily variability. Two groups were considered: 7 normal subjects (age: 22.3 ± 1.1) and 12 paraplegic patients (age: 38 ± 13.5). In particular intra- and inter-session repeatability and intra- and inter-subject variability are here presented, together with the influence of markers' repositioning.

## METHODS

The Pendulum Test consists on the analysis of the breaking effect of the knee extensor muscles when they are stretched during the freely swinging of the lower leg while the subject is lying semi-supine with the leg hanging over the edge of a table. On the basis of the alteration of knee angle time course, some parameters can be defined to quantify this breaking effect.
In this study the experimental paradigm consisted in 4 acquisition sessions for each patient during a day, two in the morning and two in the afternoon. In each session 10 trials for each leg were performed. To analyse marker positioning influence, on healthy subject the repositioning in each session was performed only on one leg, while on the other markers were maintained glued on the leg for all the day. All relevant parameters were automatically computed through dedicated software. In this paper the attention is focused on the Relaxation Index, defined as the ratio between the amplitude of the first swing (FS) and the difference between the starting (SA) and the resting angle (RA): RI = FS / (SA - RA). Higher is the spastic level of the patient, bigger is the breaking effect of the stretch reflex during the first swing and therefore smaller is the Relaxation Index.

## RESULTS

On healthy subject the average RI was 1.76 ± 0.03, in accordance to data from literature. The average intrasession variability on all healthy, quantified with the Standard Deviation (DS) of RI within one session, was very small: 0.03 ± 0.01 (COV: 1.7%). The intersession variability resulted slightly higher: 0.04 ± 0.02 (COV: 2.27%). The comparison of intersession variability between fixed and repositioned markers didn't show any statistically significant difference. No common daily trend was found between all subjects, on the contrary 5 subjects out of 7 presented common trend on the two sides.

On paraplegic patients the average RI was 1.26 ± 0.34 showing good agreement with the clinical assessment. Both intrasession (0.074 ± 0.06; COV: 5.9%) and intersession (0.09 ± 0.05; COV: 7.1%) variability were significantly higher than normal subject. The absence of any common daily trend was found, while in 10 patients out of 12 there were common trend on the two sides.
We didn't find any correlation between lesion completeness and spasticity level as well as with spasticiy daily variability. On the other hand there was a good correlation between RI and Ashworth scale ($R^2$=0.8).

## DISCUSSION

The results obtained in this study demonstrate a good repeatability of this computerized approach to Pendulum Test both for the very low intrasession variability and for the absence of errors in useful parameters due to markers' repositioning. In paraplegic patients we found a good correlation between Relaxation Index and common clinical evaluation. Concerning test repeatability the results of this study are comparable with data obtained in previous work on hemiplegic patients. The slightly higher intersession variability found both in paraplegic and hemiplegic patients compared to healthy induce us to advise caution in using this procedure in longitudinal studies: in these cases an estimation of single patient variability range should be performed at the beginning of the study and it should be considered during the analysis of follow up.

## REFERENCE

[1] Ferrarin M., Frigo C., Osio M., Crenna P., Pedotti A., Mangoni A., (1993) Assessment of spasticity by an integrated system for the analysis of pendular motion and EMG activity, Electroenchephalography and Clinical Neurophysiology, vol.87, N.2, p.S66.
[2] Ferrarin M., Brunati L., Riabiz E., Lanna A., Pedotti A., (1996) Repeatability of computerized pendulum test for spasticity assessment in hemiplegic patients, in: Proc I Med Cong Phys Med & Rehab, p.207, Herzlya.

# THE INFLUENCE OF MEDIAL AND LATERAL ORTHOTIC WEDGES ON LOADING OF THE PLANTAR APONEUROSIS. *IN VITRO* STUDY.

**G.F. Kogler [1], F.B. Veer [1], S.E. Solomonidis [2], J.P. Paul [2]**

[1] Orthopaedic Bioengineering Research Laboratory,
Southern Illinois University School of Medicine, Springfield, IL, USA
[2] Bioengineering Unit, University of Strathclyde, Glasgow, Scotland, UK

## INTRODUCTION

The primary orthotic treatment objective for plantar fasciitis is to alleviate tension in the plantar aponeurosis during weight bearing. With this intendment, the medial wedge is one of the therapies recommended by several investigators. While many presuppose the medial wedge decreases strain in the plantar aponeurosis, to the authors' knowledge their have been no scientific studies to validate this postulate. Since the plantar aponeurosis is in tension when the foot is loaded, alterations of ground forces produced from an inclined surface should be reflected as a decrease or increase in strain. The objective of this study was to quantify strain in the plantar aponeurosis in cadaveric feet with the use of various orthotic wedge combinations.

## METHODS

A differential variable reluctance transducer (DVRT) (The Microstrain Co., Burlington, VT, USA) was surgically implanted into the plantar aponeurosis of nine fresh frozen cadaveric lower limb feet. Each specimen was mounted in an Scott electro-mechanical test machine Model CRE/500 (GCA Precision Scientific, Chicago, IL, USA) which applied a load of up to 900 N axially to the tibia. Nine test conditions were evaluated with eight different wedge (6°) combinations and a level plane served as the control (Fig. 1). Five test runs were conducted with each condition. All specimens were subjected to the nine test conditions using a 'Latin square testing sequence' to control for the effect of order. Strain, load and time data were collected and analyzed at selected load levels (225, 450, 675, 900 N).

## RESULTS

Figure 1 summarize the results of this study. A Latin square analysis of variance was performed on the data to determine variations of treatment (wedges), subject and order for both strain and time; follow-up comparisons were conducted with Duncan's multiple range test. The relative ordering of the treatments remained the same across all for load levels. In general, the control (planar) remained in a mid-position between the three lateral forefoot treatments (3,5,8) and the three medial forefoot treatments (4,6,7). It was remarkable that each of these forefoot wedge groups were statistically distinct compared to the control, with the lateral wedges decreasing strain and the medial wedges increasing strain ($P<0.05$). The rearfoot wedge conditions that did not combine use of a forefoot wedge (1,2) were not statistically different from the control ($P<0.05$).

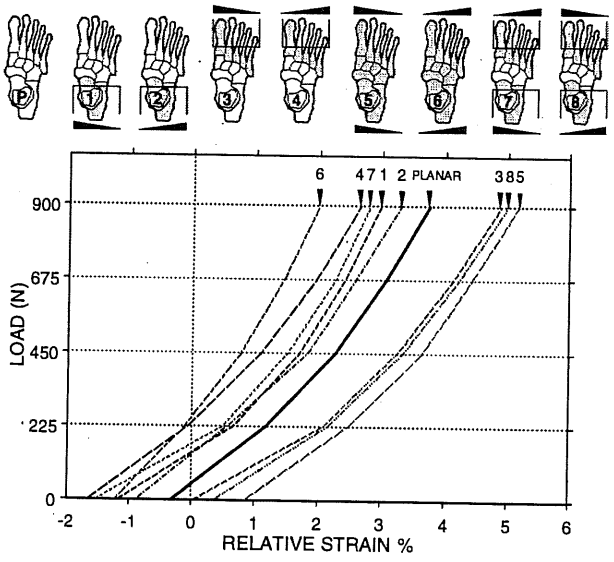



**Figure 1.** Representative load-strain curves of the plantar aponeurosis of a donor specimen limb for all nine test conditions.

## DISCUSSION AND CONCLUSIONS

Results from this study lay the foundation for fundamental orthotic principles that delineate the loading characteristics of the plantar aponeurosis and the foot with medial and lateral wedges. These data showed statistically significant ($P<0.05$) changes in plantar aponeurosis tension, noting strain decreases with lateral forefoot wedge treatments and increases with medial forefoot wedges compared to the planar situation in 'static' stance. This finding disputes the theories of several authors who have suggested that a medial wedge or post reduces strain in the plantar aponeurosis (Clancy, 1983; Marschall, 1988; Marr & Pod, 1980). We suggest that load transmission patterns in the foot are altered with an inclined plane by the shifting loads from one region of the foot to another, activating different support mechanisms. This study is the first to look at the effect orthotic wedges have on loading of the plantar aponeurosis. These data suggest that the use of a lateral forefoot wedge may be an effective plantar aponeurosis strain shielding orthosis for the treatment of plantar fasciitis.

## REFERENCES

Clancy WG (1983) Tendonitis and plantar fasciitis in runners. Orthopedics 187:176-187.
Marr SJ, Pod FA (1980) The use of heel posting orthotic techniques for relief of heel pain. Arch.Orthop.Traumat.Surg.96,73-74
Marshall P (1988) The rehabilitation of overuse injuries in athletes and dancers. Clinics in Sports Medicine, 7(1) 175-191.

# REPRODUCIBILITY OF NON-CYCLIC UPPER EXTREMITY MOVEMENT IN CLINICAL ARM FUNCTION EVALUATION.

**R. Kastein, H.A.M. Seelen, Y.J.M. Potten**
Institute for Rehabilitation Research, Hoensbroek, The Netherlands

## INTRODUCTION

In tetraplegic patients arm and hand function is impaired, depending on the level and the completeness of the spinal cord injury (SCI). This leads to problems of shoulder, elbow and wrist movement. During rehabilitation patients are extensively trained in compensating function loss involving residual capacities. However, the means to objectively quantify muscle involvement during arm movement are lacking. Aim of the present research was to develop standardized upper extremity tasks to be used in the clinical evaluation of residual muscle function in tetraparetic patients. Reproducibility of arm movements and patterns of upper extremity muscle use during a set of standardized tasks were evaluated in non-SCI subjects.

## METHODS

A group of 15 healthy, non-SCI subjects (12 males, 3 females; mean age 21.5 y (±2.4); height range 170-190 cm) participated. Subjects were asked to perform multiple unilateral, semi-circular arm movements in the frontal plane (task A), the transversal plane (task B) and a plane at an angle of 45° relative to the one in task A (task C). Participants were seated behind a table. A plastic tube (2 cm $\phi$), bent in a semi-circle, was placed in one of the aforementioned planes in front of the subject. Dimensions of the arc was adapted to each individual's reach according to a predefined protocol. Subjects were asked to move a ring over the tube from one end to the other (phase I) and back (phase II) with their right hand, thus defining the course of the arm movement. Muscle activity of the deltoideus pars clavicularis (DEC) pars acromialis (DEA) and pars spinata (DES), biceps caput longum (BIL) and breve (BIB), triceps caput longum (TRLO) and laterale (TRLA), brachioradialis (BR), extensor carpi radialis (ECR) and ulnaris (ECU), flexor carpi radialis (FCR) and ulnaris (FCU), trapezius pars descendens (TRD), levator scapulae (LS), supraspinatus (SS), infraspinatus (IS), rhomboidei (RH), teres minor (TM), latissimus dorsi (LD), serratus anterior (SA) and pectoralis major pars sternocostalis (PMS) was measured using surface EMG. Operation of buttons, placed at both ends of the arc, indicated the beginning and end of a movement phase. Button status as well as EMG signals were computer recorded. EMG signals were full-wave rectified, low-pass filtered and standardized in the time domain and to the percentage mean EMG (mEMG) per trial. Per subject ensemble averages of these signals as well as 99% confidence intervals levels ($CI_{99}$) were calculated. This was also done for the data of the whole group. Indicators of signal reproducibility, i.e. intra-class correlation coefficients (ICC) and RMS values of $CI_{99}$ signals ($RMS_c$), were computed .

## RESULTS

Group ICC's and $RMS_c$'s (in % mEMG) for both phases of task A are given in the table below.

| Phase | I | | II | | | I | | II | |
|---|---|---|---|---|---|---|---|---|---|
| Muscle | ICC | $RMS_c$ | ICC | $RMS_c$ | Muscle | ICC | $RMS_c$ | ICC | $RMS_c$ |
| DEC | .96 | 6.2 | .97 | 6.2 | TRD | .95 | 5.8 | .97 | 5.6 |
| DEA | .97 | 6.0 | .98 | 5.8 | SS | .99 | 6.0 | .97 | 5.0 |
| DES | .96 | 6.2 | .95 | 6.6 | IS | .86 | 6.6 | .94 | 6.2 |
| BIL | .91 | 8.0 | .94 | 7.4 | LS | .97 | 6.0 | .95 | 4.8 |
| BIB | .89 | 6.6 | .97 | 6.2 | RH | .95 | 7.4 | .93 | 7.0 |
| TRLO | .92 | 6.8 | .97 | 5.8 | TM | .94 | 6.2 | .99 | 5.8 |
| TRLA | .94 | 8.0 | .98 | 6.2 | LD | .95 | 6.8 | .99 | 6.2 |
| BR | .83 | 8.0 | .80 | 7.4 | PMS | .96 | 8.8 | .87 | 12.6 |
| ECR | .85 | 7.2 | .64 | 6.6 | SA | .97 | 5.6 | .99 | 5.8 |
| ECU | .87 | 6.4 | .98 | 7.2 | FCR | .92 | 10.2 | .89 | 10.4 |
| | | | | | FCU | .95 | 7.6 | .84 | 8.2 |

## DISCUSSION

In general within-group reproducibility of muscle activation patterns or profiles during task performance was high as indicated by the ICC values and the low $RMS_c$ values. However, it was observed that reproducibility was less in some muscles, e.g. the ECR and BR during phase 2. The next step in our research will be (1) to investigate the reproducibility of the tasks in groups of tetraplegic patients with different levels of SCI and sensorimotor function loss and (2) to correlate specific task parameters (e.g. movement extent or height) with (compensatory) patterns of muscle activation and reduced functional arm and hand movement.

# ANALYSIS OF FRACTAL FLUCTUATION IN THE STRIDE INTERVAL SERIES OF HUAMN WALKING

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

## PURPOSE

Hausdorff et al.(1995,1996) recently reported long-range self-similar correlations(fractal correlations) extending over hundreds of strides in human overground walkings, although the mechanism of this remained unclear. It is expected that we can find a new knowledge about the control mechanism of human locomotion.

The purpose of this study, therefore, was threefold; 1) to determine whether fractal correlation is also observed during under the condition of "constant velocity". It means "stride length" sequence has fractal correlation; 2) to explain the variance of fractal fluctuation of the stride length by the integral EMG fluctuation; 3) to present a hypothesis about the phenomenon.

## METHODS

Nine healthy men, who provided informed written consent, performed treadmill walking by self-selected velocity. The stride interval(time interval from right heel contact to the next right heel contact) was measured, and was analyzed by the coarse graining spectral analysis(CGSA), developed by Yamamoto and Hughson(1993).

About five of the nine subjects, electromyogram(EMG) data were gathered for the major muscles of the lower extremities, and were integrated during step duration(time from right heel contact to the next left heel contact, and from left heel contact to the next right heel contact). Both of Pearson's correlation coefficient and the size of integral EMG were used to determine the variance of stride interval.

## RESULT AND DISCUSSION

We first found that the similar type of long-range correlations are observed in human treadmill walking(Fig). In treadmill walking, the average walking velocity was constant, so the parameter "time interval" simply means "stride length" in this type of walking.

Secondly, we extracted some muscles which statistically contribute to making stride length. M.biceps femoris, m. gastrocnemius, and m.soleus are detected.

We then hypothesized that the long-range correlation were derived from step-by-step mappings of muscular activities.

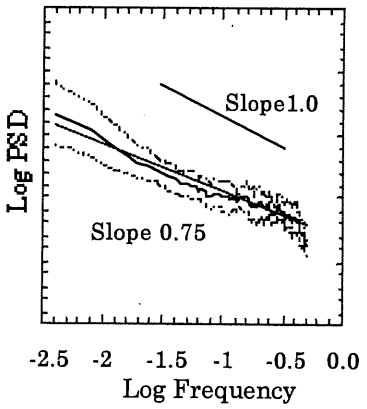



Figure The ensemble averaged power spectra of stride interval series in logarithmic scale for all subjects.

## REFERENCES

Hausdorff, J. M., et al.(1996) Fractal dynamics of human gait: stability of long-range correlations in stride interval fluctuations. J. Appl. Physiol., 80(5): 1448-1457.
Hausdorff, J. M., et al.(1995) Is walking a random walk? Evidence for long-range correlations in stride interval of human gait. J. Appl. Physiol., 78(1): 349-358.
Yamamoto, Y., and Hughson, R. L. (1993) Extracting fractal components from time series. Physica 68D:250-264.

# ANALYSIS OF MOVEMENT AT LOWER LIMB DURING UPHILL AND DOWNHILL WALKING

K. Takahashi 1, T. Isaka 2, S. Noge 1, F. Kuki 3 and T. Mori. 1
1 Kanagawa Institute of technology, Kanagawa, Japan
2 Ritsumeikan University, Shiga, Japan.
3 Hoshi University, Tokyo, Japan.

## INTRODUCTION

Recently, many people of middle and advanced age are joining in the mountain-climbing for health and enjoying. Many walking researches using the biomechanical approach have handled with the flat walking while uphill or downhill walking in human are a few case. Understanding of the mechanism of lower limb movement at uphill and downhill walking is needed for safe and joyful mountain-climbing. The purpose of this study was to clarify the difference of mechanism of lower limb movement between uphill walking and downhill walking.

## METHODS

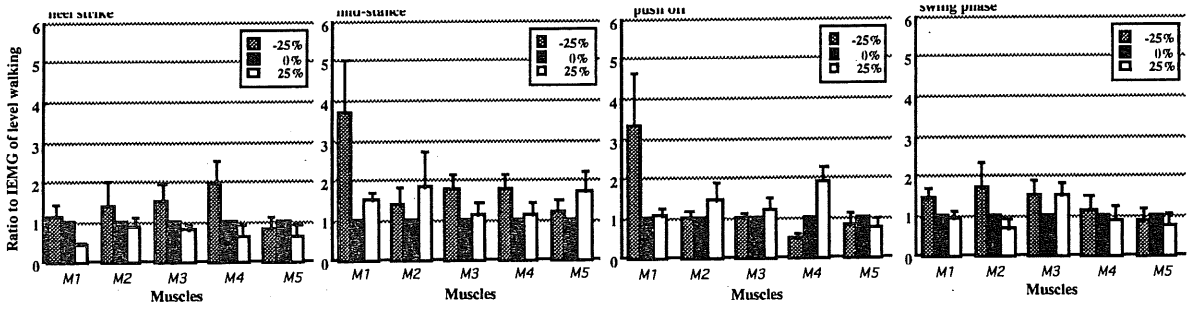
Four healthy male college students participated in this study. The angles of hip, knee and ankle joint during level, downhill and uphill walking were measured by using a electrogoniometer (P & G Co., Ltd.) sampled at 50 Hz. In order to compare muscle activity among each slope walking, EMG activities of rectus femoris, biceps femoris, tibialis anterior, gastrocunemius and gluteus maximus were measured by using surface electrodes sampled at 1 kHz. The walking phases, such as heel strike, mid-stance, push-off and swing phases, were defined with pressure sensors attached on heel and toe. These obtained data were sent to personal computer through the A/D converter. All walking were performed on treadmill with a slope gradient from +25% to -25%. The walking velocity was adjusted by each subject, which meant a preferred velocity in a given slope. The used walking velocity in level was 80.0±3.5 m/min, and in uphill decreased in proportion to up slope gradient (ex. 25%: 56±4m/min) and increased in proportion down slope gradient (ex. -25%: 99±4m/min).

## RESULTS AND DISCUSSION

In comparing with relative duration of each walking phase in normalized walking cycle, the push off phase increased and the mid-stance phase decreased with proportion to uphill slope gradient, in downhill walking these were appeared to the contrary tendency. It is considered that the longer push off phase of the uphill waking was useful for lifting up their body weights, and the extended mid-stance phase in the down hill walking worked to prevent the fall of the body against the downhill.

The patterns of hip and ankle joint angles during a normalized walking cycle were similar between up and down slope walking. However, the difference of the pattern of knee angle between uphill and downhill slope walking was obtained in mid-stance phase. The knee and hip joints in uphill slope extended simultaneously during mid-stance phase, on the other hand, in downhill slope the knee joint was keeping at a given angle and hip joint extended during same phase.

Figure 1 showed the example of the relative IEMG during each phase. The values were normalized to IEMG of level walking. In the uphill walking, the ratio of IEMG increased in the rectus femoris and biceps femoris during mid-stance phase and gastrocnemius during push off phase. In the downhill walking, the ratio of IEMG in rectus femoris, tibialis anterior and gastrocnemius increased drastically during mid-stance phase. In the rectus femoris, the ratio was still higher until push off phase. We considered that lifting up their body weight was loaded in uphill walking, knee extensor and planter flexor muscles were strongly involved. On the other hand, in the downhill walking the body position leaned backward with brace legs was performed to prevent the fall of body and reduce the downward acceleration, therefore, the rectus femoris was activated eccentrically during mid-stance to push-off phase and the stronger co-contraction between tibialis anterior and gastrocmemius was appeared during mid-stance phase.




Note) M1; rectus femoris, M2;biceps femoris, M3;tibialis anterior, M4;gastrocnemius, M5;glutei

Fig.1 The ratio of the IEMG in uphill/downhill walking to the IEMG of level walking at each phase.

# MECHANICAL ENERGY EXPENDITURE AND EFFECTIVENESS OF MECHANICAL ENERGY UTILIZATION OF WALKING FOR THE ELDERLY AND THE YOUNG MALES

**H. Okada[1)], M. Ae[2)], and N. Fujii[2)]**
**1) Doctoral Program in Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan**
**2) Institute of Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan**

## INTRODUCTION

Walking is one of the most basic movements among daily living activities and its ability may decide the independence of the elderly people in daily life. Locomotion such as walking and the other body movements are affected by the effectiveness of mechanical energy utilization as well as mechanical energy expenditure.

The purpose of this study was to compare the mechanical energy expenditure and effectiveness of mechanical energy utilization of walking for the elderly males with the young males.

## METHODS

The subjects were thirty-one Japanese healthy elderly males(73.4 ± 6.4 yr.) and twelve Japanese healthy young males(24.5 ± 1.9 yr.). The subjects instructed to walk in two self-selected speeds. One was the speed at which they walked at their daily life (Normal walk; Elderly:1.07 ± 0.19m/s, Young:1.61 ± 0.08m/s). Another one was faster speed than the normal walk (Fast walk; Elderly:1.37 ± 0.21m/s, Young:1.87 ± 0.09m/s). They were videotaped with a high speed VTR camera and ground reaction forces on the right foot were measured with a force platform. Thirteen body landmarks were digitized at 50Hz to obtain two dimensional coordinate data and these coordinate data were filtered using a low-pass butterworth digital filter. Subjects' body were modeled as a link segment model which was constituted by the foot, shank, thigh, hand, forearm, upper arm, head and neck, and torso. According to inverse dynamics method, the joint torques, joint torque powers, and joint mechanical works at the ankle, knee, hip, wrist, elbow, shoulder and neck were calculated. Mechanical energy expenditure of joint torque sources(MEE) and effectiveness index which indicates effectiveness of mechanical energy utilization(EI) were calculated by equation (1) and (2).

$$MEE = \int_{T1}^{T2} \left[ \sum_{i=1}^{N} \left| Mi(\omega_{id} - \omega_{ip}) \right| \right] dt \qquad (1)$$ (Aleshinsky, 1986)

$$EI = \frac{d}{MEE / m} \qquad (2)$$ (Ae and Fujii, 1996)

where T1= start of walking cycle, T2 = end of walking cycle, N = number of joints, Mi = joint torque at the ith joint (torque acted on the distal segment by the proximal segment), $\omega_{id}$ = angular velocity of the distal segment at the ith joint, $\omega_{ip}$= angular velocity of the proximal segment at the ith joint, d = walking distance(stride length), and m = body mass.

## RESULTS AND DISCUSSION

Figure1 shows positive and negative joint mechanical works per body mass during walking in two self-selected speeds. Both total positive and negative work per body mass for the elderly were smaller than those for the young. But these differences were significant only in the normal walk. MEE per body mass was significantly different between two groups only in the normal walk. EI for the elderly was slightly lower than that for the young in the normal walk. With increase in walking speed, EI was almost constant for the young while decreased for the elderly. Then in the fast walk EI for the elderly was significantly lower than the young. Stride length changed from 0.84 to 0.91m for the young and from 0.62 to 0.68m for the elderly with increase in walking speed. From these results, it is likely that the elderly need more additional MEE to extend stride length and may walk in less effectiveness than the young in the fast walk.

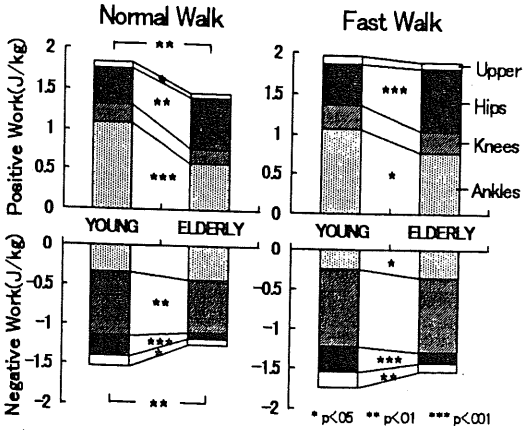



Figure1 Joint mechanical works for the elderly and the young males during walking in two self-selected speeds

## REFERENCE

AE, M. and Fujii, N. (1996) Effectiveness of mechanical energy utilization and its index in human movement. *Bull. Inst. Health & Sport Sci., Univ. of Tsukuba*, 19, 127-137.

Aleshinsky, S.Y.(1986) An energy 'sources' and 'fractions' approach to the mechanical energy expenditure problem - II . Movement of the multi-link chain model. *J. Biomechanics*, 19, 295-300.

# NORMALIZATION AND QUANTITATIVE ANALYSIS OF EMG SIGNAL ON LOWER LIMB MUSCLES IN NORMAL WALKING AS COMPARED WITH WALKING UP / DOWN STAIRS

Y. Nishijima 1) M. Yoshizawa2) S. Kumagai 2) K. Shimizu 3) R. Ogawa4) O. Kameyama 4)
1) Osaka Kyouiku University, Osaka, Japan
2) Fukui University, Fukui, Japan
3) Nara Sangyou University, Nara, Japan
4) Kansai Medical University, Osaka, Japan

## INTRODUCTION

Some researchers (Yang et al. 1984, Arsenault et al. 1986) have reported electromyographic (EMG) signal normalization to isometric contractions, in order to provide information on the degree of muscle activation required to walk. Moreover, it appears that Townsend et al. (1978) and Andriacchi et al. (1980) have published EMG data on walking up and down stairs ; they reported on some qualitative aspects of muscular activity for body support and locomotion. The purpose of the present study was to clarify the quantitative differences between level walking (Walking) and walking up (Up) / down (Down) stairs through the normalization of EMG signal.

## METHODS

Thirty male subjects with an average age of 19. 3 (SD=0. 9) years participated in the study as volunteers. The subjects were required to walk on a walkway and to Up/Down stairs with a consistent cadence (120 steps/min) utilizing a metronome. Muscles tested were tibialis anterior (Ta), gastrocnemius (Ga), soleus (So), vastus medialis (Va), rectus femoris (Rf), biceps femoris (Bf), gluteus medius (Gme), gluteus maximus (Gma) from the dominant lower limb. The digitized data of every 10 EMGs of every subject was processed in two ways. First, the real time (ms) of the gait cycle was normalized (0-100 per cent) so that poling of the data across subjects would be feasible. Secondly, the signal of the EMG , expressed in mV, was normalized to 100 percentage of a maximum isometric contraction (MIC: normally otherwise referred to as a maximum voluntary contraction; MVC) that had been obtained for each muscle. The sampling and normalization procedures allowed for the cumulation of mean and standard deviation values for each of the sample points, obtained in a cross-subject manner.

## RESULTS AND DISCUSSION

In each movement, the peak activity of normalized-EMG (N-EMG) profile of Ga and So muscles exceeded 70%MIC level , but Rf and Bf showed slightly lower activity at approximately 70 %MIC level.

No large interactions variation in the N-EMG profile patterns of any of the muscles except Ta and Bf were observed during the Up and the Walking. Every average N-%MIC (mean amplitude of one stride) of 8 muscle tested during the Up was significantly higher (Vm, Bf: $P < 0.05$, the others:$P < 0.01$) than that of the Walking.

The peak activity of Ga during the Down was around heel-strike, however, it was observed before the toe-off during the Walking. Every N-%MIC of Ga, Bf and Gma during the Down was significantly higher (Bf: $P < 0.05$, Ga, Gma: $P < 0.01$) than that of the Walking.

Results indicate that the N-EMG profile patterns during the Up and the Walking seems to be similar to each other, thought each muscular load exerted to lower limbs varies. The changing of N-EMG profile pattern and the increase of the N-%MIC of Ga during the Down, in comparison with the patterns observed during the walking, might be caused by stabilization of the ankle joint around the toe-on movement.

## REFERENCES

Andriacchi T., Andersson G., Fermier R. (1980) A study of lower limb mechanics during stair climbing. J. Bone Joint Surg., 62-A;749-757.

Arsenault A. B., Winter D. A., Marteniuk R. G. (1986) Is there a "normal" profile of EMG activity in gait ? Med. Biol. Eng. Comput. 24, 337-343

Townsend M. A., Shiavi R., Lainhart S. P., Caylor J. (1978) Variability in synergy patterns of leg muscles durig climbing, descending and level walking of highly-trained athletes and normal males. Electromyogr. Clin. Neurophysiol. 18, 69-80

Yang J. F., Winter D. A. (1984) EMG amplitude normalization methods: on improving its sensitivity as a diagnostic tool in gait analysis. Arch. Phys. Med. Rehabil. 65, 517- 521

# CHARACTERISTICS OF RECOGNITION BETWEEN AGED PERSONS AND YOUNG ADULTS ON DOWN STAIR WALKING : BY USING AN EYE-CAMERA SYSTEM

Keiji Tamura, Kazuhiko Watanabe and Kaoru Yamanobe

Laboratory of Physiology and Sports Biomechanics, Faculty of Education, Hiroshima University, Hiroshima JAPAN

## INTRODUCTION

The purpose of this study was to investigate the characteristics of eye movement of aged persons when walking down stairs.

## METHODS

An eye-camera system was used for analyzing the visual information when walking down stairs. The subjects were 5 healthy aged persons (age = 68.6±1.7) and 5 university students (age = 22.4±0.9) who were informed previously of the purpose of the experiment and the experimental conditions. The 27 stairs were located at JR Saijo railway station. The dimensions of the steps were 35cm x 300cm of width and 16.5cm in height. The total length of the stair way was 9.89m and 4.45m in height. The subjects were asked to walk at their own speed.

## RESULTS AND DISCUSSION

The following results were obtained;

1 The comparison of velocity between aged persons and young adults : The velocity of aged person when walking down stairs was 0.55±0.09m/s and 0.68±0.05m/s in young adults.(see Fig.1)

2 In both groups, the results from the eye-camera system produced the "saccadic " eye movement whilst walking down stairs.

3 The area and the length of eye-movement during walking was compared between aged persons and young adults. The area of eye movement of young adults was 3948.19±342.20cm². The area of eye movement of aged persons was 2215.68±1388.13cm².(see Fig.2) In young adults, the larger area of eye-movement and also the length was recorded. These parameters will be useful to detect the characteristics of eye movement in aged persons and young adults.

4 There are some difficulties in using the eye-camera system to analyze the eye movement in an actual situation, but still this system will be useful to study the eye movement of aged persons whilst walking.




**Fig.1**

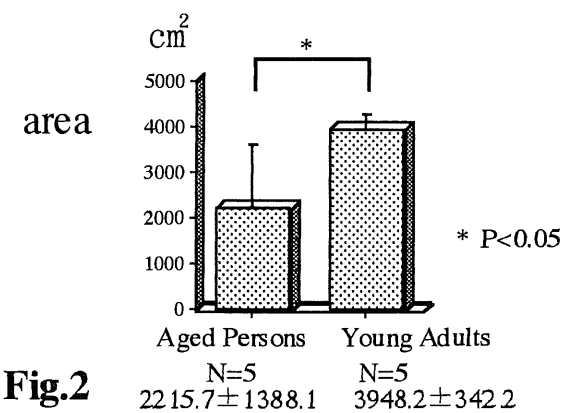



**Fig.2**

# IMPACT ACCELERATION OF LOWER EXTREMITY IN STEPPING DOWN STAIRS

Hiroshi KARASUNO and Kazuyoshi SAKAMOTO

Dept. of Communications and Systems ,
The University of Electro-Communications , Tokyo ,Japan

## INTRODUCTION

The absorption function for lower extremities in jumping has been investigated. The function was called as absorbability by Nagata(1979). The absorbability had been evaluated by the impact force at jumping to step. In the research, the jumping stands with height over 40cm were used. Such stands were not popular in daily living. Especially for the advanced aged and for the persons injured lower extremities it was difficult to measure the absorbability by the stand. In the study, the lower stairs with height of 10 to 20cm were employed to evaluate the absorbability by measurement of the acceleration of lower extremities at stepping down stair. The absorbabilities in the young subjects and the elderly ones are evaluated.

## METHODS

The young subjects(YS) were 12 healthy adults with mean age of 25.4 years, while the elderly subjects(ES) were 3 adults with mean age of 68.0 years. The movement of right lower extremity was measured. Four accelerometers(AS20B, Kyouwa) were tightly attached on skin by elastic tape; Namely, four locations on the instep(ACC1), on the heel(ACC2), on the shin of 5cm above ankle joint(ACC3), on the anterior thigh of 5cm above knee joint(ACC4) were applied. Tow foot switches were attached below both toe and heel of the right foot. They were utilized to detect the time when the toe and the heel contacted to the face of lower stair in one step cycle. One step cycle was divided into two periods. Toe contact period (TCP) was defined as the period which the contact point moved from toe to heel on the stair. Heel contact period(HCP) was defined as the period after TCP till the disappearance of the accelerations. The impact accelerations of four accelerometers were measured on the lower stair with height of 10 and of 20cm. The subjects were instructed to step down from the toe. The maximum value of the acceleration were evaluated for the both periods of TCP and HCP.

## RESULTS AND DISCUSSION

The impact acceleration was defined as the acceleration with the opposite direction to the motion which main direction meant the proximal direction of lower extremity. The maximum value of the acceleration at ACC2 in both TCP and HCP was shown in the figure. The white bar represented the value for YS and the black bar the value for ES. The maximum value of the acceleration at ACC2 on the stairs height of 20cm for YS was larger than that for ES in TCP. The relationship in TCP was reversed in HCP. The reversible relationship was a marked characteristic for ES. The similar tendency was observed in the experimental condition that the stair with height of 10 cm was used.

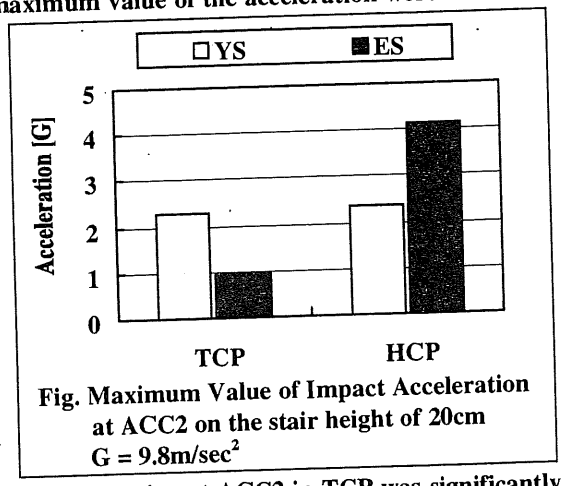



Fig. Maximum Value of Impact Acceleration at ACC2 on the stair height of 20cm
$G = 9.8m/sec^2$

In the experiment for YS, the maximum value of the acceleration at ACC2 in TCP was significantly larger than that for other positions, while that at ACC2 in HCP was significantly larger than the value for other positions. The maximum value of the acceleration at ACC2 had significantly influence on the maximum value at both ACC3 and ACC4 in HCP. The results of the experiment for both YS and ES suggested that the decrease of the impact acceleration at ACC2 in HCP was one of the role for absorbability. As regards to the significance, it was considered that the absorbability for ES declined as compared with that for YS. During the period of TCP, the adaptive dorsal flexion movement and the adaptive muscular activity around ankle joint were important function to reduce the impact acceleration.

## CONCLUSION

In the study, the impact acceleration at four locations of the lower extremity were measured in the experimental condition with the lower stairs, so that the absorbability for both YS and ES were investigated. The different for between YS and ES was the maximum value of the acceleration at ACC2(i.e., heel) in HCP. As for the results, the it was found that the absorbability of the lower extremity for ES was lower than that for YS. The property of the absorbability could been evaluated by the measurement of the impact acceleration on every part of the lower extremity in stepping down lower stairs.

## REFERENCES

1) A, Nagata et al. : Measurement and Evaluation of Absorb- ability. Jap. J. Phys. Educ. 24(1), 51-58, 1979.
2) Maeda et al. : Lower Extremity For Shock Attenuation During Landing On One Leg Jap. J. Phys. Fitness Sports Med. 43,219-227,1994.

# Biomechanical Implications of Torsional Deformities in Children Gait

J-S Her[1], Y-L Chou[1], F-C Su[1], C-J Lin[1,2], W-K Huang[1]
[1]Institute of Biomedical Engineering, National Cheng Kung University, Tainan, Taiwan
[2]National Cheng Kung University Medical Center Dept. of Orthopedic Surgery

## Introduction

In order to investigate the pathomechanism of abnormal gait, it is necessary to study the normal gait. The norms from the study of normal population provide us the references for deciding what are normal gaits and what are not[1][2]. Some of the observed gaits of patients, though considered as abnormal, are only compensatory mechanisms of other neglected problems[3]. The purpose of this study was to examine the relationship between foot progression angle (FPA) and compensatory mechanism in Taiwanese children during the maturation of gait. The gait data of this study provided us information to realize the corresponding factors of gait parameters. The relationship between mechanical energy, computed by the time integral of joint power during different gait phase, and toe in/out was the major concern of this study.

## Methods

Forty-nine healthy children between five and six years of age were included in this research. The Export Vision motion analysis system (Motion Analysis Corp. Santa Rosa, CA) was used to collect marker trajectories. Helen Hayes marker set was used. Five trials were sampled for each subject at 60 Hz. In addition, three Kistler force plates were used to collect ground reaction forces at 1000 Hz sample rate, simultaneously. OrthoTrak II software was used for kinematic and kinetic evaluation of the lower extremities in three dimensions. Three normalized (100%) gait cycles for each subject were averaged for all gait parameters. Independent sample t test was used to analyze the correlation of physical characteristics as corresponding factors and gait parameters as covariant factors to torsional deformities

## Results

The mean age of forty-nine children is 5.4($\pm$0.7) years and the mean height is 120.2($\pm$4.5) cm. The FPA is 9.154($\pm$7.517) degrees in average (toe out). For realizing the role of FPA, the children were divided into two groups according to the degree of FPA. The group I consisted of twenty-eight children with FPA under 9.154 degrees in average of 3.392($\pm$2.77) degrees. The group II consisted of twenty-one children with FPA above 9.154 degrees in average of 16.837($\pm$4.099) degrees. The time integral of time summation of joint power was significantly different between group I and group II, during the periods of initial contact loading response and mid stance at knee($p<.05$) and during the periods of mid stance and terminal stance at hip ($p<.05$).There was no significant difference at ankle.

## Discussion

Our results reveal that there are no significant differences between group I and group II at ankle during most of stance phase when the ankle absorbs energy. Only in the terminal stance phase the ankle generate large energy. There are two opposite patterns with statistical significance at knee during most of the stance phase between group I and group II. In group I, the knee is the primary power generator during the early stance phase but it switches to be power absorber during the mid stance phase. In group II, in contrast, the knee is the power absorber during the early stance phase but it switches to be power generator during the mid stance phase. The absorption and generation of power at knee in group I is more closer to normal walk. It is noted that the generation of power at knee during the early stance phase in group II disappears obviously. The generation of power is replaced by the absorption of power. The phenomenon may be interpreted as the compensatory mechanism to increase gait stability. The hip absorbs power during the mid stance in both group I and group II, but, there is significant difference in amplitude among them. The absorption of power in group I is higher than group II. The hip generates power during the terminal stance in both group I and group II, but, there is also significant difference in amplitude among them. The absorption of power in group II is higher than group I.

## Conclusion

Physical characteristics and different dynamic patterns are both important factors to affect the results of gait analysis. Children with different FPA may use different compensatory mechanism to increase gait stability.

## Reference

1. Gage J. R. Gait Analysis in Cerebral Palsy 1991.
2. Perry J. Gait Analysis : Normal and Pathological Function. 1992.
3. Salenius P. et al. J Bone Joint Surg; 57:259-61. 1975.

# RELATIONSHIP BETWEEN ANTHROPOMETRIC DATA AND GAIT PATTERN IN CHILDREN

Kazuhiko Tsunoda 1), Seiji Noriyasu 2), Hidekatu Takeda 2), Tsutomu Sasaki 3) and Tsutomu Suda 4)

1) Master course of Department of Education, Hokkaido University, kita-ku, Sapporo, JAPAN
2) School of Health Science Sapporo Medical College, chou-ku, Sapporo, JAPAN
3) Hokusei Junior College, chou-ku, Sapporo, JAPAN
4) Department of Education, Hokkaido University, kita-ku, Sapporo, JAPAN

## INTRODUCTION

Childhood is a period in which both body and performance develop rapidly. Anthropometry is important to know physical growth. Walking is a basic movement in human locomotion. Kinematically, gait is regarded as a repetition of falling-down, because it is composed of periodical lower limb swinging. Therefore, it was considered that growth of the lower limbs is related to gait pattern. Clarification of the relationship between anthropometric changes and gait is the base of gait analysis.

The purpose of this study is to clarify the relationship between anthropometric data and gait pattern in children.

## METHODS

The subjects were Japanese 73 school children (32 boys and 41 girls), ranging in age from 6 to 12 years. Anthropometric data for height, weight, lower leg length, tibia height and skinfolds were obtained. Subjects were instructed to walk at ordinary speed along a flat 5 m walkway. A video camera was positioned at a right angle to the walkway. Frame speed was 60 fps. Gait data was calculated from video. Gait data were stride length, stride frequency, velocity and double support (the period when both feet are in contact with ground). Terms defined by Zatsiorky were used. The correlation coefficients for data were calculated.

## RESULTS

Anthropometric data and gait data are shown in Table 1. There was no significant difference in the data between boys and girls. Body dimensions increased with growth. Stride length correlated with body dimensions (i.e. height (r=0.663), leg length etc.). Stride frequency decreased with growth. Velocity is expressed by the product of stride length and stride frequency. Velocity correlated greater with stride length (r=0.844) than stride frequency (r=0.341). Correlation between lower leg length and stride length was positive (r=0.661). Correlation between lower leg length and stride frequency was negative (r=-0.240).

Table 1. Anthropometric and gait data

| | | | Boys (N=32) | Girls (N=41) |
|---|---|---|---|---|
| Age | (yrs) | mean | 10.3 | 10.1 |
| | | s.d. | 1.7 | 1.9 |
| Weight | (kg) | mean | 33.9 | 33.5 |
| | | s.d. | 8.8 | 9.1 |
| Height | (cm) | mean | 138.7 | 137.5 |
| | | s.d. | 11.6 | 12.8 |
| Lower limb length | (cm) | mean | 72.4 | 72.5 |
| | | s.d. | 7.6 | 8.6 |
| Stride | (cm) | mean | 126.6 | 123.3 |
| | | s.d. | 16.5 | 15.4 |
| Stride frequency | (strides/min) | mean | 64.1 | 64.4 |
| | | s.d. | 5.1 | 4.6 |
| Velocity | (m/s) | mean | 1.35 | 1.32 |
| | | s.d. | 0.16 | 0.19 |

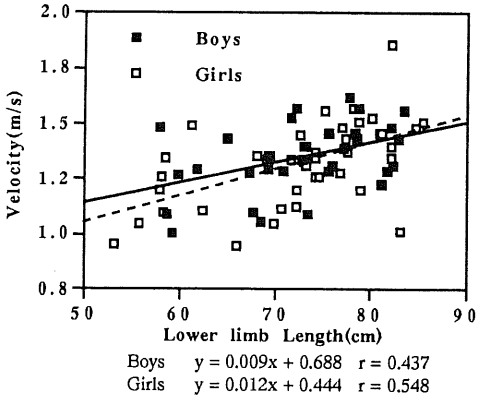



Figure 1. The relationship between lower limb length and velocity.

# GAIT ANALYSIS OF 60~80 YEARS OLD PEOPLE DURING LEVEL WALKING

**A.Y. Lu and X. Wu**
**Shanghai institute of physical education, Shanghai, China**

## INTRODUCTION

Walking is one of the most universal of all human activities, and so the study on gait plays a important role in human movement. The most famous works were done by Elerhart and his assistants from University of California(1947),they got the movement scope of lower limbs and pelvis in different condition. In the 1960's, M. pot Murray et al. (1964) studied the walking pattern of 5 groups of normal man by stroboscopic cinematograph, and got a series of standard parameter of normal gait. Daikerong(1982), from Shanghai ninth hospital, studied the gait of normal human in the beginning of 1980's. In recent years, children, old and patients who have obstruction in walking become important subject in the studying of gait. With the rising of human's average age in city, many countries have paid close attention to the question that study the age's healthy level and body's function to reduce the ratio of sick ages.

## METHODS

In this research, 60-80 years old normal people in Shanghai were filmed by a locam high-speed camera, and the kistler force plate measured the reaction on the ground synchronously while they were walking. By the analysis of data ( two groups, 60-70 and 70-80 years old ), walking speed, step length, gait phases, range of motion of the hip, knee and ankle in sagittal plane and the ground reaction forces were measured or calculated, the graph of change of angle-time, angle-angle, reacting force-time in a walking period were drawn.

## RESULTS

The result indicated that, (1) the subjects' average walking speed, step length and step frequency all descended. Parameters except step frequency showed systematic differences between the groups and different sex. Compared with the 70-80 group and 60-70 group, to female, walking speed descended 19%, step length 11%, step frequency 2.3%, to female, walking speed descended 14%, step length 9% and step frequency 5.5%. So the major reason that cause the walking speed decreased is the step length shortened. (2) Because the individual difference of olds gait characteristic is great, the dispersion of joint angle in different gait phase and period showed great. But the occupying time of gait phase and period, the motion tendency of joints, peak value and the range of over extension of hip joint reduced the flexion of knee joint and extension of ankle increase with their age growing. (3) The result of three-dimensional ground reacting forces showed that the vertical peak value reduced and the others increased, and the force in other direction showed nonsystematic difference.

## REFERENCES

Elerhart, H. D. (1947) Fundamental studies of Human Locomotion and Other Information Relating to the National Research Council, Committee on artificial Limbs. Berkeley, University of california.

Murray MP, et al. (1964) Walking pattern of normal man, J. Bone Joint Surg,46-A:335

戴克戎, 等, (1982) 平地常速行走时的步态观察. 中国生物医学工程学报 1: 15.

# LOWER LIMB JOINT MOMENTS DURING NORMAL AND FAST WALKING OF ELDERLY WOMEN

M. Uematsu * and M. Kaneko **
* Akitsu Kohnoike Hospital, Osaka, Japan.
** Osaka University of Health and Sport Sciences, Osaka, Japan

## INTRODUCTION

The reduced capacity of the elderly influences the speed of locomotion and may cause inactivity in daily life. In fact, walking speed decreases with age in association with reduced stride length and cadence. Although joint movement during gait has often been studied by kinematic procedures, there are few reports on such kinetic factors as the joint moment of force. This study evaluated the gait characteristics of the elderly in comparison with that of a younger population by determining the lower limb joint moments generated during normal and fast walking.

## METHOD

Subjects were comrised of thirteen healthy elderly women aged in 80.7±4.9 years and six younger women aged in 35.8±9.0 years. Each subject walked on a 6m-long walkway at a self-selected speed (normal gait) and at the fastest possible speed (fast gait). Trials were repeated 3-6 times each for both normal and fast gaits. Lower limb joint moments were determined using a 3D Movement Analyzing System (Anima- LOCUS -IIID). Two forceplates (1.8m-long; MG1180) were implanted midway and 4 video cameras were placed along the walkway (MA-2250D). Six LED markers were attached to the left lower limb; midpoint of the iliac crest, two points along the greater trochanter, lateral epicondyle of the femur, lateral malleolus, and head of the fifth metatarsal. The forward and vertical ground reaction forces were recorded. Using the above System with PC data processing, the joint moments at hip, knee, and ankle on the saggital plane were calculated automatically.

## RESULT

In the normal gait, the speed and stride length in the elderly group were significantly lower than those in the younger group. The gait speed and step length were significantly correlated with each other only in the elderly group. The peak moments generated at the lower limb joints during the gait cycle were averaged and values are shown in Table 1 (note that two peaks appeared in the knee joint action). Compared with the younger group, the elderly group showed significantly lower peak moments of hip flexion and knee extension during the acceleration period, and also showed lower moments in knee extension, knee flexion, and ankle dorsiflexion during the deceleration period of the gait cycle. Peak moments in knee extension and ankle plantar flexion during acceleration period were significantly correlated with gait speed in the elderly. On the contrary, in the younger group, the peak moments which showed significant correlations with gait speed were all those generated during the deceleration period; the moments in knee flexion immediately following heel contact, knee extension at the onset of the single stance phase, and ankle dorsiflexion near midpoint of the double stance phase. In the elderly, only the peak moment in ankle plantar flexion showed a significant correlation with stride length.

When gait speed was increased by 'fast walking', both stride length and cadence were increased in the younger group, while only cadence increased significantly in the elderly group. In the younger group, peak moments increased with fast walk in hip flexion during the acceleration period, and in hip extension, knee flexion, knee extension, and ankle dorsiflexion during the deceleration period, indicating that most of the peak moments increased significantly during deceleration period. In the elderly group, the peak moments increased in knee extention during the acceleration period, and hip extension and ankle dorsiflexion during the deceleration period. These results suggest that lower limb joint moments in the younger group play more important role in braking than in accelerating the body, whereas in the elderly, the moments play a larger role in accelerating the body than in braking action for both normal and fast gaits.

Table1. Peak joint moments of lower limb during normal and fast gaits (mean±SD)

| | | | Normal gait | | Fast gait | |
|---|---|---|---|---|---|---|
| | | | Elderly(n=13) | Younger(n=6) | Elderly | Younger |
| Hip | | ext. | 0.48±0.10 | 0.49±0.12 | 0.56±0.11 | 0.70±0.22 |
| | | flex. | 0.25±0.10 | 0.56±0.20*** | 0.33±0.17 | 0.74±0.22*** |
| Knee | Deceleration | ext. | 0.29±0.19 | 0.84±0.32*** | 0.34±0.25 | 1.18±0.38*** |
| | period | flex. | 0.11±0.04 | 0.18±0.05** | 0.17±0.18 | 0.26±0.08 |
| | Acceleration | ext. | 0.23±0.11 | 0.49±0.07*** | 0.25±0.12 | 0.55±0.07*** |
| | period | flex. | 0.14±0.30 | 0.01±0.11 | 0.16±0.31 | 0.03±0.08 |
| Ankle | | plant.flex. | 1.07±0.17 | 1.31±0.11** | 1.07±0.12 | 1.22±0.15* |
| | | dorsiflex. | 0.03±0.02 | 0.14±0.08*** | 0.05±0.03 | 0.23±0.10*** |

*, **, *** indicate significant differences between two age groups at 5%, 1%, and 0.1%, respectively.

# BIOMECHANICAL EVALUATION OF DYNAMIC AND ELECTROPHYSIOLOGIC RESPONSES DURING GAIT IN DIABETIC PATIENTS

I.C.N. SACCO, A.C. AMADIO
Laboratory of Biomechanics of the School of Physical Education and Sport - University of São Paulo, São Paulo, Brazil. Supported by FAPESP 94/2896-9.

## INTRODUCTION

The gait as the fundamental skill for performing independent quotidian duties, is the focus of inumerous studies in neuropathic diabetic patients, that is a high incidence clinical condition in non-insulin dependent diabetics and it usually appears after approximately 15 to 20 years of diagnosis. Its ethiology has not been known totally yet, but neurophysiological and mechanical factors seem to be the agents for the diabetic neuropathy. Biomechanical evaluation of ground reaction forces (GRF) and plantar pressure distribution (PPD), besides neurologic and vascular tests, seem to be the most appropriate to identify patients with evolutive risk to develop diabetic neuropathy, and avoid its further chronic complications. Hence, this study systematizes a methodology to support the interpretation of biomechanical parameters -GRF and PPD- as possible interfering factors on the gait evaluation of diabetic patients, as well as their pain and somatic-sensorial sensibility threshold; thus, we could intervene technical and orthopedically for further functional adaptations in their shoes.

## METHODS

By the analysis of PPD, simultaneous determination of GRF and sensibility tests, like cronaxie (SC) and pain tolerance (PT), we analysed relationships and dependencies in the gait behavior. We tested this proposed methodology in 3 groups approved by the Ethical and Research Committee of the University Hospital: CG- control group (20), DG- diabetic group without any chronic diabetic complications (20), and NDG- neuropathic diabetic group (10). We used a dedicated system for the dynamic analysis (*F-Scan* system) and we acquire the data in 25 Hz for 4 seconds. Four areas were selected to analyse the dynamic data: I- halux; II- medial forefoot; III- lateral forefoot; IV- rearfoot. And for the sensibility tests, the areas were a bite different: I- medial forefoot; II- lateral forefoot; III- midfoot; IV- rearfoot.

## RESULTS

Ground Reaction Forces (GRF) means (Body weight)

| | Fymax1 | | Fymax2 | | Fymin | | Deflec. | | Grow. (%) | | Decay (%) | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | R | L | R | L | R | L | R | L | R | L | R | L |
| CG | 1,16 | 1,12 | 1,06 | 1,08 | 0,73 | 0,65 | 2,73 | 3,12 | 29,18 | 39,14 | 35,34 | 41,41 |
| n=19 | ±0,03 | ±0,03 | ±0,03 | ±0,02 | ±0,03 | ±0,02 | ±0,3 | ±0,27 | ±3,26 | ±2,79 | ±3,21 | ±2,48 |
| DG | 1,08 | 1,1 | 0,98 | 1,02 | 0,73 | 0,75 | 2,13 | 2,31 | 24,12 | 25,65 | 31,15 | 31,44 |
| n=19 | ±0,03 | ±0,3 | ±0,03 | ±0,03 | ±0,02 | ±0,02 | ±0,2 | ±0,19 | ±2,75 | ±2,65 | ±1,98 | ±1,86 |
| NDG | 1,07 | 1,12 | 0,94 | 1,01 | 0,75 | 0,68 | 2,11 | 2,72 | 20,56 | 31,37 | 28,82 | 38,29 |
| n=8 | ±0,06 | ±0,05 | ±0,04 | ±0,07 | ±0,05 | ±0,05 | ±0,33 | ±0,37 | ±3,44 | ±4,68 | ±3,67 | ±4,76 |

Plantar Pressure Distribution (PPD) means (kPa)

| Areas | A I | | A II | | A III | | A IV | |
|---|---|---|---|---|---|---|---|---|
| | R | L | R | L | R | L | R | L |
| CG n=19 | 80,3±5,3 | 83,8±6,4 | 147,8±8,7 | 164±9,6 | 142±8,5 | 128,1±5,9 | 233,3±7,8 | 248,2±12,9 |
| DG n=19 | 74,2±8 | 73,2±7,6 | 143,3±7,8 | 169,4±10,1 | 151±9,3 | 141,7±8,1 | 222,4±12,2 | 235,3±9,9 |
| NDG n=08 | 66,6±9.6 | 85±12,3 | 127,3±9,5 | 165,8±19,1 | 144,9±14,7 | 127,3±10,2 | 218,1±12,4 | 267,7±21,8 |
| Sensitive Cronaxie (SC) medians (ms) | | | | | | | | |
| CG n=20 | 0,33 | 0,38 | 0,28 | 0,35 | 0,33 | 0,25 | 0,30 | 0,33 |
| DG n=20 | 0,43 | 0,38 | 0,30 | 0,33 | 0,28 | 0,33 | 0,43 | 0,53 |
| NDG n=10 | 0,57 | 0,48 | 0,62 | 0,52 | 0,50 | 0,60 | 0,50 | 5,95 |
| Pain Tolerance (PT) medians (mA) | | | | | | | | |
| CG n=20 | 10,75 | 11 | 12 | 13 | 12,50 | 11,75 | 14,25 | 14,25 |
| DG n=20 | 9,75 | 10,75 | 10,50 | 11,75 | 10,50 | 11,25 | 11,25 | 11,75 |
| NDG n=10 | 13,25 | 13,50 | 10,75 | 12,25 | 12,25 | 12,75 | 28,75 | 19 |

## DISCUSSION

Considering the tests to evaluate sensibility, there are remarkable differences among those groups. For example, the data for SC and PT in the left rearfoot for the 3 groups was, respectively, C- 0.33 ms/ 14.25 mA; D- 0.53 ms/ 11.75 mA; and ND- 5.95 ms/ 19 mA. However, PPD haven't shown any statistical difference among those groups; for example, analysing the same area chosen above, C- 248.2±12.9 kPa, D- 235.3±9.9 kPa, ND- 267.7±21.8 kPa. Beside any statistical differences, there is a tendency for NDG presents pressure peaks in left IV area higher than right foot and than other groups. This absence of any statistical differences among the studied groups represents that there is a inter-groups uniformity, because of the heterogeneity of the groups related to the presence or not of diabetic neuropathy or any orthopedic and vascular complications due to diabetes.

Since the data for GRF, we could realize a tendency for these studied diabetic groups not showing consistent differences between them comparing to CG, as our initial thoughts. This could be explained by an adaptation of the locomotor apparatus and their coordenative system in both pathological conditions - DG and NDG.

# GAIT OF THE ELDERLY WOMEN WITH LOAD CARRIAGE

A.Iiboshi*, M.Suenaga*, and T.Matsushita**
*Faculty of Education, Kagoshima University, Kagoshima, Japan
**Faculty of Medicine, Kagoshima University, Kagoshima, Japan

## INTRODUCTION

Falls are more frequently observed among the elderly, and a common cause of morbidity and mortality for the elderly. Suzuki et al (1991) investigated factors related to falls of the elderly, and reported 20.2% of the fallers was carrying a load when they fell. However, little attention has been given to load carriage of the elderly from the viewpoint of safety. The purpose of the present study is to investigate the effects of different type of load carriages on gait parameters of the elderly women.

## METHODS

Six healthy elderly women (age 71.7years, height 1.489m, weight 51.8kg, Body Mass Index 23.5) performed four tasks comprising one normal walking without a load (NL) and three load carriage conditions over a 15m walking distance. The load was 4.25kg, equivalent to 8.3%(6.7%～9.2%) body weight of the subjects, and three load carriage conditions were ; (a)load at the side of the body in the dominant hand(SL), (b)load at the front of the body with two hands and elbows bent at approximately 90° (FL), and (c)load on their back(BL). Gaits were filmed by two video cameras operating at 60fields/sec from frontal and saggital plane views of the subjects. Gait parameters such as step length, step frequency, velocity, the maximum toe height during the swing phase, an inclination of the trunk (trunk angle) at the heel contact were analyzed using video analysis system FRAME DIAS(Denki Keisoku Co.).

## RESULTS & DISCUSSION

The trunk angle at the heel contact in the FL was negative, which means backward inclination, while those of the NL, SL and BL were positive. And also, significant differences were observed in the trunk angle between the FL and every other tasks($p<0.05$). Such backward inclination of the trunk in the FL may effect the step length, resulting that the shortest step length among four tasks. However, the maximum toe height during the swing phase tended to increase in the FL compared with that of the NL. Since the short step length generally involves the low toe lift during the swing phase, a pattern of the step length and the toe height during the swing phase observed in the FL is an interesting pattern from the viewpoint of safety. In contrast, subject S, who was slender with Body Mass Index of 20.2, tended to walk with the low toe lift during the swing phase and the short step length in the FL compared with those of the NL. Such a pattern may be due to the largest backward trunk inclination at the heel contact and the highest step frequency in the FL among four tasks. Kaneko et al(1991) suggested that a lower foot of the elderly during the swing phase make it more difficult to step over an obstacle. Carrying a load at the front of the body with two hands may obstruct the view, then the gait pattern shown by subject S may be one of the causes of falls during a load carriage.

## REFERENCES

1,M.Suzuki, M.Endo, K.Eguchi, K.Kano, S.Tsuchiya, C.T.Okamura, and Y.Shimazu,(1991). A study of falls experienced by the aged among users of preventive health examination services. Jpn. J. Public Health,**38**:743-750.
2,M.Kaneko, Y.Morimoto, M.Kimura, K.Fuchimoto, and T.Fuchimoto(1991). A kinematic analysis of walking and physical fitness testing in elderly women. Canad. J. Sports Sci., **16**:223-228.

# VENTILATORY THRESHOLD AND MECHANICAL EFFICIENCY DURING WHEELCHAIR PROPELLING ON A TREADMILL

A. Azuma
Fukui Health Administration Association, Fukui, Japan

## INTRODUCTION

The ventilatory threshold has suggested as an alternative parameter for assessing submaximal endurance and for setting the intensity threshold of training. However, there is lack of information about ventilatory threshold during actual wheelchair propelling on a treadmill. And also, previous investigators have not attempted to estimate the mechanical work required in wheelchair propulsion to simply move the upper body including limbs. Therefore, the purpose of this study was twofold: firstly to measure ventilatory threshold during wheelchair propelling on a treadmill, and secondly to determine the mechanical efficiency of treadmill wheelchair propulsion at different velocities based on external and internal mechanical work.

## METHODS

Five male non-wheelchair users (age 28.8±5.0 years) and one paraplegic (age 19 years) participated in this study. All of them completed a continuously progressive exercise test of wheelchair propelling on a motor-driven treadmill, and peak oxygen uptake (peak $\dot{V}O_2$), ventilatory threshold (VT), and VT expressed as a percentage of the peak $\dot{V}O_2$ (VT%) were measured. And also, three tests of treadmill wheelchair propulsion at different velocities (0.56, 1.11, and 1.67 m/s), each of 5 min duration, were performed by non-wheelchair users, and metabolic and kinematic data were collected. The internal work rate ($\dot{W}_{int}$) was determined from digitized kinematic data by modeling head-neck-trunk, upper arm, forearm, and hand as a four-segment linked system and calculating their changes in potential, translational kinetic, and rotational kinetic energy. The external work rate ($\dot{W}_{ext}$) was derived from the measured drag force by a simple drag test on the treadmill and belt velocity. The total mechanical work rate ($\dot{W}_{tot}$) was defined as the sum of the $\dot{W}_{ext}$ and $\dot{W}_{int}$. Gross efficiency (GE) and mechanical efficiency (ME) were calculated as the ratio of $\dot{W}_{ext}$ to energy expenditure ($\dot{E}$), and of $\dot{W}_{tot}$ to $\dot{E}$, respectively. In addition, different cycle frequencies were imposed at a constant velocity (1.11m/s) for one of the non-wheelchair users, and $\dot{W}_{tot}$ and ME were calculated.

## RESULTS AND DISCUSSION

Average peak $\dot{V}O_2$, VT, and VT% were 1.09±0.17 l/min, 0.65±0.22 l/min, and 60.7±4.6%, respectively in non-wheelchair users, and lower VT% compared with that of the non-wheelchair users was showed in a paraplegic (VT%: 48.9%). It is possible that these physiological responses taking place during wheelchair propelling on a treadmill are measurable, and that further studies are necessary to identify critical differences in these variables of experienced and inexperienced wheelchair users. The efficiency increased rapidly when the velocity was increased from 0.56 to 1.11 m/s (GE: 5.0±0.5 to 7.2±0.8%, ME: 13.0±1.7 to 19.0±2.4%). Thereafter, the efficiency slightly increased with the velocity to 7.5± 0.9% in GE and to 21.0±2.3% in ME at 1.67 m/s (Fig. 1). Furthermore, the average $\dot{W}_{tot}$ remained constant when different cycle frequencies were imposed. This work rate level was identical to that required for treadmill wheelchair propulsion at the same velocity with self-selected (free) cycle frequency. Thus, minimum $\dot{E}$ and maximum ME resulted at the free cycle frequency. It is suggested that the evaluation of VT and ME during wheelchair propelling on the treadmill as supplementary parameters will enable researchers to ascertain the specifics as a loading apparatus.

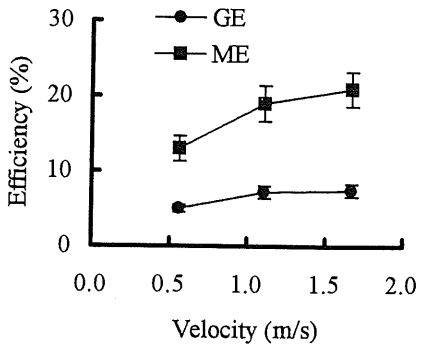



Fig. 1. GE and ME of treadmill wheelchair propulsion at different velocities

# SHOE-SURFACE INFLUENCE ON IMPACT SHOCK TRANSMISSION AND ATTENUATION

E. C. Hardin and J. Hamill

Dept. of Exercise Science, University of Massachusetts, Amherst, MA, USA

## INTRODUCTION

Independently, shoe and surface properties contribute to locomotion impact shock moderation in magnitude and frequency (Light *et al.*, 1980; Nigg *et al.*, 1983; Shorten *et al.*, 1986). These two properties likely interact to influence impact shock. Therefore, the purpose of this study was to investigate the influence of midsole hardness and surface stiffness on impact shock transmission and attenuation during running.

## PROCEDURES

Six male fitness runners, free from lower extremity injury, volunteered to participate in this study. The subjects ran at 3.4 m $\bullet s^{-1}$ for six minutes in each of six conditions (C1-C6).

| | C1 | C2 | C3 | C4 | C5 | C6 |
|---|---|---|---|---|---|---|
| Shoe[1] | S | S | S | H | H | H |
| Surface[2] | S | M | H | S | M | H |

[1] Midsole hardness: S, soft (40 Shore A); H, hard (70 Shore A).
[2] Stiffness: S=100 kN/m; M =200 kN/m; H=350kN/m.

Condition order was balanced across subjects. Leg and head shock were measured with two 1.7 gram PCB accelerometers mounted on the anteromedial distal tibia and frontal bone (Hamill *et al.*, 1995). The accelerometers were interfaced to a microcomputer via an A/D converter and sampled at 1 kHz. A 10 trial mean value was calculated for peak leg (PL) and head (PH) acceleration and transmission (PH/PL). The frequency domain signals were obtained through an FFT on 10 trials per condition. A transfer function (TF) was calculated for each 1.0 Hz frequency bin using the formula:

$$TF = 10 \log_{10}(PSD_{head}/PSD_{leg}).$$

A two factor repeated measures ANOVA was conducted on time and frequency domain parameters.

## RESULTS

PL and PH were greatest on the hard surface while shock transmission (PH/PL) was greater in the soft shoes (Figs. 1 and 2). Leg power spectra from 5-18 Hz were greatest in the hard surface regardless of shoes (Fig. 3) as also were the head spectra from 8-16 Hz. There was a significant shoe by surface interaction from 27-40 Hz in the leg spectra. For both the leg and head, the power at 4 Hz was influenced by the shoe, with the soft shoe producing greater power. The TF displayed a gain in the 0-5 Hz range and an increasing but disparate attenuation with increasing surface hardness in the frequencies from 10-20 Hz and 30-40 Hz (Fig. 4).

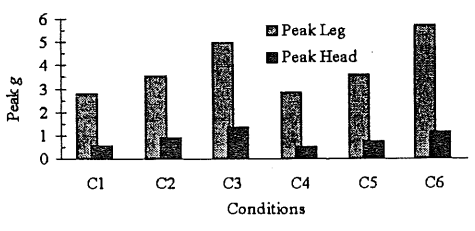



*Fig. 1. Peak leg and head acceleration in six shoe and surface combinations.*

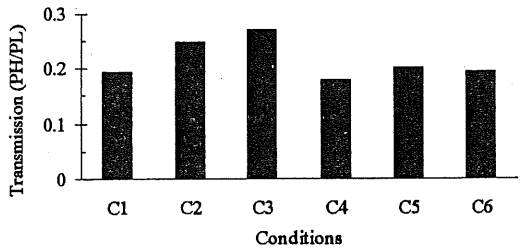



*Fig. 2. Shock transmission in six shoe and surface combinations.*

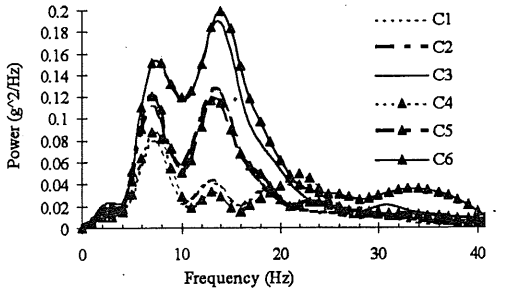



*Fig. 3. Leg spectra in six shoe and surface combinations.*

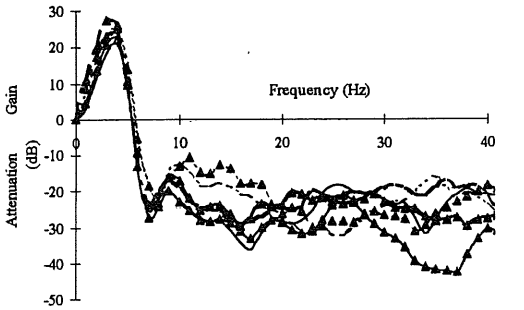



*Fig. 4. Transfer function in six shoe and surface combinations.*

## DISCUSSION

- Surface appears to dominate impact shock attenuation.
- Peak leg and head shock increased with surface stiffness, but shock transmission was greater in the soft shoe.
- The surface provided greater influence on the leg and head spectra from 10-20 Hz rather than the shoe.
- The hard shoe and hard surface resulted in greater power from 30-40 Hz in the leg spectra.
- Attenuation of frequencies from 10-20 Hz and 30-40 Hz was disparate between surfaces.

## REFERENCES

Hamill, J *et al.* (1995) *Human Mvmt Sci.* **14**, 45-60..
Light *et al.* (1980) *J. Biomechanics* **13**, 477-480.
Nigg, B. M. *et al.* (1983) In *Biomechanics VIII*-B, pp. 1041-1052.
Shorten *et al.* (1986) *Med. Sci. Sports*, **18**, S80-S81.

# EVALUATION OF REARFOOT KINEMATICS AS A FUNCTION OF SHOE MILEAGE

B. Van Gheluwe, W. Lofgren and K. Rutgeers
Vrije Universiteit Brussel, Brussels, Belgium

## INTRODUCTION

Shoe characteristics, especially shock absorption capacity, change during the lifetime of a shoe (Baumann, Krabbe & Galbierz, 1995; Cook, Marcus & Brunett, 1985; Jones & Williams, 1989; Hamill & Bates, 1988). However, the influence of mileage induced changes on the running pattern is not well demonstrated. Especially rearfoot motion is of concern because of the well established relation between subtalar pronation and overload injuries.
Therefore, it is the purpose of this study to examine rearfoot kinematics due to mileage induced deterioration of the medio-lateral stability of running shoes.

## METHODS

Twelve joggers, running with an Adj las Torsion Integral S anti-pronation shoe were tested every 200 km till they completed the 1400 km mark. During the test on a treadmill running at 3.8m/s, frontal rearfoot motion , as represented by subtalar pronation and calcaneal eversion was recorded three-dimensionally using a VICON motion analysis system. Additionally, the contact time (as a percent of stance duration) of the heel, the metatarsal heads and the hallux of the foot were recorded with an EDG system. Also the hardness (in Shore A) of the lateral and medial midsole of the heel was measured.
ANOVA statistics, in combination with a Fisher Least Significant Difference a posteriori test, were used to detect significant differences (at the .05 level) between the values recorded at different mileage's.

## RESULTS AND DISCUSSION

Figure 1 shows maximal pronation and eversion to increase significantly with mileage. However, after about 800km a decrease can be observed, especially for calcaneal eversion. This decline in eversion is also noticeable for pronation and eversion at the moment of heel-off. Though this decline in frontal rearfoot stability is not statistically significant, observation of the graph depicting a sudden rise in hardness of the medial midsole at 800km (fig. 2), offers a plausible explanation. It is well-known that increasing hardness at the medial posterior side of the shoe, improves frontal shoe stability during the pronation period of the stance phase. Therefore, hardening of the midsole with mileage seems to partially compensate for the loss in medio-lateral stability of the upper shoe construction. However, as hardness increases laterally too, shock absorption at heel impact should diminish, which is in agreement with literature findings.
The results also show a steady and significant increase of the contact time of the medial metatarsal heads and the hallux with mileage, implying a longer loading time of the medial forefoot. This finding, together with the increase of subtalar pronation mentioned above, suggests a growing presence of midtarsal pronation. Therefore, forefoot stability should attract more attention from biomechanical researchers and manufacturers.

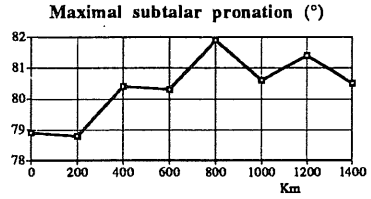


Fig.1a

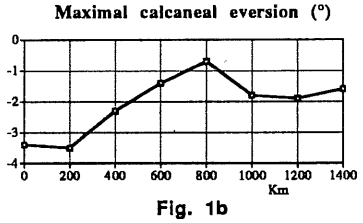


Fig. 1b

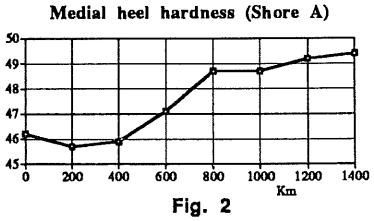


Fig. 2

## CONCLUSION

From all the results above, one may conclude that for the type of shoe mentioned above, medio-lateral stability of the rearfoot significantly deteriorates with mileage. However, after 800km, a partial compensation can be observed due to hardening of the heel midsole.

## REFERENCES

Bauman, W., Krabbe, B. and Galbierz, P. (1995) Life Characteristics of running shoes and shoe material. *Proceedings of the Second Symposium on Footwear Biomechanics*, Deutsche Sporthochschule, Cologne, pp.38-39.
Cook, S.D., Marcus, A.K. and Brunett, M.E. (1985) Shock absorption characteristics of running shoes. *Am. J. Sports Medicine* 9, 248-253.
Hamill, J. and Bates, B. (1988)A kinematic evaluation of the effects of in vivo loading on running shoes. *J. Orthop. and Sports Phys. Therapy* 10, 47-53.
Jones, J.E. and Williams, K.R. (1989) Changes in the material properties of running shoes due to shoe wear. *Proceedings of the 12th Int. Congress of Biomechanics*, University of California, Los Angeles, abstr. 140.

# BIOMECHANICAL ANALYSIS OF DROP LANDING FROM DIFFERENT HEIGHTS IN TAIWAN COLLEGE STUDENTS

Chenfu Huang, Ing-Jyh Chang, and Rey-Hung Tu
National Taiwan Normal University

## INTRODUCTION

The Physical Education (PE) major students in Taiwan college require to take eight to twelve hours of activity classes each week compare to the two hours class by the non-PE major students. The activity classes including lots of running and jumping which involve landing impact from different heights. The purpose of this study was to analyze the kinematics and kinetic differences of drop landing from different heights (0.59, 0.72, 1.0m) by the Taiwan college male students.

## METHODS

Twenty-nine PE major (height : 172±5.6cm ; weight : 67.7±4.8kg ; age : 21.5±1.7yrs) and twenty-one non-PE major (height : 172±5cm ; weight : 65.6±15.1kg ; age : 20.7±2.7yrs) college male students served as the subjects for this study. Kinematic and kinetic data were acquired simultaneously by using a Peak Performance high-speed camera (120Hz) and a Kistler force platform (9287, 1200Hz). Test procedures and variables calculation were according to McNitt-Gray's study (McNitt-Gay,1991). The selected variables were tested by a mixed design ANOVA at $\alpha$=.05 significant level.

## RESULTS AND DISSCUTION

The results(Table1.) indicated that there were significant increases in lower extremity joints flexion (with the exception of ankle joint flexion ), peak angular velocity, first (PK1) and second (PK2) vertical reaction force, fifty microsecond passive impulse(Nigg,1985) and the total impulse as landing height increased. In addition, time to peak angular velocity and time to PK1, PK2 were significant decreases as landing height increased. The PE major had a greater fifty microsecond impulse and a shorter time to PK1 and PK2 than the non-PE students major which indicated that PE major students may used different landing strategy than the non-PE major students. It is suggested that when landing from a greater height, the subjects should increase their hip and knee joint flexion in order to reduce the peak vertical reaction force and 50 ms passive impulse.

| Group | Height (m) | Hip (rad/s)* | Knee (rad/s)* | Ankle (rad/s)* | Pk1 (BW)* | Pk2 (BW)* | 50 ms impulse (N*sec) *+ |
|---|---|---|---|---|---|---|---|
| PE major (n=29) | 0.59 | -12.1 | -15.5 | -21.3 | 2.16 | 5.86 | 9.14 |
| | 0.72 | -13.3 | -16.0 | -23.4 | 2.76 | 6.79 | 9.61 |
| | 1.0 | -14.5 | -17.9 | -27.3 | 4.16 | 8.47 | 13.31 |
| non-PE major (n=21) | 0.59 | -12.3 | -16.5 | -22.0 | 2.15 | 4.76 | 6.94 |
| | 0.72 | -12.8 | -16.1 | -24.2 | 2.94 | 5.55 | 8.59 |
| | 1.0 | -14.3 | -17.3 | -26.9 | 4.35 | 8.11 | 12.08 |

*height $p<.05$
+group $p<.05$

Table1. mean values and significant level for the peak joint velocities, Pk1, Pk2 and 50 ms impulse.

## REFERENCES

McNitt-Gray, J.L. (1991). Kinematics and impulse characteristics of drop landings from three height. Int. J. Sports Biomechanics, 7, 201-224.

Nigg, B.M. (1985). Biomechanics, load analysis and sport injuries in the lower extremities. Sports Medicine, *2*, 367-379.

# THE PATTERN DIFFERENCE DURING SUSTAINED LONG STANDING BETWEEN TRAINED AND UNTRAINED YOUNG MALES

Junko KATAOKA*, Machiko SACRAGI**, Hideko KIKUCHI***,Atsushi HAYAMI*** and Kazuo MAIE****
*Japan Women's Junior College of Physical Education, **Bunkyo Women's University, ***Japan Women's College of Physical Education, ****Otsuma Women's University.

## INTRODUCTION

We have already studied the postures or spontaneous movements during sustained long standing on children (1981,1995), young females (1983,1984) and middle-aged males and females. The purpose of present study was to compare the patterns during sustained long standing between well-trained young male athletes and untrained ones.

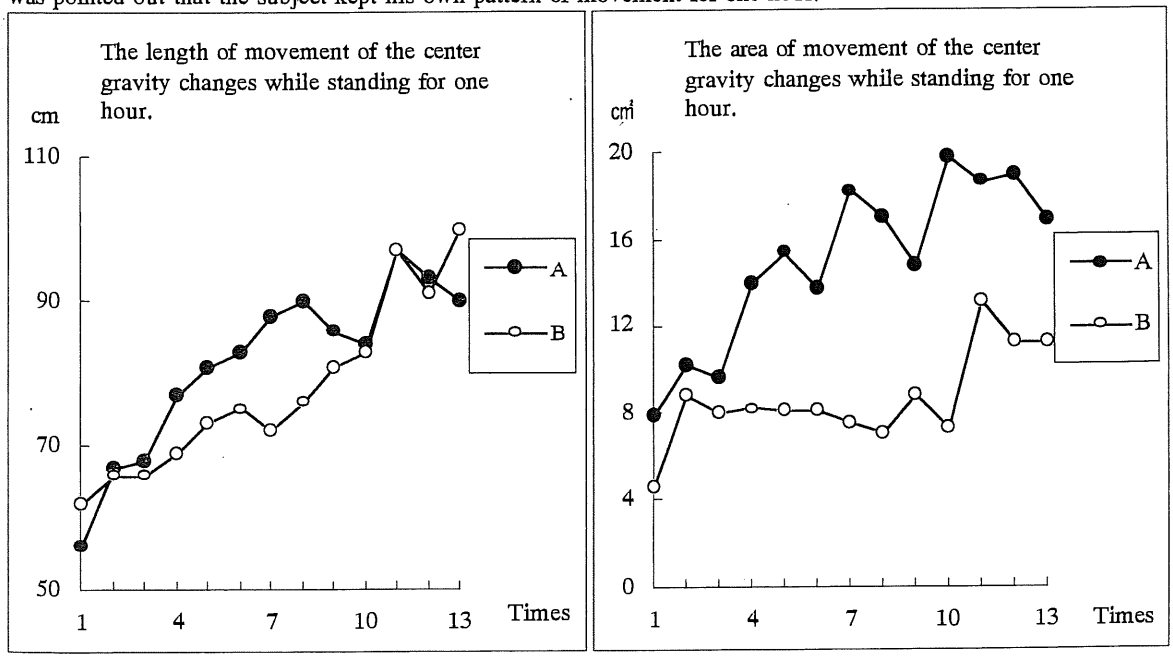
## METHODS

Twenty male subjects (21 to 23 years old) were asked to sustain standing as still as possible for one hour. Ten students of a physical education college were as the well-trained subjects (A-group) and ten students of an engineering course were as the untrained subjects (B-group). When the subject wanted to quit, the experiment was stopped. The instruction to the subjects were to stand still at ease looking at a fixed point of the eye level. The spontaneous movements and physiological responses of heart rate and respiratory rate were recorded.

Subjects: The athletic career and present athletic habits of both groups were counted as the points; 49.5 for A-group and 11.4 for B-group. The Rohler-Indicies were 122.7±75.2 for A-group and 124.0±21.2 for B-group. The points of hand-grip-strength, back-strength, trunk-extension, standing trunk flexion, side-step-test, closed-eyes-foot-balance and vital capacity of A-group were higher than those of B-group, but the tapping point of B-group (110.0± 12.7) was higher than that of A-group (105.7±8.3), the subjects of B-group were familiar with key-boards operation of computers.

## RESULTS

The results were as follows; The mean minute heart rates at rest were 75. 2 ±11.9 (A-group) and 93.0±14.9 (B-group). The means of maximum minute heart rates during standing were 93.6±13.3 (A-group) and 110.2±13.9 (B-group). The mean sway areas and lengths of the center of gravity (measured by Kistler) during an hour were 13.6 ±14.2 cm² (A-group) and 8.7± 6.5 cm² (B-group), and 81.7± 36.1 cm (A-group) and 76.4± 20.4 cm(B-group), respectively. But all these differences were not significant. The close relation between sways and feet, and that between head sways and center of gravity sways were suggested. Standard deviations of respiratory rate during standing of B-group were significantly larger than that of A-group. The sway of the center of gravity became larger according to the time. It was pointed out that the subject kept his own pattern of movement for one hour.




REFERENCE: Junko KATAOKA et al (1984). Evaluation of the Ability to Sustain a Prolonged Standing Posture. Annals Physiol. Anthrop.3 (1):39-45

# A STUDY ON THE BIOMECHANICS OF SLIPPING ACCIDENTS

Kuk-Woong Seo*, Yang-Jin Yoon**, Joong--Sook Lee***, Yong-Jae Kim****,
Seung--Jae Jo****, Hoon-Sig Lee****, Mi-Ra Jung*****, Seung--Bum Park****

* Professor, Pusan National University, Pusan, Korea
** Assitant Professor, Pusan National University, Pusan, Korea
*** Assitant Professor, Pusan Wemens University, Pusan, Korea
**** Graduate School, Pusan National University, Pusan, Korea

## Introduction

The foot, while often viewed as only a semirigid base of support for the upright body, provides an import dynamic complex which is an integral part of the locomotor system. Several authors have recommended the utilization of specialized foot supports as a possible means to normalize irregular ground reaction force patterns(James, Bates, & Osternig, 1978: Schoenhaus, Gold, Hylinski, & Keating, 1979; Scrantom, Pedegana, & Whitesel, 1982; Sgarlato, 1971; Subotnick, 1979). The purpose of this study was to analyze walking steps to prevent unnessary energy consuming and prevention of injury from sliding when someone was walking. The questionaire, high video system, E.M.G. system, and force platform system were used to analyze sliding mechanism in biomechanical perspective.

## Research Methods

The subjects of this study was 4 Korea Standard Physical Characteristic subjects who slected 10 ages, 20 ages, 30 ages, 40 ages in Korea. The Instrument for this study was Force Platform (size:4060, Bertec Co), Electromyograph System (Noraxon EMG), High Speed Video Camera System (Nac. Co. Japan). A Bertec Type 4060A piezoelectric force platform (60cm by 40cm)was used to measure the ground reaction forces. The signal from the force platfom was amplified using a Bertec Type 5053A Charge Amplier. Four of the amplified outputs (two Fx and two Fy) and a 6V synchonization pulse were sent directly to ADAC (U.S.A) Converter interfaced with a Analysis Computer. A NAC projection head projected the film images onto a GTCO digitizer, connected to a computer. The positions of body landmarks in the projected images were digitized for every frames of the cameras.

## Results and Duscussion

Several results were found as following (1) According to questionnaire regarding slindin, 93.3% of participants have sliding experiences. People have sliding experience by greasy floor (48.4%) and abstacle (22.5%). Sliding types were found from different kinds of shoes such as sneakers (48.8%) and boots (34.2%). The results were presented as following 44.6% of participants were fall down front direction and 40.3% of people fell down rear side. (2) According to the high speed video system analyzer, step sliding increased with increasing step speed from distance and acceleration speed experiment with 80 step/min, 100 step/min. Also, the most were occurred with sneakers, boots, and bare feet as order. (3) According to E.M.G test results, the vastus lateralis M., the vastus medialis m., and the tibialis anterior m. of electric action potential presented strongly when participants fell front, rear, and right side. The vastus lateralis m. and the tibialis anterior m. of electric action potential presented as weaker. (4) According to the force test results, the most big change of floor antipower of x-axis presented when fell the left side. The most of big change of floor antipower of y-axis presented when fell down rear side. The most of big change of floor antipower of z-axis presented similarly when someone fell the front, rear, left and right side.

## References

Bates, B.T., L.R.Ostering, B.Mason, and L.S.James (1979), Foot orthotic devices to modify selected aspects of lower extrmity mechanics. Am.J.Sports Med. 7(6): 338-342

Batemam, J.E., and Trott,A (1980), The foot and ankle, thieme strattam.Inc., New York.

Roger A.Mann (1982), Biomechanics of running, Symposium on the Foot and Leg in running sports, The C.V.Mosby Company, P.I.

Winter,D.A. (1979), A New defintion of mechanical work done in human movement. J.Appl. physiol, 46:79-83

# INVESTIGATIONS ON KINEMATICS OF GIANT SLALOM'S TACTICS IN ALPINE SKIING

Wlodzimierz S. Erdmann (1), Vassilios Giovanis (2)
(1) J. Sniadecki University School of Physical Education, Gdansk, Poland
(2) School of Physical Education and Sport Sciences, University of Athens, Athens, Greece

## INTRODUCTION

Description of slalom or down hill races in alpine skiing can be performed according to the technique, tactics, psychological problems, equipment used. Tactics of skiing includes choise of a way within a track, disposition of a velocity between gates and along the whole course. Tactic elements could be described in a kinematic way (displacement, velocity, acceleration/deceleration) according to the detailed geometry of a track including setting of gates.

Up to now investigations on slalom skiing were performed on a small fragment of a track, i.e. 2 - 3 gates (Cotelli and Cotelli 1979, Nachbauer 1987). In the pilot study (Erdmann and Giovanis 1995) velocity and acceleration of skiers based on detailed geometry of 10 gates setting of giant slalom's course were presented. Those investigations were performed during the regional competition in Arachova (Greece) and the next day during the training session of short practice skiers (on the same setting of giant slalom). Detailed time was obtained by video recording.

The aim of the present investigations was obtaining of the kinematic data on upper level of skiers and for the whole length of a course.

## MATERIAL AND METHODS

Investigations of 41 men during Greek Championships in Alpine Skiing in Naoussa were performed. From this number the best three, the medium three, and the worst three skiers in giant slalom were evaluated.

The geometry of the whole slalom, i.e. distances between all gate poles (vertical and surface), as well as all angles between them were obtained by geodetic theodolit. All giant slalom's runs were recorded. Based on displacement and time - mean velocities and accelerations obtained between the gates were calculated. The coefficient of acceleration was calculated by dividing the absolute value of mean acceleration of the course by mean velocity of the course.

## RESULTS

The vertical drop of the whole slalom equaled 288.18 m and surface length (from pole to pole - 36 segments) 936.42 m. The mean gradient of the course's pole to pole profile equaled 17.86 degrees (S.D. = $4.23^{o}$), the mean vertical drop equaled 8.01 m (S.D. = 1.78 m), the mean surface length from pole to pole equaled 25.95 m (S.D. = 4.97 m), and the mean angle of deviation, i.e. the angle between the extension of a line of the preceeded distance between gate's turn poles and the line of actual distance between gate's turn poles, equaled 24.8 degrees (S.D. = $12.3^{o}$).

There was a big diference in mean velocities of the whole course noted between the best skiers (mean = 16.8 m/s) and the worst (12.0 m/s). Medium group reached 14.5 m/s. The mean acceleration equaled for the best group 2.5 $m/s^2$, for medium 1.33 $m/s^2$, and for the worst group 1.1 $m/s^2$.

## DISCUSSION

Comparing the coefficient of acceleration between the representatives of the three groups investigated in Naoussa it was observed, opposite to the data obtained in the pilot study of Arachova for lower level competitors, the better skiers the bigger values of this coefficient: 0.150, 0.098, 0.096 respectively. But comparing the best three skiers from Naoussa the best of them had smaller value of this coefficient (0.119).

Some skiers of the best group (2nd and 3rd in Naoussa) obtained in few examples big value of acceleration (up to 15 $m/s^2$). This allowed to gain a good result in giant slalom competition but on the other hand such a situation is close to the accident. The coaches, competitors, and slalom setters must be aware of above situation.

## REFERENCES

Cotelli, F. and Cotelli, M. (1979) Sci competizione. La tecnica delle tre discipline : slalom, gigante e discesa (Ski competition. The technique of three disciplines: slalom, giant slalom and down hill). Longanesi & C. Milan.

Erdmann, W., Giovanis, V. (1995) Wstepne badania kinematyki narciarza w slalomie gigancie (Preliminary investigations of skier's kinematics in giant slalom). Proc. Polish Conf. Biomechanics, Cracow, 20-22 September 1995 (Edited by Mazurkiewicz, S.) pp. 57-62. B. Czech University School of Physical Education, Cracow.

Nachbauer, W. (1987) Fahrlinie in Torlauf und Riesentorlauf (Course line in slalom and giant slalom). Leistungssport, 6:17-21.

# BIOMECHANICS OF SKI SLALOM
## - COMPARISON OF SLALOMS WITH POLE SETTING AND WITHOUT POLE SETTING-

Hisako IKEGAMI*, Hiroshi SODEYAMA**, Yoshiro ANDO***, Yasuo IKEGAMI****, Shinji SAKURAI**** and Kyounosuke Yabe****

* Nagoya Seirei Junior College, Seto Japan
** Kinjo Gakuin University, Nagoya Japan
*** Chukyo University, Nagoya Japan
**** Nagoya University, Nagoya Japan

## INTRODUCTION

As in slalom of Alpine ski racing distance between consecutive gates is shorter than in giant slalom or in down hill, skier try to take his running course nearest to the inside pole of a gate to minimize the running time. To make this slalom technique possible skier knocks down the pole with the reverse side hand (ex. in right turn left hand is used). This movement of knocking down a pole is one of the most important technique for higher performance. Therefore skier practices running on a slope with pole setting and learns timings of ski control and ski stick planting and knocking down poles. However it is not always possible to set pole in a slope. When pole setting is impossible skier sometimes run on a slope without pole setting but imaging pole setting (imaginal slalom).

The purpose of this study is to investigate the difference between imaginal slalom and actual slalom from the biomechanical view points.

## METHOD

Two skiers who experienced Alpine ski racing performed slalom in the same course settled in the experimental slope with pole setting (actual slalom) and without pole setting (imaginal slalom). In imaginal slalom piste marks were located in the same positions as set poles in actual slalom. Movements of the skier's body as well as skis and ski sticks during running were recorded with two synchronized high speed movie cameras. The obtained images were digitized and 3-dimensional coordinates of reference points of body and ski were reconstructed by DLT method. Kinematic data such as edging angle, angle of attack, and joint angles and the center of the gravity of the skier's body (C.G.) were calculated from three-dimensional coordinates of these reference points.

## RESULTS

Stick was planted at the mid point of the consecutive two gates in actual slalom, while in imaginal slalom it planted just after passing the piste mark. In both slalom, however, the stick was planted in the same timing which was just after the maximum phase of the turn in which edging angle and angle of attack reached their maximum. These results indicated that the timing of the turn was different in these two types of skiing, that is, in imaginal slalom the switchings of the directions of edging and angle of attack was earlier in comparison to those in actual slalom. The lateral movement of C.G. according to turn was also initiated in earlier phase of a turn in imaginal slalom. In the forward and backward movement of C.G. it was observed that in imaginal slalom C.G. was kept forward throughout the turn. Inclination of upper body was larger in imaginal slalom while angular change in inclination of lower leg was larger in actual slalom. Even though changing patterns in flection angle of hip and knee joints were similar in both types of slalom, time delay in changing pattern was observed in actual slalom. The timing of body movements such as displacement of C.G. or flection/extension of the knee and hip joint also became earlier in imaginal slalom.

## CONCLUSION

From the results of this study, it can be summarized that in actual slalom skier had to knock down a pole and this motion caused time delay of body movement necessary to successive turn. The practice in imaginal slalom may have advantage in correcting the body positions as well as in improvement of the body movement for the turn with higher speed.

# KINEMATIC ANALYSIS OF SNOWBOARD TURN
## - COMPARISON WITH SKI TURN-

Yasuo IKEGAMI[1], Shinji SAKURAI[1], Kyonosuke YABE[1],
Hiroyuki NUNOME[1], Hisako IKEGAMI[2], Hiroshi SODEYAMA[3],
Atsushi OKAMOTO[4], Toru TERASHIMA[5] and Yshiro ANDO[5]

1 Nagoya University, Nagoya Japan
2 Nagoya Seirei Junior College, Seto Japan
3 Kinjo Gakuin University, Nagoya Japan
4 Nagoya Women's Junior College of Commerce, Owari Asahi Japan
5 Chukyo University, Nagoya Japan

## INTRODUCTION

In recent years snowboarding has become dramatically popular in the younger generation. The mechanism of snowboarding can be considered fundamentally the same as that of skiing. In skiing to achieve successive turn the following three factors are essential ; 1) edging of ski, 2) angle of attack of ski and 3) displacement of the center of the gravity of the skier's body (C.G.) inside of turn arc. These three factors may also be essential in snowboard turn, while the body movement required to actualize these three factors may differ in skiing and snowboarding.

The purpose of this study was to investigate movements of a snowboard and snowboarder's body during snowboard turn in comparison with ski turn from the biomechanical view points.

## METHOD

One excellent skier who was also highly skilled in snowboarding performed turn with ski and snowboard on the same slope with the same course setting. Trials with ski and snowboard were recorded by a pair of synchronized cine and video camera respectively. Three-dimensional analysis technique was applied to obtain three-dimensional movements of skis or snowboard and body movements. To record movements of skis or a snowboard two small pole on which two reference mark were attached were fixed perpendicularly to sole in front and rear of skis or snowboard. From three dimensional coordinates of reference mark of the body and skis or snowboard edging angle, angle of attack, movement of C.G. and joint angles of body were calculated.

## RESULTS AND DISCUSSION

Top views of trajectories of C.G. in ski turn and snowboard turn were similar. Though movement of skis and snowboard were also similar, traces of snowboard turn was narrow in comparison with those of ski turn. In snowboard turn the angle of attack was smaller than in ski turn. This results showed it might be possible for excellent snowboarder to make a turn with thin track. The lager edging angle was required in snowboard turn than in ski turn with the same curvature and the same running speed, while lateral movement of C.G. necessary to generate the centripetal force for turn did not differ from each other. In ski turn joint angles of hip and knee joint changed with almost the same manner, while in snowboard turn joint angles of hip and knee joints changed with the rivers phase. There observed difference in angular change in flection/extension of right and left knee joints..

## CONCLUSION

From the results of this study, it may be considered that the movements of hip and knee joints specific to snowboard turn make possible to maintain upper body stable during snowboard turn as it is common in excellent skiers turn.

# JOINT TORQUES OF THE SUPPORT LEG IN SPEED SKATING

**Masahiro YUKI[1] and Michiyoshi AE[2]**
1)Doctoral program in Health and Sport Sciences, University of Tsukuba, JAPAN
2)Institute of Health and Sport Sciences, University of Tsukuba, JAPAN

## INTRODUCTION

Although speed skating is different from other types of human locomotion in the sense that the forward velocity is achieved by sideward push-off of the support leg, no investigation has been done three dimensional analysis of the joint torques of the support leg.

The purposes of this study were to estimate three dimensional joint torques of the leg during stroke by using the specially-made sensor skate and videography, and to identify the most important joint motion to accelerate the skater.

## METHODS

Fast skating trials for twelve Japanese speed skaters, including world record holder were videotaped by eight VTR cameras (60 fields/s) at the straight-way of skating oval. Three dimensional coordinates of the body segment endpoints were computed by using DLT technique. Both vertical and horizontal components of blade reaction forces (BRF), point of BRF application and free moment in an ice reference frame were measured with sensor skate at 100Hz. Three dimensional joint torques were calculated from standard Newtonian equations and transformed to seven motion axes fixed to the lower limb joints, i.e. abduction-adduction, extension-flexion, int-out rotation axes at the hip joint, extension-flexion, int-out rotation axes at the knee joint, plantar-dorsi flexion, pronation-supination axes at the ankle joint.

## RESULTS AND DISCUSSION

Figure 1 shows mean the joint torques about the hip abduction and extension, knee extension and plantar flexion axes of the support leg during stroke time. Shaded area indicates the range of torque. Peak of the hip extension torque was almost twice of the knee extension and plantar flexion torques. There was large range in abduction and adduction torque. It seems that the hip abduction torque would relate to skating skills and performances.

Figure 2 demonstrates correlation coefficient between joint torques and horizontal BRF at three instants of the gliding and push-off (from the start of the hip or knee extension to the end of stroke) phases. Hip abduction torques at mid-point of both phases and hip extension torque at the start of push-off showed high positive correlation coefficient with horizontal BRF. These positive relationships indicate that the hip abduction and extension motions by the torques exerted at the joints contribute to the exertion of the horizontal component of BRF which increases skating velocity during stroke.

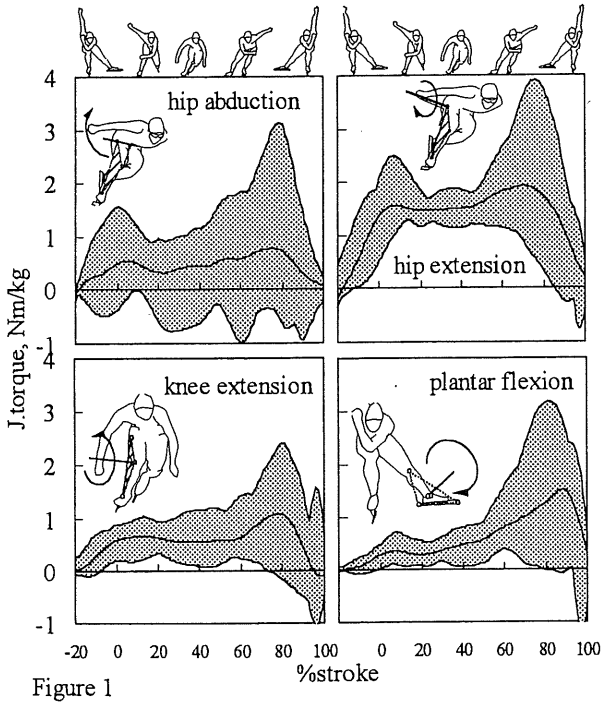



Figure 1
Mean of the joint torques around the hip abduction and extension, knee extension and plantar flextion axes.

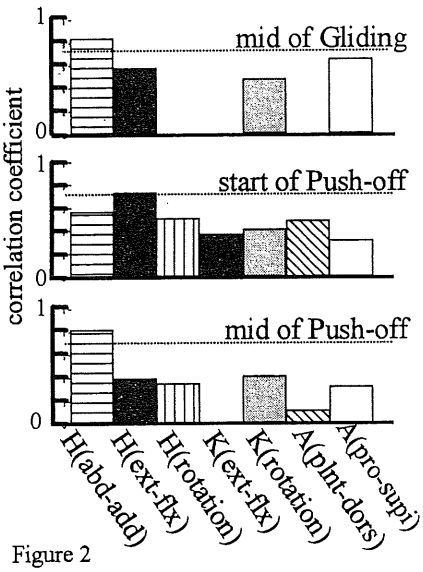



Figure 2
Correlation coefficients of joint torques to the horizontal component of BRF at the mid of gliding phase, start and mid of push-off phase. Broken lines indicate significant level at 1%.

# A BIOMECHANICAL STUDY OF DYNAMICS AND EMG IN GYMNASTICS: BACKWARD TO HANDSTAND

L. MOCHIZUKI; I.C.N. SACCO, A. FARO; A.C. AMADIO
Laboratory of Biomechanics of the School of Physical Education and Sport - University of São Paulo, São Paulo, Brazil.

## INTRODUCTION

In this paper, a dynamical and eletromyographical (EMG) analysis were applied to study a gymnastic movement, the backward roll to handstand. In order to improve and combine technical knowledge and biomechanical description and analysis of sport movements, dynamical approaches have been developed in this sense.

## METHODS

Two trained subjects performed this movement. In the EMG analysis, it was used bipolar surface electrodes on the motor point of right *m.gluteus maximus* and *m.triceps suralis* and full wave rectification. In the Dynamical analysis, it was used a strain gages force plate to collect the signal of vertical and horizontal ground reaction forces during the backward roll and the impulse to handstand. It was used a A/D convertor that simultaneously interfaced the two signals to a computer. The frequency rate for both procedures was 750 Hz.

## RESULTS AND DISCUSSION

The description of the vertical ground reaction force showed two different peaks during the execution of the backward roll until handstand. These two different peaks are related to the extension of hip and lower limb, and the extension of elbow and flexion of shoulder. The EMG signals demonstrated a different time duration of muscular activation, as we can se in the figure 1. In fact, postural corrections of body position during the movement may maintain a high muscular activation. A performance evaluation of sport movements is important to improve technical knowlwdge and to accurate biomechanical methodologies to permit the description, analysis and evaluation of sport movements.

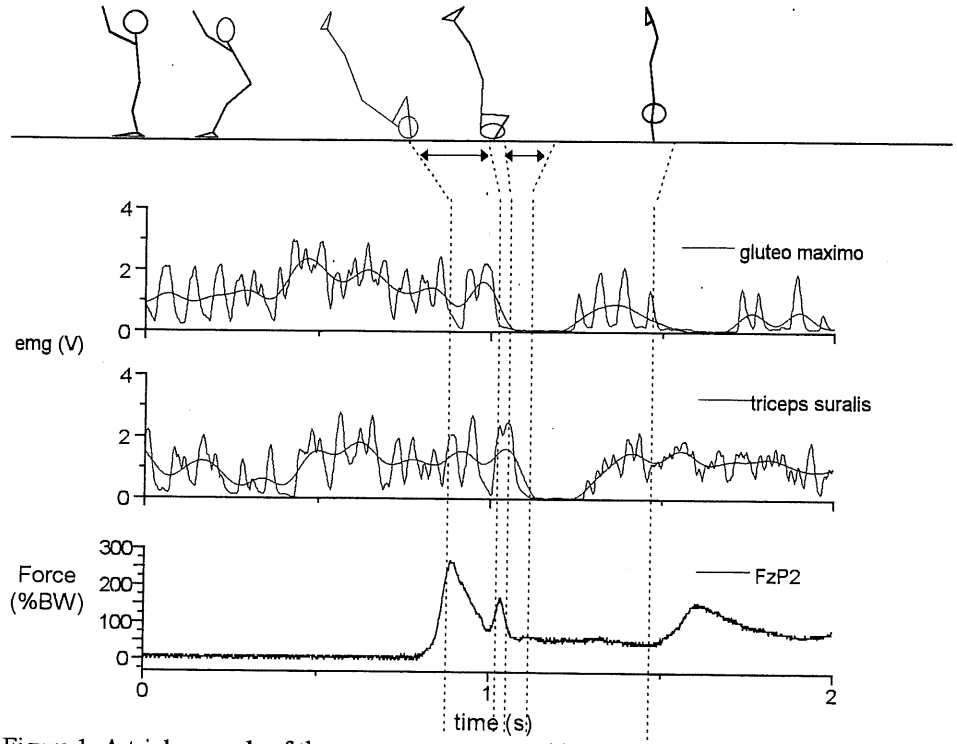



Figure 1. A trial example of the parameters measured in backward to stand.

# EVALUATION OF SYMMETRY IN SNATCH PULL TECHNIQUE

T. Isaka[1], A. Albrecht[2], M.M. Ryan[2], B. I. Prilutsky[2] and R.J. Gregor[2]

[1] College of Science and Engineering, Ritsumeikan University, Shiga, Japan
[2] Department of Health and Performance Sciences,Georgia Institute of Technology, Atlanta GA, USA

## INTRODUCTION

In Olympic weightlifting, proper mechanics during the pulling phase of both the snatch and clean and jerk are critical to successful performance. This bimanual task requires that muscles of each lower limb work synergisticly with muscles in the trunk and upper extremities. While most individuals have a dominant side with respect to their upper and lower extremities, skeletal development, muscle and neuromuscular function are generally considered to be asymmetrical. Athletes, therefore, whose events require symmetrical movements make a concerted effort to develop well-balanced nueromuscular function between both legs and arms. Many studies on the pull phase of the competitive lifting movement have focused on the kinetics and kinematics of lifting techniques (Baumann, 1988), but rarely discuss issues related to symmetry. An evaluation of symmetry during lifting provides information related to skill level and selected technical problems and offers potential feedback to athletes and coaches to prevent injury and enhance performance. The purpose of this study was to evaluate symmetry during the pull movement of snatch lift with respect to external forces (ground reaction force) and muscle activation.

## METHODS

Eight healthy male subjects participated in this study and each signed a written informed consent. Two subjects were skilled competitive weightlifters and six were football players with some experience in weightlifting with clean and high snatch lifts used in their weight training program. Each subject performed at least two trials at weights representing 50, 70, and 90% of their best records during the last year. Two high speed video cameras (60Hz) were employed to collect barbell displacement data for each subject during the snatch pull. Each foot of the subject was placed separately on a force platform (Kistler Instr.) for measurement of ground reaction forces under each foot during each lift. Analog force signals were amplified and sampled at 300Hz. In order to evaluate muscle involvement during the pull phase of the snatch lift, a pair of surface electrodes were applied to the skin overlying 15 separate muscles in the lower extremities and trunk of each lifter. Bilaterally, the gastrocnemius, vastus medialis, rectus femoris, semitendinosus, biceps femoris, gluteus maximus and erector spinae were sampled while the rectus abdominus was sampled unilaterally. A sixteen channel telemetry system (Konigsberg Instr.) was used in these experiments with signals from each muscle transmitted to a computer based receiver for storage and subsequent analysis and synchronized to the force plate and video records using a DC pulse. Signals were filtered (6db 20-450 Hz) and sampled at 1 kHz.

## RESULTS and DISCUSSION

Figure 1 shows the typical vertical component of the ground reaction force with the total vertical force divided into three phases. Enoka (1979) reported similar vertical force patterns during the pull phase of a clean movement while Garhammer and Gregor (1992) reported similar patterns during vertical jumping. In the unskilled subjects, phase II could not be identified in some trials because the vertical component of the ground reaction force did not go below the total system weight. The magnitude of the vertical forces, timing and vertical impulses were considered in our evaluation of lifting symmetry between the right and left sides of each lifter. Vertical force symmetries were calculated as follows: Fz symmetry index (%) = 100 - | (left - right) / right X 100 |, using peak Fz during each of the three phases during the 70% best weight snatch condition are presented in Figure 2. Unskilled subjects were less symmetrical in the Fz2 phase when compared to the Fz1 and Fz3 phase while skilled lifters were more symmetrical, overall, especially lifter JK during the Fz2 phase. These tendencies were similar in the 90% best weight snatch condition. Results from the timing and impulse symmetries indicate unskilled subjects were generally less symmetrical during the unweighting phase of the lift (Fz2), but the skilled subjects maintained similar values of symmetry during the weighting phases. During the unweighting phase, the skilled lifters utilized the double knee bend technique, a technique generally unfamiliar to their unskilled counterparts. During the unweighting phase, the skilled lifters displayed knee flexor and extensor synergy to more effectively control that phase of the lift. Certainly the training of the neuromuscular system, i.e. appropriate recruitment of motor units, in the two lifters more familiar with the double knee bend technique, influenced the kinetic output measures monitored in this experiment and resulted in the observation of greater symmetry in the skilled lifters in all three phases (Figure 2) with the greatest during the unweighting phase (Fz2) so critcal to a successful lift.

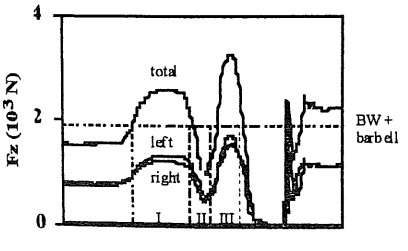



Figure 1. Vertical component of the ground reaction force. These phase are identified, Fz1 and Fz3 are weighing phases, Fz2 is the unweighing phase.

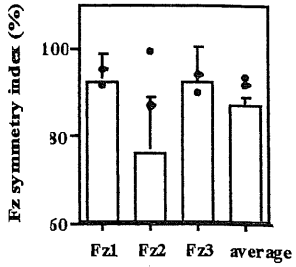



Figure 2. Fz symmetry index for the three phases and average of the 70% best snatch lift. 6 unskilled(□) and two skilled (OJ-◆, JK-●) weightlifters.

# A NEW APPROACH TO APPRECIATING THE MECHANICAL OUTPUT OF INDIVIDUAL JOINTS IN VERTICAL JUMP

A. Nagano, Y. Ishige and S. Fukashiro
University of Tokyo, Tokyo, Japan

## INTRODUCTION

To appreciate the mechanical output of each joint in vertical jumps, net joint torques and their powers have usually been calculated, then integrated into energies exerted by joints (ex. Hubley and Wells, 1983). In vertical jumps, the criterion of the performance is the jumping height, so output of each joint should be appreciated in terms of vertical components of vectors, not in terms of energy as a scalar parameter. The purpose of this study was to present a new approach that enables the calculation of the vector parameters in vertical jumps in order to estimate the output of each joint.

## METHODS

Eight male subjects performed vertical jumps from squat position with their knees flexed 90 degree. Kinetic data were recorded with a force platform (Kistler,9281b) at 1000Hz. Coordinates of five anatomical landmarks (Hubley and Wells, 1983) were filmed at 200Hz with a high-speed camera, and digitized automatically (NAC, MEMRECAM c2s, Image Express System). Momentum exerted by each joint was calculated, as described below. The vertical component of the ground reaction force was decomposed by means of relative acceleration of the mass center of each segment, such as

$\mathbf{F1}_{hip} = (\mathbf{m}_{HAT}) \times (\mathbf{a}_{HAT} - \mathbf{a}_{thigh})$,
$\mathbf{F1}_{knee} = (\mathbf{m}_{HAT} + \mathbf{m}_{thigh}) \times (\mathbf{a}_{thigh} - \mathbf{a}_{shank})$,
$\mathbf{F1}_{ankle'} = (\mathbf{m}_{HAT} + \mathbf{m}_{thigh} + \mathbf{m}_{shank}) \times (\mathbf{a}_{shank} - \mathbf{a}_{foot})$,
$\mathbf{F1}_{toe} = (\mathbf{m}_{HAT} + \mathbf{m}_{thigh} + \mathbf{m}_{shank} + \mathbf{m}_{foot}) \times (\mathbf{a}_{foot})$,
$\mathbf{F2}_{ankle} = \mathbf{F2}_{ankle'} + \mathbf{F2}_{toe}$

$\mathbf{F2}_{hip} = \max(0, \mathbf{F1}_{hip})$
$\mathbf{F2}_{knee} = \max(0, (\mathbf{F1}_{knee} + \min(0, \mathbf{F2}_{hip})))$
$\mathbf{F2}_{ankle'} = \max(0, (\mathbf{F1}_{ankle'} + \min(0, \mathbf{F2}_{knee})))$
$\mathbf{F2}_{toe} = \max(0, (\mathbf{F1}_{toe} + \min(0, \mathbf{F2}_{ankle'})))$

where **m** : the mass of the segment, **F** : the virtual force exerted by the joint, **a** : the absolute acceleration of mass center of each segment. Net joint torques, net joint powers, and their works were also calculated. The momentum exerted by each joint was calculated by integrating **F2** values for each joint through the push-off phase. Relative contributions of each joint determined by these approaches were compared.

## RESULTS AND DISCUSSION

Mean curves of **F2**s for three joints are shown in Fig.1. Percentage of the output of three individual joints are listed in Table1.

Table1. contribution of each joint.

| | Ankle | Knee | Hip |
|---|---|---|---|
| Momentum | 40.4(±5.0)% | 29.0(±2.2)% | 30.6(±4.9)% |
| Work | 38.4(±5.7)% | 25.2(±10.9)% | 36.4(±12.8)% |

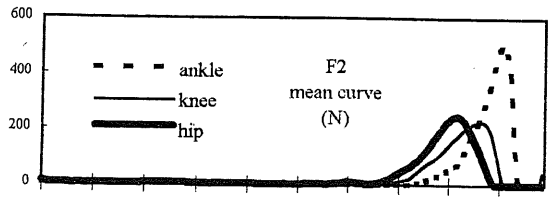



Fig.1.mean curves of F2. 1.0sec before take off is shown.

Both F2 curves and joint power curves show the peak output from hip, then knee, and lastly ankle. At hip and ankle, the percentages of contributions show different value between the two approaches, suggesting the necessity to select between these two approaches (with and without vector) according to the specific focus of the study. In particular, the approach introduced in this study handles one (vertical) component of the vector parameter. Unlike energetics, this approach can recompose vectors without losing any information about the direction. So, this approach is highly recommended in order to analyze translational parameters.

## REFERENCE

Hubley, C.L. and Wells, R.P. (1983) A work-energy approach to determine individual joint contributions to vertical jump performance. Eur. J. App. Physiol. 50, 247-254.

# DOES THE ISOMETRIC STRENGTH AFFECT THE VERTICAL JUMP PERFORMANCE?

J. Janiak, and J. Gajewski
Institute of Sport, Warsaw, Poland

## INTRODUCTION

Various jumping tests have been commonly utilized for evaluation of strength-velocity abilities of athletes (Bobbert et al., 1995). The maximal mechanical power developed in the take-off phase of the countermovement jump (CMJ) has been shown to be dependent on the attained height of jump, the countermovement depth and body mass (Dowling and Vamos, 1993, Gajewski et al., 1996). Also the maximal strength has been shown to depend on body mass. The aim of the present study was to clarify possible connections of the variables mentioned with the maximal isometric strength of lower extremity.

## METHODS

Two groups of untrained students (22 males and 47 females) took part in three different jumping tests on the Kistler piezoelectric platform interfaced with a PC XT microcomputer (1 - CMJ with arms swinging, 2 - CMJ with arms akimbo and 3 - SJ with arms akimbo). The height of jump (H1, H2, H3, respectively) and maximal power (P1, P2, P3, respectively) values for each jump were computed from the vertical ground reaction force. The recorded data were processed by a special designed computer programme (JUMP). The maximal isometric strength of knee and hip extensors of the left and right limbs separately (KL, KR, HL and HR, respectively) was tested on a special torquemeter device designed and constructed locally. The study was approved by the local Commission of Ethics. The repeated measures ANCOVA was employed to test the differences between mean values of jump or strength parameters. Body height was used as the covariate for the height of jump and maximal mechanical power while body mass was the covariate for strength variables.

## RESULTS AND DISCUSSION

Table 1
Mean (±SD) values of strength and jump performance variables

| | Male (n=22) | Female (n=47) |
|---|---|---|
| KR[N·m] | 389.1± 62.0 | 224.9±35.4 |
| KL[N·m] | 372.3± 59.5 | 230.4±43.5 |
| HR[N·m] | 630.5±117.6 | 375.0±74.4 |
| HL[N·m] | 649.0±160.0 | 417.3±89.5 |
| H1[cm] | 45.9±7.5 | 35.0±5.0 |
| H2[cm] | 39.6±5.5 | 31.5±4.6 |
| H3[cm] | 37.5±6.7 | 29.5±5.3 |
| P1[W] | 2452±759 | 1552±376 |
| P2[W] | 1813±493 | 1319±330 |
| P3[W] | 2211±598 | 1380±400 |

Mean values of the analyzed variables were presented in Table 1. Statistical analysis ($p<0.05$) revealed significant differences between variables describing the jumping performance in three different jumps and two groups of students. Effects of gender, jump and their interaction occured signifficant on height of jump ($F_{1,66}=15.3$, $F_{1,134}=122.8$, $F_{1,134}=6.6$, respectively) and maximal power ($F_{1,66}=8.7$, $F_{1,134}=55.8$, and $F_{1,134}=7.9$, respectively). Significant differences were also found for strength variables. Correlation analysis was utilized for evaluation of relationships between isometric strength and jumping performance variables. However, significant relation was found only for right knee extensors torque and maximal power developed in SJ in female group ($r=0.301$). Factors other than isometric strength, like muscle fiber composition (Aura and Viitasalo, 1989), performance technique and stretch reflex, were considered individual determinants of maximal mechanical power developed during various types of vertical jumps.

## REFERENCES

Aura, O. and Viitasalo, J.T. (1989) Biomechanical characteristics of jumping. Int. J. Sport Biomech. 5, 89-98.

Bobbert, M.F., Gerritsen, K.G.M., Litjens, M.C.A. and Van Soest A.J. (1995) Explanation of differences in jump height between countermovement and squat jumps. In: Book of Abstracts. XV ISB Congress, Jyväskylä.

Dowling, J.J. and Vamos L. (1993) Identification of kinetic and temporal factors related to vertical jump performance. J. App. Biomech. 9, 95-110.

Gajewski, J., Janiak, J., Eliasz, J. and Wit, A. (1996) Determinants of the maximal mechanical power developed during the countermovement jump. In: Abrantes J. M. C. S. (Ed.) XIV Symposium on Biomechanics in Sports. Proceedings. Edicoes FMH.1499 Lisboa Codex. Portugal, Funchal-Madeira, 420-423.

# A RESEARCH OF FORCES AND THEIR TIMINGS OF SKI JUMPING

Kaoru Yamanobe, K. Tamura and K. Watanabe

Laboratory of Physiology and Sports Biomechanics, Faculty of Education of
Hiroshima University, Hiroshima, Japan

## INTRODUCTION

The purpose of this study is to know the forces produced by ski jumper and to clear their relationship to the jumping performance especially the distance of ski jump.

## METHODS

The subjects were participants of a normal hill competition of FIS Continental Cup Asia Series NTT Cup Summer Jumping Games in Hakuba vil. Nagano pre. at 21th September 1996. The five force plates (2m length) that were installed to the jumping platform. Photocells were mounted parallel to a run way at 0m, 6m, 10m, 13m from the edge of the jumping platform to acquire the position of the jumper. The jumper's motion was filmed during last 6m by high-speed camera at 200fps. The force analysis included the Fz peak generated 6-0m before the edge of the takeoff platform. The averaged force levels(N) were calculated during the following distances: F'1:6-4m, F'2:4-2m, F'3:2-0m,F'4:6-0m,F'5:4-0m before the edge of the takeoff platform and their relative values as to the total body weight (N/kgBw). The timing of Fz peak from the edge of the takeoff platform defined as T1 , the timing of the force curve coming down to the level of the jumper's total body weight as T2. With respect to T2, we used absolute value T2abs because some were before and others after the edge of the takeoff platform. Further investigation was made to compare 16 longest and 16 shortest jumps at each round.

## RESULTS

In 1st round the relative force level was correlated to the distance of jump. The Fz peak was also correlated to the distance of jump(table1). In 2nd round the relative force level was also correlated to the distance of jump(table2).

**Table1: Correlation between variables and distance at 1st round**

| | | |
|---|---|---|
| Total body weight(kg) | r=-0.392 | ** |
| T2abs(msec) | r=-.331 | * |
| F'2(N/kgBw) | r=.279 | * |
| F'4(N/kgBw) | r=.440 | ** |
| F'5(N/kgBw) | r=.294 | * |
| Fzpeak(N/kgBw) | r=.357 | * |

**Table2: Correlation between variables and distance at 2nd round**

| | | |
|---|---|---|
| Official velocity(m/s) | r=.357 | * |
| F'4(N/kgBw) | r=.387 | ** |

n=50

* p<.05
** p<.01

## DISCUSSION

We could have the data that support the length of ski jump may depend on the relative force generated during the phase of 6-0m before the edge of the takeoff platform. In addition, the pattern of the force curve was very different even in the excellent jumpers.

# Motion analysis at clearance the fence during the show jumping

K. Terada and A. nagata K. Akihiro
Waseda univ. Tokyo, Japan

**Introduction**

The purpose of this study was to find out the horse movement directly before jumping by the kinematics method. In order to know what way horse do before jump, we can provide for his jumping movement and jumping actions to record their physique experimentally.

**Methods**

Two thoroughbred horses (A, B) were subjected to be filmed using four high speed video cameras. Cinematographical analysis were planned to use horses on two experimental conditions with or without a rider. Experimental parameter were selected as following stems : trajectories of horse's head, displacement of right and left fore-limb, angle of rider's elbow and horse's neck, and angle of horse's leading fore-limb against ground.

**Result**

Vertical velocity of horse's head decreased before time when the trailing fore-limb was landed directly, and then these velocities increased after time when the leading fore-limb was landed. when the trailing fore limb was landed, horizontal velocities of the horse's head decreased. whereas, these velocity increased, when the leading fore-limb was take off before this jumping (Fig) . Broad distance between the light and the left fore-limb at one stride were shown to be much more shorter than that at two or three strides before jumping motion. The angle of horse's leading fore-limb against ground increased significantly comparing with general galloping motion, when those fore-limb landed in the ground. While those angles decreased remarkably when these fore-limb take off the ground.

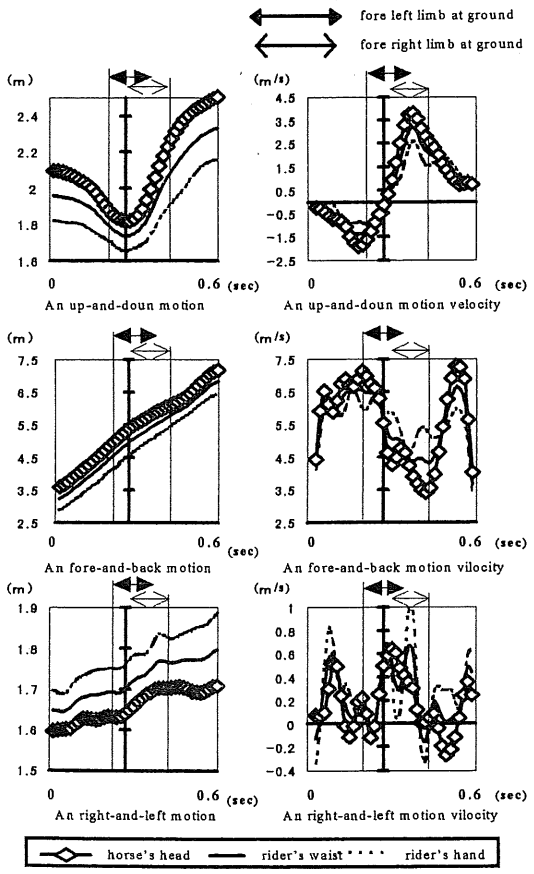



fig Motion and motion vilocity of horse's head , rider's waist and rider's hand which direction is up-and doun, fore-and-back and right-and-left.

**Discussion**

These twice results that vertical velocity of the horse's head decreased before time when the trailing fore-limb was landed, and then increased after time when leading fore-limb was landed would indicate that the vertical velocity of the horse's head was accelerated at the first time, and at the following period , horizontal velocity increased to clearance the fence with different timing from general galloping motion. In other words, the horse change toward direction of the jumping power forward to upward using of one's head movement.

These data should suggest that the horse's head had controlled mainly before jumping movement against the fence as preemptory one. And also, the rider's movement would have to match with those horse's one before and during jumping at those time sequence.

# Leg Length Determines Ankle Speed in Planar Soccer Instep Kick

Masanori Isokawa and Takeji Kojima*
Tokyo Metropolitan University, Tokyo, Japan and the University of Tokyo*, Tokyo, Japan

In soccer kicks with the objective of maximal speed, there is a close correlation between the speed of ankle of the kicking leg and ball speed. The purpose of this study was to investigate the main determinant of the ankle speed in a planar soccer instep kick.

## METHODS

Six soccer players of a professional team (A group), six substitute soccer players of another professional team (B group ) and five soccer players of a university (C group) participated in this study. The subject kicked a stationary ball three times for maximum ball speed using sagittal plane motion of the kicking leg with several steps of running approach. Each kick was filmed at 300 fps with a 16 mm cine-camera positioned perpendicular to the sagittal plane. The kick which produced the highest ball speed was selected for further analysis for each subject. Coordinates of the greater trochanter, lateral condyle, lateral malleolus of the kicking leg were determined using a film motion analyzer and smoothed with a Butterworth digital filter. The following variables were determined from the smoothed data : lower limb length (i.e. sum of the lengths of the thigh and lower leg), relative velocity of the lateral malleolus to the greater trochanter, and angular velocity of the knee joint.

## RESULTS and DISCUSSION

Mean relative velocities of the ankle joint just before impact for A, B and C groups were 13.8±0.8 m/s, 14.2±0.8 m/s and 13.6±0.7 m/s, respectively. The difference in the mean velocity among the groups was not statistically significant ( $F = 0.82$, $p > 0.05$). The relative velocity of the ankle joint was significantly correlated with the lower limb length (Fig.1, $r=0.86$, $p < 0.05$). The angular velocity of the knee joint just before impact was not significantly correlated with the length (Fig.2, $r=0.02$, $p > 0.05$).

These results suggest that for the experienced soccer players used in the present study the lower limb length is the main determinant of the relative velocity of the ankle joint (Fig.1) and that effects of the muscular properties and technique on the ankle speed might be small in this kind of kick.

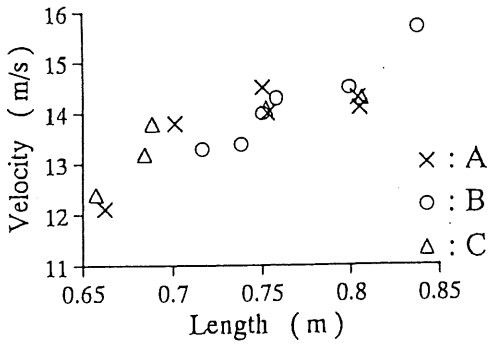



Fig.1 Relationship between the relative velocity of the ankle joint and the lower limb length.

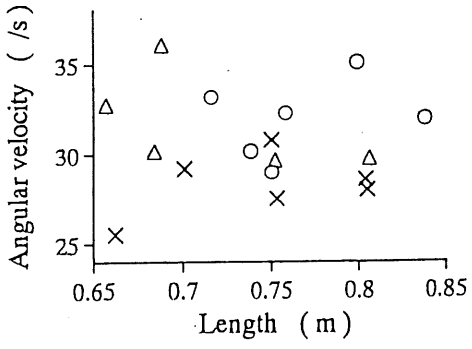



Fig.2 Relationship between the angular velocity of the knee joint just before impact and the lower limb length.

# BIOMECHANICAL ANALYSIS OF THE WUSHU JUMP-SLAP-KICK

Chung-yu Chen, Chenfu Huang, Di-yu Chen*, & Jen-ping Tang+
National Taiwan Normal University, Taipei, Taiwan
*National College of P.E. & Sports, Taoyuan, Taiwan
+Chinese Culture University, Taipei, Taiwan

## INTRODUCTION

Wushu is the good character of the traditional Chinese culture. It is also one of the event in the Asia Game. The basic movement of the Wushu "jump-slap-kick" (Fig1.) is the important jumping action in Wushu competition. The height of center of mass (CM) during the jump-slap kick is the important criteria which may decide who can win in the competition. The purpose of this study is to analyze the biomechanical characteristics of the Wushu jump-slap-kick.

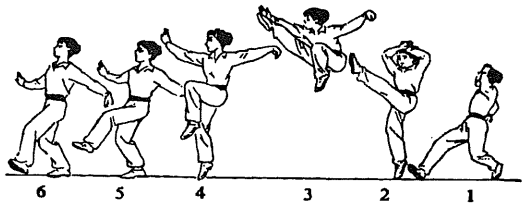



Fig1. Jump-slap-kick

## METHODS

Seven elite Wushu athletes served as the subjects for this study. Their mean age, height and body mass were 20.4±2.7 years, 167.9±5.5 cm, and 62.4±4.7 kg respectively. One high speed camera (120Hz) was used to videotape the movement and the Peak Performance System was used to digitize and analyze the data. Person-Production of correlation was used to test the selected kinematic variables with the jumping height of CM at 0.05 significant level.

## RESULTS AND DISCUSSION

The following variables have positive correlation with the jumping height of CM: the vertical velocity of CM at takeoff (4.64 m/s), the angle of CM at takeoff (62.5°) , the maximum trunk angle during the takeoff (117.7°), and the vertical velocities of the knee (5.7 m/s), ankle (7.87 m/s) and toe (9.23 m/s) of swing leg at takeoff. There was negative correlation between the jump height of CM and takeoff time(0.252s). Fig2. Showed that when the CM at the lowest point the takeoff leg continued to flexed the joints and in the same time the vertical displacements of the trunk and the swing leg (fig 3) also went up.

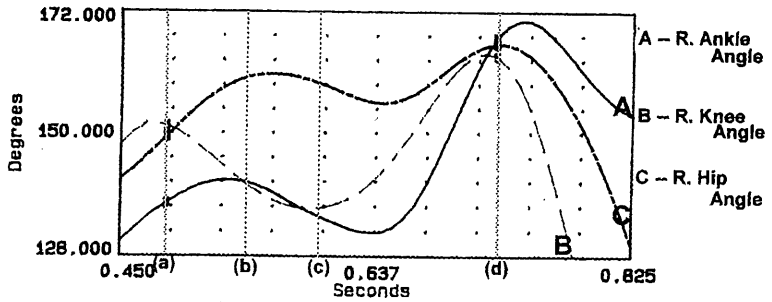



Fig2. Joint angle of the takeoff leg.

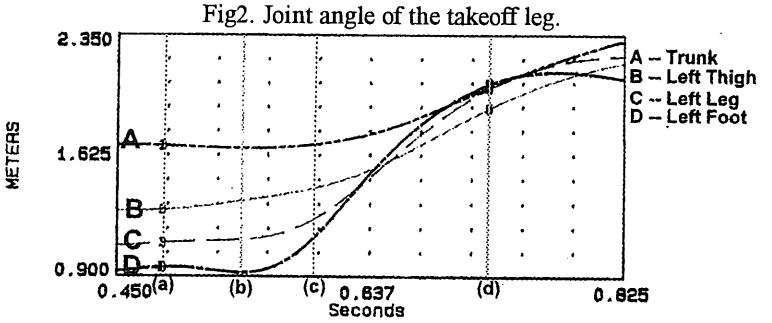



Fig3. Vertical displacement of the trunk and swing leg.
(a) touch down (b) lowest point of CM (c) max. flexion of knee (d) takeoff.

The high correlation between the jumping height of CM and the vertical velocity of CM at takeoff and takeoff time is in agreement with the high jump studies. It is suggested that the Wushu player should rapidly swing the leg up at takeoff in order to increase the jumping height.

# BIOMECHANICAL ANALYSIS OF TWO DIFFERENT FLYING-KICKS OF CHINESE MARTIAL ARTS

Chenfu Huang, Chung-yu Chen, Ti-yu Chen*, & Jen-ping Tang**
National Taiwan Normal University, Taipei, Taiwan
*National College of P.E. & Sports, Taoyuan, Taiwan
**Chinese Culture University, Taipei, Taiwan

## INTRODUCTION

In traditional Chinese martial arts, stationary flying-kick (SFK) and one-step flying-kick (OFK) are two important skills which are required by the Wushu athletes. The purpose of this study is to analyze the biomechanical characteristics of the two different flying-kicks.

## METHODS

One elite martial arts athlete (25 years, 1.75 m, and 89 kg) served as the subject for this study. One camera (60Hz) was used to record the movement and the Peak Performance System was used to digitize and analyze the data. A Kistler force platform (9287, 600 Hz ) was used to collect the ground reaction data. The NORAXON telemetred EMG system was used to collect the right leg (takeoff leg) six muscles action (gluteus maximus, rectus femoris, vastus medialis, biceps femoris, gastrocnemius, and soleus).

## RESULTS AND DISCUSSION

The results indicated that the OFK had the greater jumping height of center of mass (CM) and a shorter support time (0.641 m, 0.42 s) than the SFK (0.498 m, 0.64 s). The OFK also had the greater peak vertical ground reaction force (2.42 BW) than the SFK (2.09 BW) during the support phase. The hip (52%), knee(60%), and ankle (60%) of the OFK had longer joint flexion time (figure 1) during the support phase than the SFK (hip, 28%; knee, 36%; ankle, 46%). The result of the EMG signal (figure 2)of the SFK and the OFK indicated that the bicepts femoris was more active at the beginning of the action which was responsible for extend the hip joint and bring the body forward. Other five muscles EMG signal were more active before the takeoff.

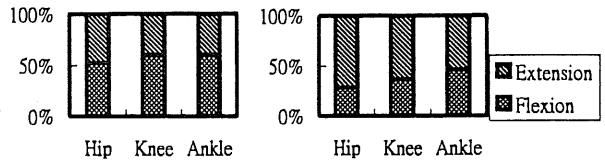



Figure 1. Percentage of joints movement of OFK (A) and SFK (B)

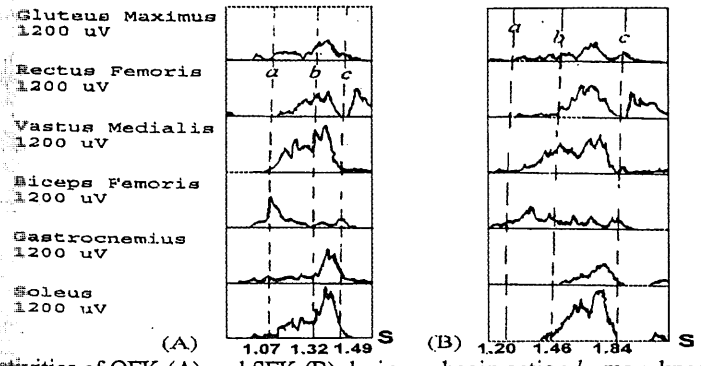



Figure 2. EMG activities of OFK (A) and SFK (B) during *a*. begin action *b*. max. knee flexion *c*. takeoff

Most Wushu coaches suggested that when performing the SFK, the athletes should not flex the lower limb joints after action begun. The results of this study indicated that the flexion of the joints in the SFK and the OFK was to bring the swing leg and body forward. The results provide valuable information for the teachers and coaches when teach SFK and OFK.

# THE IMPACT EFFECT OF THE CHINESE MARTIAL ARTS FORWARD-KICK

Yu Liu & Long-Ren Chuang
Dept. of Physical Education, Chinese Culture University, Taipei, Taiwan

## INTRODUCTION

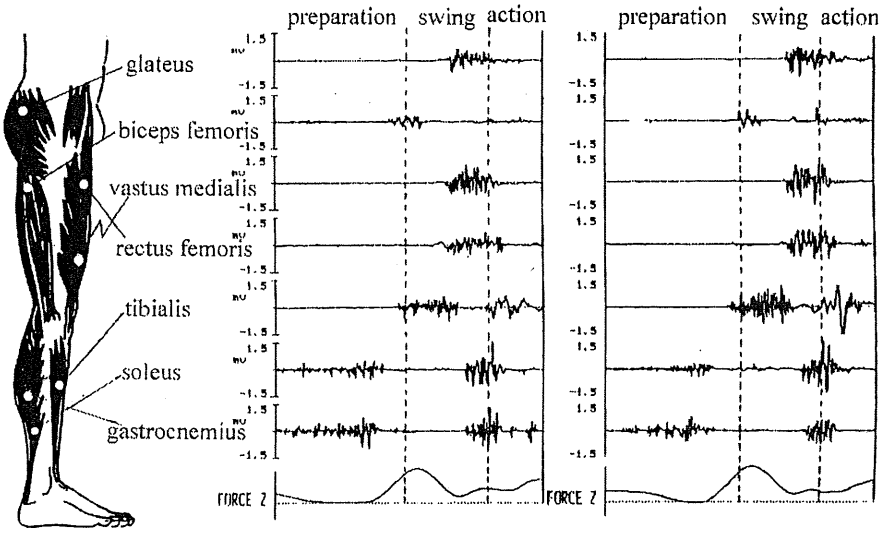
The forward-kick (FK) occupies an important position in the leg techniques of Chinese Martial Arts (CMA). Previous studies of Taekwondo kicks and CMA kicks focused on the description and comparison of the existing basic kicks with regard to kinematic and kinetic variables (Park, 1992; Liu,et al 1995).The purpose of this study was: i) to investigate the influencing factors on the impact effect of FK with regard to kinematic, ground reaction force and EMG data; ii)to go further into the principals of the kick effect of CMA.

## METHODS

Ten CMA students (height: 172±6.8cm; weight: 71.02±11.4kg; age: 24.3±3.9) performed FK from a standing ready position and aimed toward the target (punching bag). This performance can be divided into three phases: i) preparation, ii)swing and iii)action. Peak Performance video Motion Analysis System (60Hz), AMTI Forceplatform (sample rate 600Hz) and NEC DP1100 Signal Processor was used to synchronously record the movement, the ground reaction force on supporting leg and the EMG signal of 7 muscles (1-glateus;2-biceps femoris;3-vastus medialis;4-rectus femoris;5-tibialis;6-soleus;7-gastrocnemius) in the kicking leg. Pearson-Production correlation between the kinematics and kinetic parameters and the velocity of the punching bag (immediately after foot contact)was analyzed using SPSS/PC+ software.

## RESULTS

The velocity of BCG reached its maximum before contact with the punching bag. Hip, knee and ankle joint angles continuously extended throughout action phase, showing that the leg was continuously to acting on the punching bag after contact. The smaller the angle of the knee flexion during the kicking, the greater the velocity of the punching bag after foot contact ($r=-.63$, $p<.05$). On the kinetic parameters, a significant correlation was found between the maximum of ground reaction force, which was reached before contact with the bag, and the velocity of the bag ($r=.67$, $p<.05$). The result of EMG displayed that the major activities of the muscles occurred not in the action phase but in the swing phase (s. Fig.). This phenomenon is very helpful for us to understand the impact effect of FK.




## DISCUSSION

Associate the kinematic and kinetic analysis with the EMG results we can summarize that the maximum velocity of the body, the maximum ground reaction force and the major activities of the muscles was occurred before the foot contact. The attack effect of the FK depends, therefore, mainly on the momentum of the body rather than the active producing the contact force through extension the joints. From our experiment, it can be concluded that in order to utilize the body movement created previously and to increase the impact momentum, striking valid mass and firmness of the leg at the moment of contact is more valuable in FK than foot velocity.

## REFERENCES

Park, Y.J.(1989). A Biomech. Analysis of Taekwondo Front-Kicks. UMI Dissertation Services.
Liu, Y., Chuang, L. R. & Lian, D.H. (1995). Biomech. Analysis of the Chinese Martial Arts Forward Snap Kick and Forward Hell Kick. Research on Chinese Martial Arts. Taiwan

# THE SECOND IMPACT FORCE OF TAI CHI FIST PUNCHES

LONG-REN CHUANG, NORMAN YANG, SAMPSON CHEN, CHIA-JEN CHUANG
Dept. of Physical Education , Chinese Culture University, Taipei, Taiwan

## INTRODUCTION

Now in ever increasing numbers scholars and researchers have begun studying the striking movements of martial arts; Boxing (Atha 1985, Whiting 1988), Karate (Yoshihuku etal. 1984, 1985, 1987; Cavanagh 1975; Walker 1975) and Chinese martial art Shaolin (Chuang 1991, 1992). But up to now there has been no study on the striking movements of the Chinese martial art Tai Chi. During normal Chinese martial arts body conditioning (toughening of the body through regulated strikes to the body), normal strikes to the body were bearable and of single impact force. Students from the Rue-Yue Tai-Chi Academy seemed to be able to create an additional second impact force in a single punch. Therefore we devised this study with the goal of proving that there is indeed a second impact force.

## METHOD

We asked Eight Rue-Yue Academy students and Eight Shaolin stylists to volunteer as test subjects. Because Rue-Yue Academy's Tai Chi punch movement requires a longer follow-through distance, normal stationary impact plates were unsuitable. So to gain the best impact results, we constructed a structure similar to a pull up bar with two hinged bars connected to a punching plate. We recorded the impact force with an F-scan pressure sensor taped to the surface of the target punching plate. While kinematics analysis was carried out of the with the Peak Performance movement analysis system and a Panasonic 510HS video camera.

## RESULT

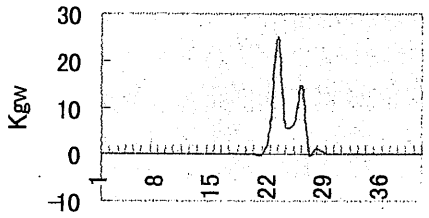



Figure A. Rue-Yue Academy Tai-Chi punch impact force.

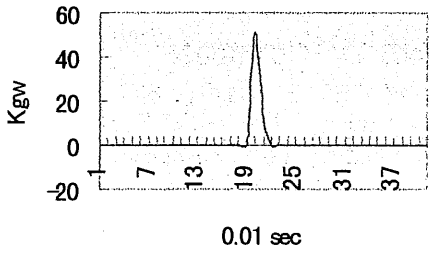



Figure B. Shaolin stylists punch impact force

1.The velocity curve of the arm showed two peaks. One peak was reached before impact, and another smaller peak after impact. Shaolin style punching resulted in a single peak before impact.
2.The F-scan pressure sensor detected two force peaks in the Rue-Yue punching impact, with the first one being greater. The Shaolin style punching impact had only one force peak.

## DISCUSSION

Our comparison of the results from different subjects, and the control test of a thrown shot-put has shown that the Rue-Yue Academy's Tai Chi students do create a 2 peak impact force. In other published works Yoshihuku(1985) had also published about a similar 2 peak impact force curve in a Karate punch. In his opinion, the second peak was induced by the body's moment. In our study we have yet to explain this phenomenon, but will investigation this mechanism in following research.

## REFERENCES

Cavanagh, P. R. & Landa J. (1975). A biomechanical analysis of the Karate chop. Research Quarterly.47(4).
Chuang, L. R. & Hong, D. M. (1992). Comparison and analysis of upper limb kinematics between various steps and trunk postures during the Bow-Step straight punch. Research on Chinese Martial Arts.1(1). Taiwan.
Walker, J. D. (1975). Karate Strikes. American Journal of Physics. 43(10)
Yoshihuku, Y. (1984). Impact forces of various kinds of fighting techniques. Jpn. J. Sports Science. 3(6). 485-491.
Yoshihuku, Y. (1985). The science of the greatest fighting technique. Hukushou Dou publishing co. Tokyo.
Yoshihuku, Y. (1987). Impact forces of fighting techniques. Jpn. J. Sports Science. 6(4). 252-259.

# A BIOMECHANICAL STUDY OF KARATE STRIKES

Akira SHIBAYAMA and Senshi FUKASHIRO

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

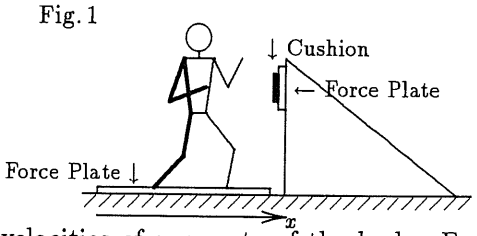
## INTRODUCTION

The basic techniques of karate (a traditional Japanese martial art) are striking, kicking, blocking, etc. The purpose of this study is to examine the physical aspects of a karate strike (*gyakuzuki*) through the use of force plates and a high speed video camera.

## METHODS

Seven healthy male subjects (Karate Club members of Tokyo University) volunteered for this experiment. Floor reaction forces and impact forces of *gyakuzuki* were measured with two force plates (Kistler). The subjects were asked to stand on a force plate and to strike a hard cushion on the other force plate with maximal effort (see Fig. 1).

Fig. 1




A high speed video camera was used to determine the velocities of segments of the body. Front and side views were recorded simultaneously by means of a large mirror placed to one side of the subject at about 45° to the line of sight. The three-dimensional trajectory was then reconstructed using the DLT method. Distance to the target and width of stances were left to the preference of each subject.

## RESULTS AND DISCUSSION

Typical results are shown in Fig. 2, 3 and 4. Fig. 2 shows the floor reaction force of the $x$ component (direction to the target). The force is initially positive to accelerate the body but turns negative before impact (dotted line) and decelerates the body. This can be considered as a tactics to stabilize the body. Also, it may help to accelerate the upper part of the body*. It is a matter of course that if this negative force is developed too early and too excessively, it will cause a negative effect, i.e., reduce the impact force.

The relationships between the impulse of impact force (from beginning of the contact to 0.01 sec after the force reaches its peak) and the momentum of the arm or of the body are shown in Fig. 3 and Fig. 4. These figures indicate that the initial phase of the impact is mainly composed by the arm momentum lost at impact and that the total body momentum contribution to this phase is small. An identical line is shown in Fig. 3 (dotted line). The distance between this line and the plotted points may be attributed to striking techniques.

Furthermore, the shape of the waveform of the impact force differs between the subjects. This cannot be explained by collision between a single particle and a damper. This phenomenon shows that the way of transmission of momentums (of parts of the body) through the fist depends on the techniques of the subjects.

Fig. 2

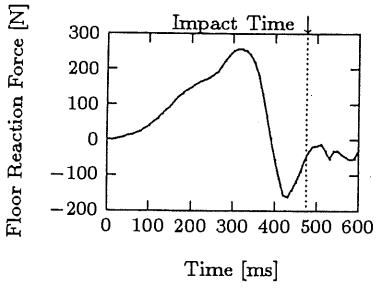



Fig. 3

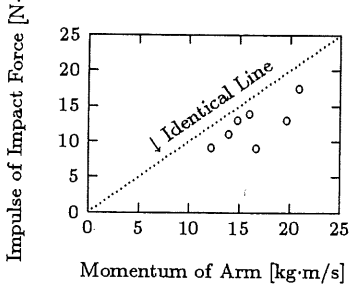
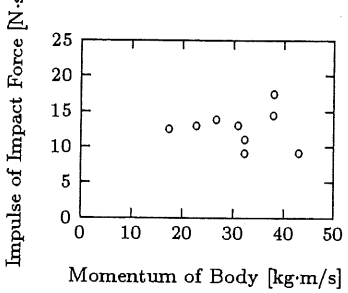



Fig. 4




*Let us suppose that a rigid rod is moving forward (the direction is orthogonal to the rod). When one end of the rod is stopped by an external force suddenly, rotational movement occurs and the velocity of the other end increases although the velocity of the center of mass decreases.

# A STUDY OF THE RELATIONSHIP BETWEEN THE WEIGHT OF SHINAI AND STRIKING MOTION IN ELEMENTARY SCHOOL AGED KENDOISTS

Kensaku Miyamoto,Shinichi Yamagami
Kagawa University
Japan

## INTRODUCTION

Kendo is one of the sports that is originated from japanese taraditional martial arts and is defined as a conducted by use of the shinai(bamboo sword).Shinai manipulation is very important to the technique of kendo.But,there is no rule about the standard and selective method of shinai in elementary school aged kendoists.The aims of this study are to seek the relationships between the weight of shinai and striking motion in elementary school aged kendoists.So we want to think about the standard and selective method of shinai in elementary school aged kendoists.

## METHODOLOGY

6 boys(6th grade:3 persons,2nd grade:3 persons) were perticipated. The striking motion were filmed by means of two high speed video camera at 100 frames/sec.Selected kinematic parameters such as the time required,velocity of shinai and center of gravity,shinai angle and joint angles were computed.The weight of shinai sumple has 3 types(mean value,m.v.-s.d.,m.v.+s.d.).This study was especially intended to compare among the timing of forward step in motion and shinai stroke.

## RESULTS AND DISCUSSION

Figure 1 showes time differences of the shinai manipulation and the step in motion.High level,6th grade kendoist(6A,6B) and 2nd grade kendoist(2A) adjusted the timing of the step in motion of right leg to the swing motion of shinai when the weight of shinai was heavy.But middle level,6th grade kendoist(6C) and lower level, 2nd grade kendoist(2B,2C) couldn't adjust to them.And Maximum velocity of shinai of 6th grade kendoist(6A,6B) was larger than that of 2nd grade kendoist(2B,2C) when swung up the shinai.The timing of swing up and down of shinai was change at the difference in weight of shinai on 2nd grade kendoist(2B,2C).The influence of weight of shinai on the striking motion was verified in the 2nd grade kendoist more than the 6th grade kendoist in elementary school.From this findings,the weight of shinai in the lower grade or beginner had better be a little more light.

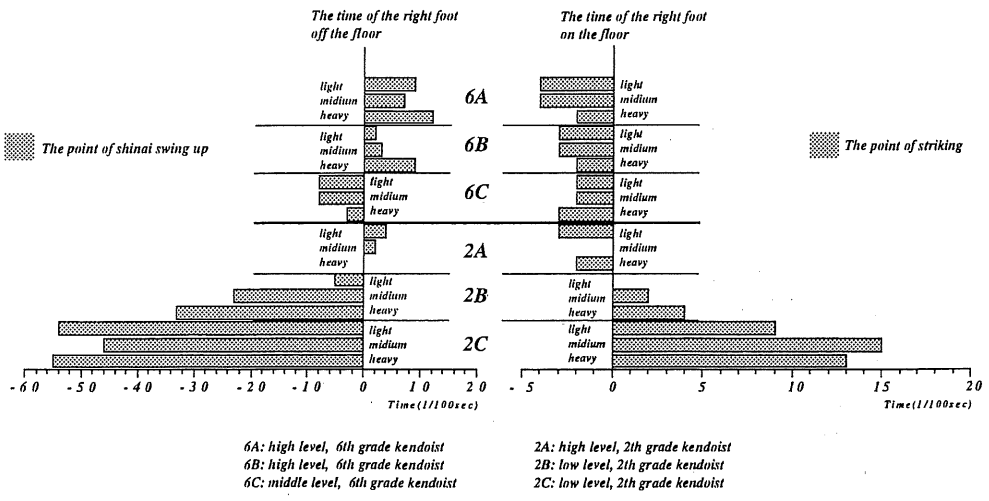



***Figure 1. Time differences of the shinai manipulation and the step in motion***

# A BIOMECHANICAL CHARACTERISTICS ON THE STRIKING MOTION OF HIGH TRAINED AGED KENDO MAN

Shinichi Yamagami,Fuminori Nakiri,Naoya Yokoyama,Kensaku Miyamoto
Kagawa University,Tokyo University of Agriculture and Technology,
Yokohama National University ,Japan

## INTRODUCTION

Kendo is one of the striking sports that was derived from Japanese traditional martial arts and is able to practice at the advanced age.So the striking technique of the high trained and aged kendo man is of great interest to us.There have been many studies on the biomechanics of kendo.However,there were few researches on the characteristics in striking motion of high trained aged kendo man.The purpose of this study was to investigate the biomechanical characteristics on the striking motion of high trained aged kendo man.

## METHODOLOGY

High trained aged kendo man(82 yrs,9dan degree) and university kendo man(21yrs,4dan degree) served as subjects of this study.The shomen-uchi motion was investigated.Force components along X,Y,and Z axes of step in motion with right foot were measured by use of a forceplate,the form of the motion were filmed utilizing two VTR camera(60frame/sec),an electromyogram(EMG) activities of both leg muscles(Rectus femoris,Biceps femoris,Tibialis anterior,gastrocnemets) were recorded using bipolar surface electrodes,simultaneously.

## RESULTS AND DISCUSSION

Compared with university kendo man,High trained aged kendo man can execute effective shinai(banboo sword) manipulation like a whip motion(see Figure 1).And during the shomen-uchi motion,smaller positional difference of the center of gravity in vertical direction was found in high trained aged kendo man than university kendo man.Because his trunk motion in the process of the whole motion showed such small wavy locus as going down gradually at the shinai raising and going up to the final stage of the stroke.In addition,he hits an object with small lean-forwarded position.In other words,he slides both foot forward like a walking step motion.SecondIy,it was found that high trained aged kendo man used rectus femoris of the right leg and gastrocnetmets of the left leg efficiently during the forward step in motion.The mean values of the maximum force components along X,Y and Z axes of right foot force of high trained aged kendo man are smaller than those of university kendo man.In conclusion,striking motion,especially during the shinai stroke and forward step in motion, high trained aged kendo man was composed of the principle in walking motion.From this findings,the technical characteristics of high trained aged kendo man is considered to be effective for lasting a long kendo practice.

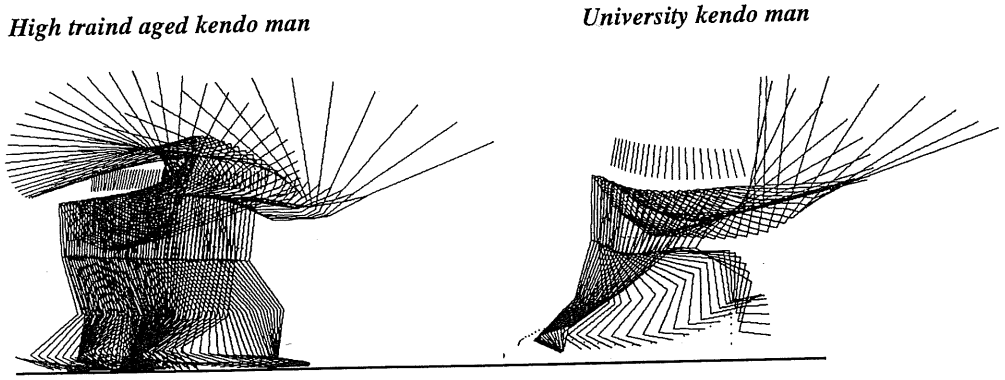



***Figure 1. Stick figure during the striking motion***

# DISCHARGE PROPERTIES OF HUMAN MOTOR UNIT DURING SUBMAXIMAL VOLUNTARY CONTRACTION

KAMO, M. and MORIMOTO, S.
Yokohama National University., Yokohama, Japan

## INTRODUCTION

During submaximal voluntary contraction, the behavior of the spike interval change seems to be not yet settled fundamentally because the results were not in agreement between investigators. In vastus medialis muscle, we observed the initial elongation in mean spike interval (MSI) in all tested motor units during the prolonged contraction at target torque of the recruitment threshold (Tth) of the objective motor unit (Kamo & Morimoto, 1990). In this case, motor unit would fire (or recruit) at lower excitability because of no difference (or little difference) between the target torque and Tth. It was insufficient experimental condition to discuss the motor unit activity at wide range of the torque above Tth. In present study we attempted to investigated the motor unit activity at the target above the Tth (≦ 10%MVC). And we would examine one of factors regulated the motor unit discharge pattern.

## METHODS

The subjects were three healthy adult volunteers aged 20-45 years. Action potentials of single motor unit were recorded from vastus medialis muscle using bipolar fine wire electrode unless otherwise stated. The subjects were requested to sit on a high stool . To record the developed torque at the ankle, the right lower leg was fastened with a strap connected to a strain gauge at articulatio pedis. The angle of the knee joint was set at 90° . The present target torque was 2, 4, 8 and 10%MVC. The subjects were requested to increase the developed torque at a gradient of 1%MVC/s linearly to the target and then sustain the target for 15min.

## RESULTS

Measurement could be made completely on 56 motor units. At the target torque of 2, 4, 8 and 10%MVC, 10, 14, 14 and 18 motor units were analyzed, respectively.

(1)Tested motor units represented the gradual increase in MSI during initial several minutes from the moment of onset of the developed torque to the target.

(2)Discharge behavior of motor unit was characterized and was divided into three pattern from the degree of the initial elongation and the subsequent behavior as follows: 1)Decuruited, 2)Decuruited but recruited again (re-recruited) with a pause and 3)Continuously recruited through the contraction. The present result was similar to the discharge pattern of motor units reported previously (Kamo & Morimoto, 1990). Additional, these three discharge pattern could be seen commonly in all experimental condition, independently to the target torque. To test the reproducibility of the discharge pattern obtained on different day, we employed the surface recording method to identify the single motor unit (Morimoto et. al.,1980). We obtained the high reproducibility in the motor unit discharge pattern.

(3)Tth of motor units tested in the present study distributed in a wide range between nearly 0%MVC and 9.3%MVC. However, there was no distinct relationship between the discharge pattern and the Tth. And the order of re-recruitment of motor unit was not necessarily the same of the recruitment at the beginning of contraction. The MSI at the in start of constant contraction did not influenced the discharge pattern.

(4)Recruitment of newly motor units could be seen at any time during contraction. But it was rare to observe the newly recruitment of the motor unit during initial elongation phase.

(5)When submaximal electrical stimulation was add on quadriceps femoris muscle through femoralis nerve , the evoked torque was maintained constant value for 3min at the stimulation frequency of 10Hz. When the stimulation was synchronized to the motor unit discharge pattern (spike intervals), the evoked torque was decreased with the elongation of spike interval.

## DISCUSSION

During the submaximal prolonged contraction, motor unit discharge pattern could divide into three pattern independent to the developed torque (or target torque). At least in the present condition, therefore, it suggests that the strategy on motor unit discharge is fundamentally same to regulate the constant torque development submaximally. Electrical stimulation applied to the muscle bundle (not to motor unit) appeared to decrease in the torque development at the stimulating frequency of a real motor unit discharge pattern. And there was the newly recruited motor units for the initial elongation of spike interval. These results suggest that the initial elongation of the spike interval was not originated in the change of the contraction property.

In the low level of muscle contraction employed in the present experiment, individual motor unit presented the individual discharge pattern. we could not pursuit the factors to decide the discharge property. Until now, we considered the factors from resting and/or control conditions. In further study, it is necessary to investigate peripheral (including muscle contractile properties) and central factors which might be modified with time.

## REFERENCES

- Kamo, M. and Morimoto, S. Discharge pattern of single motor units during prolonged activity at the tension of the recruitment threshold and its conduction velocity. Jpn. J. Physiol. Fitness Sports Med., 39, 298-306, 1990
- Morimoto, S, Umazume, Y. and Masuda, M. Properties of spike potentials detected by a surface electrode in intact human muscle. Jpn. J. Physiol., 30, 71-80, 1980

# FRACTAL DIMENSION OF SURFACE EMG DURING ISOMETRIC CONTRACTION IS NOT RELATED TO MUSCLE ACTIVATION, BUT TO MUSCLE FATIGUE

Minoru Shinohara, Motoki Kouzaki, Takeshi Yoshihisa, and Tetsuo Fukunaga
Laboratory of Sports Sciences, Department of Life Sciences, The University of Tokyo, Japan

## INTRODUCTION

Fractal analysis of surface EMG during isometric contractions has shown that fractal dimension (FD), a measure of "complexity", of EMG is linearly related to muscle activation (Anmuth, et al., 1994). Their results, however, might be due to the intrinsic characteristics of the fractal analysis method employed (Katz, 1988), *i.e.*, amplitude of the signal will strongly affect FD. We hypothesized therefore that other method independent of signal amplitude (Higuchi, 1988) would derive different results, and FD would be related to muscle fatigue rather than muscle activation.

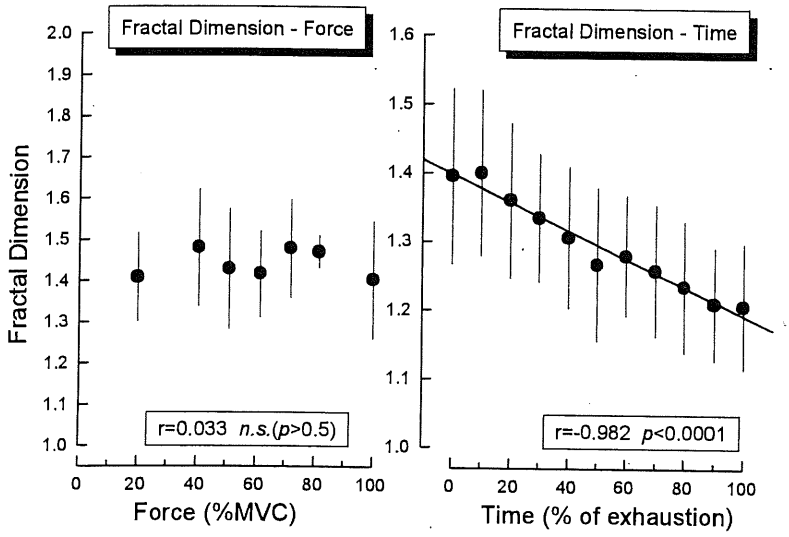
## METHOD

Six healthy males performed isometric elbow flexion with the elbow joint angle of 110deg. Separate bouts of isometric contractions at seven activation levels were conducted for 3sec with resting intervals in between. Subsequently, repeated contractions at 60% MVC for 3sec with 2sec relaxation period was continued until the subject could no longer produce 60% MVC. In both tests, surface EMG was recorded from biceps brachii muscle, digitized and sampled at 10,000Hz. FD was calculated according to Higuchi's method developed for fractal analysis of time series data (Higuchi, 1988).

## RESULTS

*Fig. left* : FD during separate bouts was 1.4～1.5 and was not correlated to muscle activation.

*Fig. right* : FD at the onset of repeated contractions was 1.395±0.128, and it decreased to 1.207±0.092 at the endurance limit. FD was linearly related to the exercise time relatively expressed to exhaustion time.




## DISCUSSION

It is found that FD does not correlate to muscle activation if the analysis method is not affected by signal amplitude. Linear decrease in FD during repeated contractions would be partly related to the changes in power frequency domain of EMG signal. In conclusion, fractal dimension, a measure of "complexity", of surface EMG during isometric contraction is not related to muscle activation, but to muscle fatigue when analyzed by the method independent of signal amplitude.

## 【REFERENCES】

Anmuth CJ, et al. (1994) Fractal dimension of electromyographic signals recorded with surface electrodes during isometric contractions is linearly correlated with muscle activation. Muscle & Nerve 17: 953-954.

Higuchi T. (1988) Approach to an irregular time series on the basis of the fractal theory. Physica D31: 277-283.

Katz MJ (1988) Fractals and the analysis of waveforms. Comput Biol Med 18:145-156.

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# ELECTROMECHANICAL DELAY OF CONCENTRIC, ECCENTRIC AND ISOMETRIC MUSCLE ACTIONS IN MEN AND WOMEN

K. Takase and M. Taguchi
Department of Physical Education, Fukuoka University. Fukuoka, Japan

## INTRODUCTION

Electromechanical Delay (EMD) was defined as the time lag from the onset of the muscle electromyographic activity (EMG) to the onset of development of tension in the muscle. This EMD was thought to be one factor that had a great influence on the appearance of movement action, sports performance and sports disorders. The purpose of this study was to measure electromechanical delay (EMD) of concentric, eccentric and isometric muscle actions in men and women of knee extensors, and to investigate each of the muscle action modes of EMD in men and women.

## METHODS

The subjects were 21 men and 22 women undergraduate physical education students. The measuerment of EMD were carried out using KIN-COM for knee extensors. There were 3 kinds muscle action mode(concentric, eccentric and isometric muscle actions). Each muscle action was enforced with the movement velocity of 0 ° /sec (isometric muscle action), 30° /sec, 120° /sec and 210° /sec (concentric, eccentric muscle actions). The kinds of stimulus that carried out were the proprio receptor response in tibial of sense organ (concentric, eccentric muscle actions) and voluntary movement (isometric muscle action). Also, The force levels were enforced with the 100% maximum voluntary contraction, and the subjects were asked to react to the force levels as quickly and as strongly as possible. The calculation of EMD was obtained by the difference between PMT(the time from onset of stimulus to the onset of muscle action) and TRT(the time from the onset of stimulus to the onset of development of muscle torque).

## RESULTS

In both men and women, EMD lengthened as movement velocity in concentric muscle action increased. On the contrary, there was a tendency for eccentric muscel action to become shorter. Also, the EMD of eccentric muscle action showed shorter significant value than concentric and isometric muscle actions. Moreover, the EMD in men showed shorter significant value ($P<0.01$) than women in concentric, eccentric and isometric muscle actions of all movement velocity(Table 1).

## DISCUSSION

According to these results, the EMD in concentric muscle action was lengthened. It was considered that this change was due to a great infuluence by contractile component, series elastic component and parallel elastic component in the case of slow movement velocity, and the dependence of contractile component was getting higher as movement velocity increased. On the contrary, the EMD in eccentric muscle action was getting shorter. It was considered that, series elastic component was stretched fast as movement velocity increased, and activated the elastic structure and generated efficiency force. Moreover, as for the difference between men and women in the EMD, it was considered that complex the internal cause such as muscle and tendon might be concerned .

Table 1. Electromechanical Dalay (means±SD) of Men(n=21) and Women(n=22)

| EMD (msec) | 男子EMD | | 女子EMD |
|---|---|---|---|
| ISO 0° | 73.4±9.6 | ** | 86.6±9.1 |
| CONCENTRC 30° /sec | 71.0±11.7 | ** | 87.3±12.5 |
| CONCENTRC 120° /sec | 78.0±9.7 | ** | 90.8±11.0 |
| CONCENTRC 210° /sec | 81.9±10.7 | ** | 100.5±10.9 |
| ECCENTRIC 30° /sec | 67.6±10.4 | ** | 79.6±11.7 |
| ECCENTRIC 120° /sec | 61.5±10.9 | ** | 75.8±8.9 |
| ECCENTRIC 210° /sec | 56.2±8.4 | ** | 70.0±10.1 |

(** p<0.01)

# EFFECTS OF STRETCH REFLEX MODULATION ON THE MOVEMENT PERFORMANCE DURING ADJUSTMENT MOVEMENTS

T. Kizuka *, T. Aasami ** K. Tanii*** and M. Okada **

* National Institute of Bioscience and Human-Technology, Ibaraki, Japan

** Institute of Health and Sport Sciences, University of Tsukuba, Ibaraki, Japan

*** Department of Industrial Engineering, Musashi Institute of Technology, Tokyo, Japan

[ INTRODUCTION ]

The gain of stretch reflexes, particularly long latency components, was reported to be modified by motor preparation and some studies mentioned that the change in long latency activity was more remarkable for the subjects with short reaction time. These findings suggest that the formation of motor preparation is appropriate in the subject who can react to a stimulus quickly and, as a result, it may be useful for the initiation of quick movement to change long latency activity of the muscle involved in the movements by that preparation. Therefore, in order to perform adjustment movements quickly and accurately, which are mainly based on open loop control, stretch reflex activities of the muscle involved in the movements must be modified too. Moreover, the degree of modulation in stretch reflexes with appropriate preparation is thought to be linked with good performances of adjustment movements. In the study, the effect of the long latency reflex modulation on the performance of a quick adjustment movement following a muscle stretch was investigated in 26 healthy male subjects.

[ METHODS ]

Subjects were instructed to perform an adjustment so that the position output of the handlebar aligned with a target position as quickly and as accurately as possible when they felt the rapid angle displacement in the direction of a wrist extension, which was produced by the DC torque motor. The target position of this study was set up to be on the extension side of the wrist joint. The indexes of performance were the time taken to move the handle to the target position from stretch onset (adjustment time) and it's reproducibility (adjustment time-S.D.). The short and long latency activities (M1 and M2 components) were measured by averaging the rectified EMGs which recorded with the surface electrodes over the wrist flexor. The response onset time was represented by measuring the time from stimulus onset to voluntary EMG burst on the rectified EMGs.

[ RESULTS ]

For all subjects, the reflex activities in the flexor decreased during the adjustment movement, because the target position was the extension side of wrist joint. The decrease in the M2 component, which is considered to be a transcortical reflex, was significantly larger than the decrease in the M1 component, which is a spinal reflex. Furthermore, the level of M2 decrease was large in the group with short adjustment time, while that was small in the group with long adjustment time. The difference of M2 decrease level was statistically significant. On the other hand, the scatter of response onset time was smaller in the group with small adjustment time-S.D. than in the group with large adjustment time-S.D.

[ DISCUSSION ]

The results from this study show that the level of M2 modulation is closely connected with the adjustment ability of the neuromuscular system, which demands quick and accurate movements simultaneously. Moreover, we claim that the subjects, who construct the appropriate motor set to complete the adjustment movements quickly and accurately, are able to greatly inhibit stretch reflex activities in the flexor, particularly the long latency component that produced the tension obstructing the target movements at the extension side. The strong inhibition of long latency component helps to make the function of voluntary muscle activity more effective and so they are able to perform the adjustment movements within a short time. Added to these, in order to complete adjustment movements with high reproducibility, it is important that the onset of voluntary muscle activity is consistent .

# EFFECT OF INITIAL JOINT ANGLES ON STRETCH REFLEX SENSITIVITY IN ELBOW FLEXOR MUSCLES

**K. Nakazawa*, S. Yamamoto*, T. Fukunaga**, M. Itoh**, and H. Yano***
*Research Institute, National Rehabilitation Center for the Disabled, Tokorozawa, Japan
**The University of Tokyo, Tokyo, Japan

## INTRODUCTION

To understand the neural mechanism underlying modulation of the stretch reflex sensitivity during voluntary movements is one of key issues in motor control study. However, it is difficult in vivo to measure the stretch reflex sensitivity of muscles, which is generally defined as the relationship between the muscular stretching velocity and the evoked muscle response. In the present study, we propose a new method to estimate the stretch reflex sensitivity, i.e., using a biomechanical model and the ultrasound technique to estimate the muscle stretching velocity induced by perturbations, and examine whether the stretch reflex sensitivity vary with limb positions.

## METHODS

Subjects, the experimental setup, the motor task employed and measurements are same as those described in the accompanying paper (Yamamoto et al. in this symposium). In the present study, three different angular velocity perturbations, 50 deg/s, 80 deg/s and 180 deg/s were used to evoke the stretch reflexes at each joint angle. Ten perturbations were applied for each angular velocity at each joint angle. Averaged EMG responses to those perturbations were used to estimate the stretch reflex sensitivity.

The muscle stretching velocities induced by the three angular velocities were estimated by the biomechanical model of elbow joint and muscles (van Zuylen et al. 1988a). Moreover, for four subjects, the length changes in the BBL accompanying with the elbow joint extension were measured by the ultrasound technique. Sagittal images of the BBL were obtained at each 10 deg from 110 deg to 0 deg (full extension) of the elbow joint. From those images moving distances of a muscle-tendon junction with joint angles were measured, and the muscle stretching velocities were calculated with the actual data for the elbow joint displacement in the stretch reflex test.

## RESULTS AND DISCUSSION

The length changes in BBL with joint angles estimated by the biomechanical model and the ultrasound technique were both larger as the elbow joint more flexed. Therefore, the estimated muscle stretching velocity induced by the same perturbation at the flexed joint positions were higher than those at the extended positions. Both the short (M1) and the long (M2, M3) latency reflex components were directly proportional to the muscle stretching velocity (Fig. 1). It was revealed with this analysis that the slope of the linear regression line increased as the elbow joint extended, indicating the higher stretch reflex sensitivity at the extended joint position, i.e., the sensitivity was highest at 35 deg flexed joint angle. This is in accordance with the result for the motor unit recruitment threshold (MURT) demonstrated by van Zuylen et al. (1988b), in which MURT of the biceps brachii was decreased as the elbow joint extended. Further, we observed that the EMG reaction time at the extended joint position was higher than that at the flexed position (Nakazawa and Yano 1991). Taken together these data suggest that peripheral afferent inflow accompanying with joint extension has some facilitatory effect on the spinal and/or supraspinal nervous center, and so that the stretch reflex sensitivity increased as the elbow joint extended.

## REFERENCES

Nakazawa, K. and H. Yano, Nat. Rehab. Res. Bull. Jpn., 12, 45-50, 1991.
van Zuylen, E. J., et al. J. Biomech., 21, 183-190, 1988a.
van Zuylen, E. J. et al., J Neurophysiol, 60, 1523-1548, 1988b.

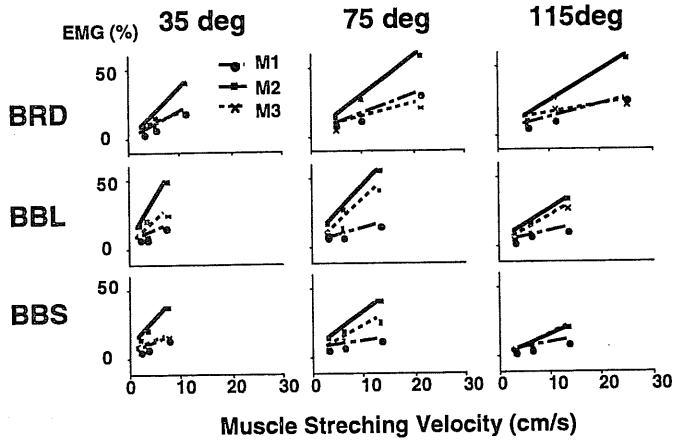



Fig. 1 Relationships between the muscle stretching velocity and the stretch reflex EMG responses at the three joint angles.

# MODULATION OF STRETCH REFLEX RESPONSES WITH INITIAL JOINT ANGLES IN ELBOW FLEXOR MUSCLES

**S. Yamamoto*, K. Nakazawa*, T. Ohtsuki** and H. Yano***
* Research Institute, National Rehabilitation Center for the Disabled, Tokorozawa, Japan
** The University of Tokyo, Tokyo, Japan

We focused on the relationship between the stretch reflex and the limb position in the synergistic elbow flexor muscles. The purpose of this study was to investigate the effect of the joint angle changes on the short- and long-latency stretch reflex responses in elbow flexor muscles.

## METHODS

Ten healthy subjects aged 20 - 33 years participated in this experiments. The EMG activities of the brachioradialis (BRD), the long and short heads of biceps brachii (BBL, BBS) were recorded. The subject was instructed to exert the elbow flexion torque corresponding to 6%MVC EMG level of BRD. The isometric elbow flexion was performed horizontally at three elbow joint positions; 35 deg, 75 deg and 115 deg (full extension angle was 0 deg). The perturbations were given to extension direction by the torque motor. The all subjects were instructed "do not resist to the perturbation". The session consisted of 10 perturbations at each elbow position. The background activity level (BGA) occurring in the 100 ms before the onset of the perturbation was calculated, and the reflex EMG activities were subdivided into the short-latency reflex response (M1) and the long-latency reflex responses (M2 and M3). The mean EMG activity was calculated in these reflex and volitional EMG responses. The BGA was subtracted from all stretch-induced EMG responses.

## RESULTS

Figure 1 shows the ensemble average for the EMG responses and joint displacements in all conditions. The variation patterns of the reflex EMG responses with the joint angles were differed between BRD and the both heads of the biceps brachii. The means and SEs of each reflex EMG component were shown in Figure 2. In spite of the same BGA level across the elbow positions, the M1 and M2 responses of BRD were significantly increased at the 75 deg and 115 deg as compared to those at 35 deg, and the M2 responses of BBL and BBS were significantly increased at 35 deg and 75 deg as compared to those at 115 deg.

## DISCUSSION

The results of this study provide the clear evidence that the short- and long-latency stretch reflex responses were dependent on the initial elbow joint positions, and further, the patterns of position dependency were varied in different muscles.

There are two factors that may explain these results. Firstly, the muscle length changes induced by the perturbation might be different among the three joint positions. The moment arm of the elbow flexors varies with the joint angle. Therefore, although the same perturbation was applied to the forearm at the three joint positions, the changes in the muscle fiber length might not be equal. This means that the input to the muscle spindle might be different at the three elbow joint angles. Secondly, the changes of the stretch reflex sensitivity with the joint angle sholud be considered. van Zuylen et al. (1988) indicated that the recruitment threshold of the motor units was dependent on the joint positions and the coordination patterns of synergistic muscles were different at the different joint angles. They suggested that the muscle with the larger moment arm receives the larger input. To make clear about these two factors, the muscle stretching velocities induced by the perturbation should be measured in each muscle. The accompanying paper (Nakazawa et al.) examines this.

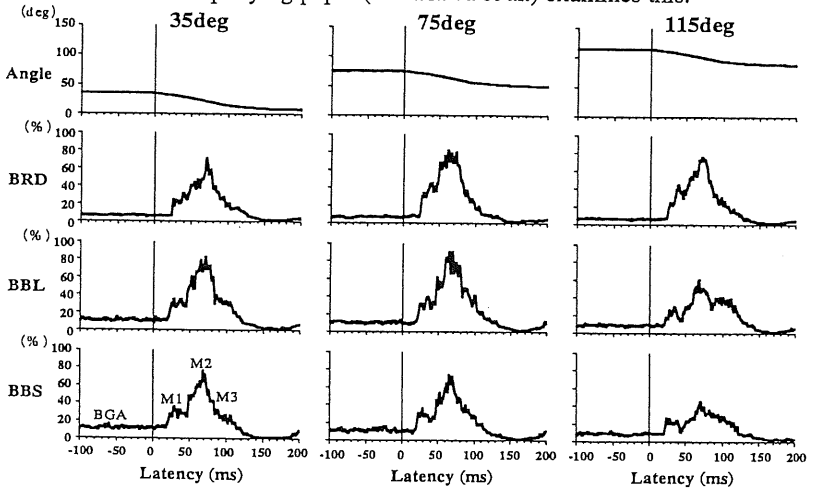



Fig. 1 Ensemble average wave forms

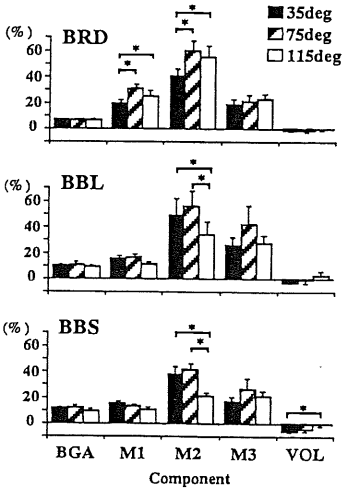



Fig. 2 Mean EMG responses, *: $p<0.05$

# REFLEXIVE CONTROL OF THE DELFT SHOULDER MODEL

**Leonard A. Rozendaal**

Dutch Shoulder Group, Faculty of Human Movement Science,
Vrije Universiteit, Amsterdam, The Netherlands.

**INTRODUCTION** In the majority of large-scale musculoskeletal models, stability of the model is not incorporated in the optimization goals. Hence there is no guarantee that the obtained muscle force patterns represent stable situations, in the sense that the model will return to its original position (or desired trajectory) after the application of a transient perturbing force. In this presentation, the Delft Shoulder Model is extended to a dynamic control model. Using an eigenvalue decomposition, it can be shown that the original model generally is unstable when the muscle forces are determined on the basis of the assumptions that 1) there should be force and moment equilibrium; 2) the use of the muscles should be energy-efficient. A reflexive control module is added to the model, which can resolve the instability for an appropriate choice of the sensory feedback gains.

**THE DELFT SHOULDER MODEL** The Delft Shoulder Model[1] is a large-scale three-dimensional mechanical model of the human shoulder. The geometrical characteristics, i.e. bone shapes and attachments of ligaments and muscles relative to the bones have been determined in an extensive cadaver study. The mechanical effect of the muscles is described by dividing each muscle into an appropriate number of elements. The dividing method that was used[2] results in a total of 95 muscle elements for the muscles of the shoulder region.

**EXTENSION TO A DYNAMIC CONTROL MODEL** From the existing version of the Delft Shoulder Model a linearized dynamic relation for the inertial dynamics of the shoulder and arm can be obtained[3], which is transformed to a state-space model (6 degrees of freedom, 12 states). For each of the 95 muscle elements, a linearized three-state muscle model has been incorporated after the model of Winters and Stark[4]. From this muscle model also the contractile element length, velocity and force are obtained, which are fed back proportionally to the muscle elements. In addition, the time delay for these sensory signals of each muscle element has been included using a two-state Padé approximation. This results in a total of $12+95\cdot(3+2) = 487$ states for the neuromusculoskeletal model. By determining the eigenvalue decomposition of the state-space system, its stability can be assessed and the stiffness and damping can be quantified. Simulations of seven stages of an abduction movement show that without reflexive control the model is in an unstable equilibrium at and above 90 degrees of humeral elevation, which is caused by the destabilizing effect of gravity and of forces in muscles with a destabilizing geometry.

**DETERMINATION OF THE OPTIMAL FEEDBACK GAINS** For determining the feedback gains that result in the optimum behaviour of the system (stability plus a maximum stiffness at an acceptable damping level) numerical optimization techniques are required. However, each evaluation of the criterion function requires an eigenvalue decomposition of the system, which is computationally very expensive. As the eigenvalue decomposition time is approximately proportional to the third power of the number of system states, reducing this number is essential. From earlier simulations with the model it was found that the force course over the elements of a muscle usually can be parametrized by a considerably smaller number of parameters than the original number of elements. Using this fact, the number of representative muscle elements can be reduced from 95 to 35 with hardly any loss of accuracy. A reduction of the number of model states from 487 to $12+35\cdot(3+2) = 187$ ensues, and as a result the eigenvalue decomposition time reduces by a factor 18. To each of the representative elements of the model an individual length feedback gain is attributed, while the parameters of the velocity and force feedback gains are the same for each element. This results in a total number of 37 feedback gains, the numerical optimization of which is just feasible. This optimization results in stability for all 7 investigated postures, together with a realistic (though somewhat low) endpoint stiffness. The pattern of the feedback gains as a function of the elevation angle for the muscles from thorax to clavicle and scapula differs from that of the muscles which insert on the humerus. The former group shows optimum feedback gains that are relatively independent of the joint angle, whereas for the latter group the optimum feedback gains depend strongly on the elevation angle.

**DISCUSSION** In the present approach a reflexive control model for a large-scale musculoskeletal model is presented. This control is very effective, as it combines a minimum-energy muscle activation pattern which in itself results in unstable system behaviour, with a reflexive control which stabilizes the system and achieves good dynamic behaviour. The reflexive control system does not require additional energy as long as no disturbing forces are actually present, in contrast to muscle co-contraction.

**REFERENCES** **(1)** Van der Helm FCT (1994) A finite element musculoskeletal model of the shoulder mechanism. J. Biomech. 27, 551-569. **(2)** Van der Helm FCT, Veenbaas R (1991) Modelling the mechanical effect of muscles with large attachment sites: application to the shoulder mechanism. J. Biomech. 24, 1151-1163. **(3)** Rozendaal LA (1997) Stability of the shoulder, intrinsic muscle properties and reflexive control. PhD thesis, Delft University of Technology, The Netherlands. **(4)** Winters JM, Stark L (1985) Analysis of fundamental human movement patterns through the use of in-depth antagonistic muscle models. IEEE Trans. Biomed. Eng. BME-32, 826-839.

# SOLEUS H-REFLEX DURING VOLUNTARY LUMBAR MUSCLE CONTRACTION

S. Tanaka * and A. Nagata **
* Graduate School of Human Sciences, Waseda University, Saitama, Japan
** School of Human Sciences, Waseda University, Saitama, Japan

## INTRODUCTION

It is well known that H-reflex changes due to by various factors such as contraction of the target and antagonist muscles, and subject's posture and breathing. Increases in the H-reflex may reflect increased motoneuron pool excitability on the spinal cord and recruited motor unit size and firing rate in the central nervous system. The purpose of this investigation was to observe motoneuron pool excitability noninvasively, using the H-reflex measured at the soleus during voluntary lumbar muscle contraction.

## METHODS

Experiments were carried out on healthy male volunteers, lying in the prone position. H-reflex was recorded from the soleus muscle of the right leg, and was produced by post-tibial nerve stimulation at the popliteal fossa. Bipolar surface electrodes were used to elicit the soleus H-reflex and data were recorded using a DAT machine. We used the maximum entropy method (MEM) for spectral analysis to measure H-reflex alterations by calculating peak power frequency (PPF).

## RESULTS

When the subject attempted to perform contralateral contractions of the lumbar muscle the soleus H-reflex amplitude decreased from 0.05mV to 0.03mV. In contrast, when the subject performed ipsilateral contraction H-reflex increased to 1.89mV($P<0.05$). Furthermore, while subjects sustained contraction of the lumbar muscle for a period of 20sec, and then H-reflex amplitude increased significantly ($P<0.001$). After continuous muscle contraction, H-reflex was altered during recovery from 2.0mV to 0.8mV. Then, PPF of H-reflex spectrum were calculated at rest, maintenance of contraction, and during recovery by MEM frequency analysis. PPF values were 68.3 Hz, 108.8 Hz, 119.4 Hz at these three stages, respectively. H-reflex was shifted to a higher frequency with muscle contraction.

## DISCUSSION

Excitability of the motoneuron pool was estimated by determining the soleus H-reflexes elicited during rest, lumbar muscle contraction and recovery. The amplitude of the soleus H-reflex increased significantly during ipsilateral lumbar muscle contraction as compared to those at rest and during contralateral contraction. On the other hand, when the subjects performed contralateral lumbar contraction, the evoked electromyogram signals were depressed. The observation that the H-reflex amplitude was increased probably indicated that enhancement of excitability on motoneuron pool caused facilitation drive signals from the central nervous system and the afferent pathway. Decreases in H-reflex amplitude suggested that the effect of reciprocal inhibition should appear at the presynapses in the spinal cord. We performed spectral analysis using the maximum entropy method, and H-reflex was observed during sustained lumbar contraction. PPF values obtained from the H-reflex spectrum were estimated with excitability at recovery period after voluntary contraction. These PPF values were higher than these at rest or during lumbar muscle contraction. These results indicated that alteration of H-reflex amplitude and H-reflex spectrum reflect upon functional changes in motoneuron pool excitability caused by facilitation from the central nervous system.

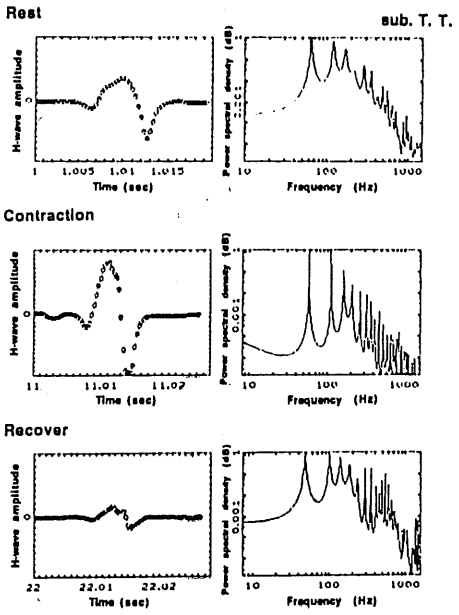



Fig. Raw data and power spectra of the H-wave during each procedure (sub. T. T.)

# Non-linear property of the triceps surae H-reflexes during fast plantarflexion in humans

T. Komiyama* and M. Humoto**
*Department of Health and Sports Sciences, Chiba University, Japan and ** Doctoral program in Health and Sport Sciences, Tsukuba University, Japan

## INTRODUCTION

It has long been demonstrated that the size of Hoffmann (H-) reflex increases before and during voluntary contraction of the homonymous muscle (Schieppati, 1987). However, it is still controversial whether the degree of the H-reflex facilitation is closely correlated to the level of the output force or EMG (Gottlieb and Agarwal, 1970; Rüegg et al., 1990). In the present study, we re-examined in more detail the relationship between the size of the H-reflex and the amount of background (BG) electromyographic (EMG) activity of the homonymous muscle at different levels of contraction force.

## METHODS

The experiments were carried out on ten healthy male subjects aged 20 to 37 years, all of whom gave informed consent to the purpose and the procedures. The subjects sat in a reclining armchair with the knee and ankle joints at 170 and 110 degrees, respectively. The feet were fixed to a pair of immobile metal foot plates to make the movement isometric as much as possible. Subjects were asked to perform fast plantarflexion under a reaction time condition. The amounts of contraction force were 10, 20, 50 and 100% of maximum. EMG signals were recorded from the gastrocnemius medialis (GM), the gastrocnemius lateralis (GL) and the soleus (Sol) muscles with surface electrodes. The posterior tibial nerve was stimulated transcutaneously to elicit the H-reflex from the GM, GL and Sol. A rectangular pulse (1 ms duration) was used for the electrical stimulation. The intensity of the test stimulus was carefully adjusted to result in an M-response that was 20 % of its maximum size in all cases. The constancy of the test stimulus was confirmed by checking the shape and peak to peak amplitude of the M-response. The time intervals between the onset of the EMG and the H-reflex stimulation were varied from 10 to 120 ms. The amount of the BG EMG was calculated from the full-wave rectified and averaged (n=10) EMG signal.

## RESULTS

The maximum size of the M-response (Mmax) altered systematically during the contraction. In general, it decreased up to 50 ms after the EMG onset, and then increased gradually. We maintained the relative size of the test M-response constant (about 20 % of the Mmax) throughout the experiment so that the actual size of the test M-response was slightly different among the stimulation intervals. Although the Sol H-reflex strongly facilitated during plantarflexion, the degree of the Sol H-reflex facilitation was not related linearly to the amount of the BG EMG. Irrespective of the same level of the BG EMG, the size of the H-reflex is always larger in the early rising phase of the EMG activity than the later falling phase. Accordingly, the relationship between the size of H-reflex and the level of the BG EMG has a non-linear feature. Almost the same results were obtained from the Sol, GM and GL. The H-reflex was strongly facilitated even when the exerted force was only 10 % of its maximum. We therefore found no significant difference among the maximum size of the H-reflex obtained in the different levels of plantarflexion forces. This result was obtained in all three muscles.

## DISCUSSION

It was reported that the H-reflex is a linearly increasing function of the BG EMG for a fixed stimulus strength (Gottlieb and Agarwal, 1971; Capaday and Stein, 1986). The term "automatic gain compensation" has been proposed to show that the gain of the reflex increases with the excitation level of the motoneuron pool (Matthews, 1986). However, it was shown in this study that the relationship between the size of H-reflex and the level of BG EMG was not a linear function during fast movement. Reasons for this discrepancy could be attributable to methodological differences; we used fast movement as a motor task, but locomotor activity and steady contraction were used in other studies. So, it is likely that a different supraspinal control acts on the H-reflex arc according to the type of voluntary movement performed. The non-linear relationship between the size of H-reflex and the BG EMG suggests that the level of the presynaptic inhibition onto Ia terminals changes very quickly during the course of the movement. Further, the constancy of the maximum size of the H-reflex during different output forces suggests that the level of the presynaptic inhibition decreases and increases while exerting lower and higher levels of the output force, respectively.

# FORCE TRAJECTORY AND EMG ACTIVITY IN BALLISTIC KNEE EXTENSION

S. Mizumura, T. Yoneda* and T. Kimura**
Laboratory of Health and Physical Education, School of Arts and Letters, Meiji University, Tokyo, *Laboratory of Physiology, School of Sports and Health Sciences, Juntendo University, Chiba, **Laboratory of Physical Education, Tokyo Institute of Polytechnics, Kanagawa, Japan.

## INTRODUCTION

In ballistic contractions performed as fast as possible, it has been postulated that the entire force trajectory and EMG activity depend on a preplanned central motor program, and that no corrections are possible. An investigation of the characteristics of force trajectory and EMG activity during ballistic contraction should help us, then, to understand the properties of the central control mechanisms relevant to human power output. Though many researchers have carried out various experiments focusing on upper limbs during ballistic contraction, we have adopted thigh muscles as the objects of our analysis, because human explosive movements such as jumping and kicking are mainly derived from the knee extension force produced on knee extensor muscles.

The purpose of our study is to investigate properties of the motor commands responsible for force production on knee extension from low to high levels of force exertion.

## METHODS

Twelve normal male volunteers between 19 and 35 years of age participated in our experiments. The subjects sat in an experimental chair with their right knee fixed firmly at an angle of 90 degrees. They faced a storage oscilloscope on which exerted forces would be displayed. We instructed them to exert extension force with their right knee "as fast as possible". Initiations of force exertions were self-paced. The tasks were performed for 60-100 trials, and subjects were instructed to exert at varying levels, in a range from 10 to 80% of their maximum voluntary force (MVF).

Force signals were detected by strain gauges attached to a stainless steel rod located at a point just above the right ankle. Surface EMG signals were led from Vastus Medialis(VM), Vastus lateralis (VL) and Rectus Femoris (RF) that had each been attached with a pair of 8mm diameter silver-silver chloride electrodes. Both force and EMG signals were stored on floppy disks after A/D conversion at 1kHz sampling rate. We measured peak force, force rise time, rate of force rise, EMG duration, amount of muscle discharge (iEMG), discharge rate, and rectified averaged EMG.

## RESULTS AND DISCUSSION

Force curves showed a bell-shaped one-peak wave form up to 50%MVF. At over 60%MVF, detected force curves then deviated from this wave form into a two-peak wave form. It is noteworthy that here initial force trajectories, from force initiation to the first peak achieved, followed the same time course over all trials. At under 50%MVF, force rise time and EMG duration showed near-constant values independent of peak force, with little deviation, while at over 60%MVF those values increased with increasing peak force, with greater deviation. Rate of force rise increased linearly with peak forces at under 50%MVF, but above such force levels it then leveled off into a scattered form. Rectified averaged EMG showed a single volley at under 50%MVF, while it showed two successive volleys at over 60%MVF. First volleys were almost all nearly the same in amplitude and duration, while those of second volleys increased with increasing second peak force. iEMG increased with increasing peak force at under 50%MVF, and at over 60%MVF, it showed greater increase in response to the same increases in peak force. Discharge rate increased with increasing peak force at under 50%MVF, while at over 60%MVF it levelled off into a scattered form.

Our results indicate that there is a critical force level at which the output mode of the central motor commands change. We believe that the neural controls of ballistic knee extension would include two components. The first, a "limited time component", only appears below the critical force level, while at levels above this, the second, a "varied time component", appears immediately following the first. The first component modulates the magnitudes of motor commands within a near-constant time, while the second modulates them with increasing execution time. It seems likely that the central nervous system might regulate actual explosive leg movements as well.

# MUSCLE ACTIVATION PATTERNS AND PERFORMANCES DURING SUCCESSIVE JUMPS ON A VARIOUSLY INCLINED PLATFORM

H. Nishizono, A.Matsuo, H. Kintaka, M.Kon, Y.Ohmura, H. Shibayama and A. Maeda*
National Institute of Fitness and Sports. Kanoya, Japan.
* Department of Hygiene, School of Medicine, Akita University. Japan.

## INTRODUCTION

In most jumping exercises and training, high performance is achieved by use of effective stretch-shortening-cycle (SSC) contraction, which involves for the extensor muscle the effective transfer from eccentric to concentric contraction. Under SSC-condition, EMG activation can be divided into three different functional phases(Gollhofer,1987): 1. the preactivation phase(before ground contact); 2. the reflex-induced activation phase (30-120 ms after touch down); 3. the late EMG-response(over 120 ms: mainly innervated during the latter phase of the push-off). These activation patterns can be modulated by the degree to which the platform is inclined. Some modulations which impose stretch load on the ankle joint will be related to the development of jumping performance capability. The present study was designed to investigate the changes in neuronal activation patterns, force-time characteristics and jumping performances on a variously inclined platform.

## METHODS

The subjects were 18 healthy males aged 19 to 22 years. The subjects performed 5 jumps successively on 0, 10 and 20 degree inclined platform. The first two jumps were preparatory, and the third, forth and fifth jumps were trial jumps and were performed with maximum effort. The electrical signals of the three maximal trials of (1) surface EMG of M.Soleus(SOL), M.Gastrocnemius(GA), M.Vastus Lat.(VL), M.Rectus Fem.(RF), M. Biceps Fem. and M.Tibialis Ant. and (2) vertical ground reaction force were recorded and processed. Jumping height and contact time on the platform were also calculated for each trial.

## RESULTS AND DISCUSSION

On the 10 degree inclined platform, the mean jumping height was higher than on either the 0 or 20 degree platform. The mean contact time and SD on the platform were $158 \pm 14$ ms on 0 degree, $162 \pm 14$ ms on 10 degree and $175 \pm 13$ ms on 20 degree. The mean muscle power showed the highest value on 10 degree platform. Figure depicted a representative example the average rectified EMG-patterns of SOL, GA and VL and the vertical forces in one subject during the landing condition with 0 and 10 degree. The amplitude of SOL muscles during the 10 degree jumps increased around 100 ms after ground contact, compared with 0 degree. However, the relative contributions of the VL and RF muscles revealed almost similar activations during all three degrees of elevation. It is assumed that with increased stretching load (10 degree) the triceps surae muscle will be the most sensitive and be related to enhancement of jumping height.

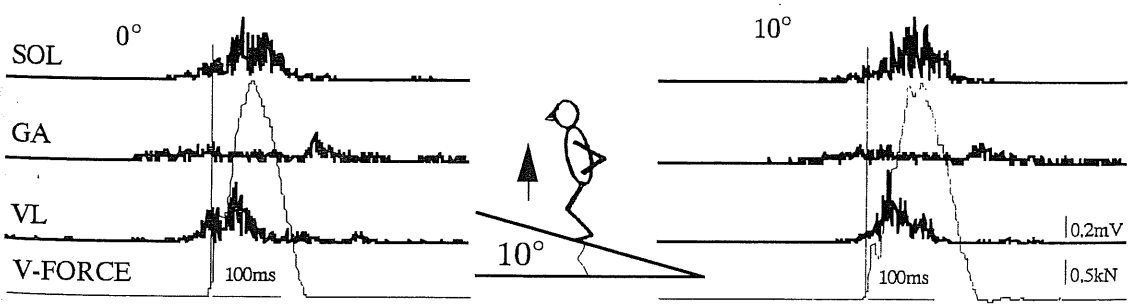



Fig. EMG and vertical force ; each datum represents the average of 3 trials.

# EMG responses in the rotator cuff muscles elicited by electrical stimulation of human shoulder joint afferents

Voigt, M. Jakobsen,J.* and Sinkjaer T.
Center for Sensory-Motor Interaction, Aalborg University, *Dept. Orthopedic Surgery, Aalborg Hospital, Denmark. Ermail: mv@miba.auc.dk

## INTRODUCTION

The role of the joint afferents in the control of the human glenohumeral joint has not been clarified in detail yet. In this study we investigated the effect of electrically induced afferent activity on passive and active and rotator cuff muscles in humans.

## MATERIAL AND METHODS

Three female and four male patients (age 35-63 years) participated in the study. The patients had no history of trauma on the glenohumeral joint capsule. During general anesthesia arthroscopy of the glenohumeral joint was performed due to either subacromial pathology (two subjects) or for diagnostic purposes. Four or six fine wire electrodes were inserted in the shoulder joint capsule in patches with an inter-electrode distance of 5 mm. In three subjects the electrodes were either placed in the posterior inferior part, the posterior superior part, or the anterior superior part (including the biceps tendon) and in four subjects in electrodes were placed in the anterior inferior (ant. inf.) part of the joint capsule (including the glenohumeral ligament). Pairs of fine wire intramuscular EMG-electrodes were inserted in m. supraspinatus, m. infraspinatus and m. subscapularis. The electrode wires were covered with bandages and the arm was fixated. The measurements were performed the next day when the subjects were fully recovered from the anesthesia. Trains of square constant current stimuli (8 times 1ms at 200Hz or 4 times 0.5ms at 400Hz) were delivered through pairs of wire-electrodes in all electrode combinations. The perception of the stimulus intensity was rated with a linear scale between '1' (the sensory threshold) to '10' (beginning of pain perception). The stimulus intensity was chosen corresponding to a perceived rating about 5. The interstimulus interval was 3s and recordings were performed in series of 10 stimuli and repeated up to 6 times in two conditions: a) relaxed muscles and b) during a moderate static background contraction. The EMGs were sampled at 4KHz and analyzed off line.

## RESULTS

No EMG responses were observed when the stimuli were applied in the posterior parts and the ant. superior part of the joint capsule. When the ant. inf. part was stimulated and the muscles were relaxed excitatory responses were observed in m. infraspinatus and in m. subscapularis with a latency to the onset of 48±7ms (mean±SD) (fig.1, curves presented upside-down). When the muscles where contracted a clear inhibition of the muscle activity followed by an excitation was observed in all the muscles (fig.1). The latency of the onset of the inhibition was on average 57±26ms. The depression lasted 43±7ms.

## DISCUSSION

These preliminary results indicate that there are a strong innervation of the ant. inf. part of the gleno-humeral joint capsule including the glenohumeral ligament. This

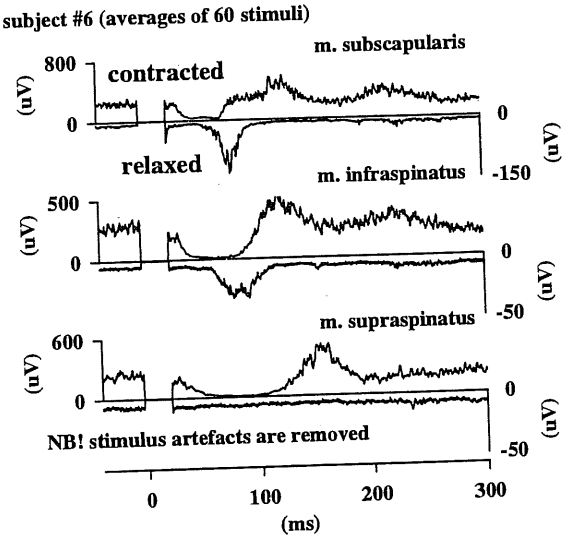



Fig.1.

is consistent with the observation of well defined nerve branches from the axillary and musculocutaneous nerves to this part of the capsule (Wrete 1949). The rather long latencies observed does not exclude that voluntary interaction is involved. However, the patients were not able voluntary to suppress the responses elicited by stimuli applied at times unknown to them. Human joint afferents are small diameter nerves (3-4μm) and about 70% are unmyelinated (Hines et al 1996) and correspondingly the conduction velocities are low (max. 12-16 $ms^{-1}$). Jerosh et al. 1995 observed EMG responses shoulder muscles with latencies of 100ms and more in fully anaesthetized patients elicited by intraarticular stimulation of the anterior inferior part of the joint capsule. Our results indicate that the afferent activity from the joint capsule may induce both excitatory and inhibitory responses in the rotator cuff muscles depending on if the muscles are voluntary activated or not. This is consistent with the observation that joint afferents from the posterior articular nerve from the cat knee joint acts through both excitatory and inhibitory di- and trisynaptic reflex pathways from Ib-afferents (Lundberg et al.1978). Therefore, the shoulder joint afferents may play an important role in the active control of joint stability through regulation of the muscle force in the rotator cuff muscles.

## REFERENCES

Hines.AE. et al. (1996) Anat. Rec. 246: 573-578
Jerosch,J et al. (1995) Unfallchirurg. 98(11),480- 485.
Lundberg,A et al. (1978) J. Physiol. 284, 327-344.
Wrete,M (1949) Acta Anat. 7, 173-190.

## ACKNOWLEDGEMENTS

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# EFFECT ON MOTOR NEURON ACTIVATION OF HTE TRICEPS SUARE IN CO-ACTIVATION OF THE STERNOCLEIODOMASTOID MUSCLES DURING CLENCHING OCCLUSION.

M. Muro, M. Yona and C. Tadano, H. Seki, K. Oka, K. Kawada, and T. Yamagata

Toho University School of Medicine, Tokyo, Japan.

## INTRODUCTION

It is well known that motor skill is dependent upon the stability of the unsupported head by the function of the cervical muscles, and that the functional activations are supported by the masticatory muscles. In fact, neuro-anatomical studies of animals and humans have established that the trigeminal spinal tract and nucleus extend to the upper cervical segments(Rustioni et al., 1971; Abrahams et al., 1979). It is reasonable to conjecture that there would be a coupling between the limb and the masticatory or cervical muscles, because of the dependence of the motor skill. Further essential analyses and evaluations of the degrees of coupling between the neck and/or masticatory muscles and the lower limb muscles are necessary, because motor skill is increased greatly in the balance of intercuspation. In an attempt to demonstrate the degree of facilitation with isometric and shortening contractions in the triceps suare muscle during clenching occlusion and co-activation of the neck muscles, our study of the coupling between the masticatory and limb muscles was carried out using a myoelectrographical method. The purpose of this study was to quantify the coupling of the activity levels of the masseter and sternocleiodomastoid muscles during various jaw clenching occlusions, and to explain the relationship between the facilitation of the triceps suare and the activation level of the masseter or sternocleiodomastoid muscles.

## METHODS

Ten healthy adult male subjects were selected from a pool of individuals with normal jaw and neck function. The subject was comfortably seated in a semireclining position with the knee at 120 deg and the ankle fixed at 90deg. The force of voluntary foot plantarflexor contractions was recorded from a force transducer, the gross EMG was recorded from two surface electrodes over the belly of the soleus(SOL) and medial gastrocnemius(MG) muscles, and the single motor unit potentials from this muscle were recorded with the use of fine wire electrodes. The single motor units were monitored on a storage oscilloscope and were recorded on a thermal recorder. Single motor unit firing rates during sustained contractions were measured by manual calculation. Moreover, the maximal voluntary contractions(MVC) for the sternocleiodomastoid and masseter muscles were then recorded during resisted neck flexion effort and biting occlusion. Three repetitions of the maximal contraction efforts were performed, and each maximal effort involved an abrupt onset and offset and was no more than 3s long. Simultaneous recordings of myoelectrical activity were made bilaterally from the superficial sternocleiodomastoid and masseter muscles, by using bipolar surface electrodes. Vibration was applied over the lower mandibular location using a vibrator that was adjusted so that the vibration frequency was 70-120 Hz and the amplitude of vibration was approximately 2 mm. As the vibration-induced responses of the muscle spindle afferents, the vibrator was held manually with a constant force of application(1-5 kg) in the masseter muscle. To ensure a bite force level, visual feedback of its force was provided via an oscilloscope, and subjects performed several practice contractions to familiarize themselves with the experimental paradigm. All clenching and neck flexion were performed by the subjects in positions of maximal intercuspation and maximal flexion. The H reflex responses, however, were elicited in the muscles that act via the SOL(SOL-MNs) and MG motor neurons (MG-MN) by electrical stimulation of the posterior tibial nerves of all subjects, during voluntary clenching occlusion and tonic vibratory reflex(TVR) occlusion, by vibration on the mandible location. The H reflex response was measured five times with stimulation intervals of 10 sec in all experimental protocols. The amplitudes of the H response by the voluntary clenching occlusion and the bite force by TVR are expressed as a percentage of the mean amplitude obtained during the resting activity levels of the sternocleiodomastoid and masseter muscles.

## RESULTS AND DISCUSSION

All subjects showed co-activation of the sternocleiodomastoid muscles during the strong clenching occlusion effort. However, the co-activation of the sternocleiodomastoid muscles was very small during the weak clenching occlusion and the TVR of the masseter muscles. In the TVR of the masseter muscle, a weak contraction developed slowly as the vibration continued and when the frequency of the vibration increased, the contraction disappeared within a few seconds. The low-threshold and high-threshold motor units were recruited into action by the TVR. The low-threshold motor unit was recruited by the low frequency vibrations. In contrast, the high-threshold motor unit was recruited by high frequency vibrations. When a weak contraction maintained by vibration continued , all subjects demonstrated excitation of the SOL-MNs and the MG -MNs during maximal voluntary occlusion, but in the weak bite occlusion from 1 to 4 kg, the excitation of the MG -MNs was only slightly increased. During TVR, however, the excitations of the SOL-MNs and the MG -MNs were clearly increased as compared to the voluntary occlusion. On the other hand, when voluntary isometric contractions and shortening contractions in plantar flexion were performed during voluntary occlusion or TVR, the excitation of the MG-MNs was markedly increased with both bite occlusions, as compared to the excitation of the SOL-MNs. Consequently, the degree of excitation in the MG -MNs is increased, particularly with shortening contractions in plantar flexion, during TVR of the masseter MUs. These results suggest that dynamic muscle movement is strongly linked to the neural connection mediating the activation of the masseter motor neurons .

## REFERENCES

1) Miles, T.S., Poliakov, A.V. and Nordstrom, M.A.: Responses of human masseter motor units to stretch. J. Physiol., 483:251-264, 1995.
2) Turker, K.S., Brodin, P. and Miles, T.S.: Reflex responses of motor units in human masseter muscle to mechanical stimulation of a teeth. Exp. Brain Res., 100:307-315, 1994.
3) Rustioni, A., Sanyal, S. and Kuypers, H.G.J.M.: A histochemical study of the distribution of the trigeminal divisions in the substantia gelatinosa of the rat. Brain Res., 32:45-52, 1971.
4) Abrahams, V.C., Anstee, G., Richmond, F.J.R. and Rose, P.K.: Neck muscle and trigeminal input to the upper cervical cord and lower medulla of the cat. Can. J.Physiol.Pharmacol., 57:642-651, 1979.

# EFFECTS ON MOTOR UNIT ACTIVITY DURING CHANGES OF SKIN TEMPERATURE BY INSTANTANEOUS COLD TREATMENT

M.Yona, M.Muro, C.Tadano and H.Seki

Tokyo University of Pharmacy and Life Science, Tokyo, JAPAN

## INTRODUCTION

It is well known that the threshold forces of different MUs reflect increases and decreases with electric cutaneous stimulation, because this alpha-motoneuron activation tends to mix both the inhibitory and facilitatory components following skin stimulation. However, it has been reported that during maintained isometric contraction, the threshold force and the firing rate of motor units (MUs) are influenced by lower skin temperature (Grimby and Hannerz 1976, Masakado et al., 1991, Yona et al., 1993). We have reported that the recruited MUs within 30%MVC showed a close correlation between the threshold force of the MUs and the skin temperature, because a decrease of the temperature by cold air showed a linear correlation (Yona et al., 1995). Inparticular, the recruited MUs, the MUs of high-threshold force(HT-MUs) , near 30%MVC showed a marked decrease in the threshold force decline of the skin temperature. The recruited MUs of low-threshold force (LT-MUs) at the weak contractions slightly increased the threshold force, but it was unstable. Therefore, the purpose of this study was to investigate the spontaneous and compulsive change of the LT-MU firing rates by instantaneously decreasing the skin temperature.

## METHODS

The subjects who volunteered for this study were five healthy males, and were predominantly right-handed. In the experimental posture, the subjects were seated comfortably in a chair and were secured in position with a wrist band that was connected to an iron bar. The experimental arm was fixed in the cooling box and the subjects were asked to maintain this position. The force of the isometric contraction was established by a force transducer attached to the apparatus. An X-Y recoeder was used to display the exhibited force to the subject at the voluntary isometric force (VIF) at which the stable activation MUs are recruited and at the 20% voluntary isometric ramp contraction (20%VIRC). The 20%VIRC was arranged so that the force flexion exerted by the subject increased linearly to a slop rate of 1.68 ± 0.091kg/s for a period of 3sec. The electric signals were detected by needle electrodes placed on the belly of the elbow flexor. The surface skin temperature of the subject's muscle was measured on the belly of the biceps muscle, using a digital thermometer. In the experimental temperature condition, a spray was used to deliver instantaneous cold stimulation of the skin, and for the continuous cold stimulation of the skin, the air temperature was maintained at-10°C.

Before each trial, the subjects performed 2 maximal voluntary isometric contractions (MVC) for 2 seconds in the cooling box room, and trained for the tracking 20%VIRC on the X-Y recorder for two weeks before the collection of data. Before the start of the experiment, the subjects trained to recruit the stable activation MUs. The conditioning test (Tc) was performed twice at a controlled 20%VIRC for 3sec and the stable activation MUs, and the MUs activities were sampled. After a 15minute interval, the stable MUs activation was measured during instantneous cold stimulation. Thereafter, when the skin temperature was reduced gradually from Tc to 25°C with cold air at -10°C, the MUs activities were sampled during the stable MUs activation and the 20%VIRC. The relationship of the amplitude and the wave form of the EMG signals was analyze with a computer.

## RESULTS and DISCUSSION

The spike-interval (ms) of the MUs was variably affected by the decrease of the skin temperature by the instantaneous cold stimulation. The spike-interval of the affected MUs was clearly extended by the decrease of the skin temperature after the instantaneous cold spray in all practiced subjects. The effect of instantaneous stimulation by the cold spray appeared to be common to all of the subjects. The extension of the MU's spike-interval recovered less quickly from about 500 to 1000 msec. In the continuous slow decline of the skin temperature using the cold air box, however, the extension of the MU's spike-interval was not markedly extended as compared to the instantaneous cold stimulation, because of influence of afferent inputs. Furhtermore, the activation of the MUs in the slow cooling of the skin at 20%VIRC only slightly tdecreased he LT-MUs, by the cutaneous afferents. Consequently, it is indicated that instantaneous skin cooling and slow skin cold stimulation reflect only the LT-MUs activation, via the reflex tract from the cutaneous afferents. These results were the same in our experimental data, which showed that the continuous stimulation of cold air caused, only the LT-MUs activation to be decreased markedly by the reduction of the skin temperature, but that it was increased oppositely by it (Yona et al., 1995). Thereby, the mechanism of the cutaneous receptor complicates the influenced responses of the MUs by both the instantaneous and slowing of the skin cold stimulation. However, it is suggested that the mechanism of the LT-MU's activation is reflected for only few seconds by the cold stimulation of the skin.

## REFERENCES

1) Grimby,L. and Hannerz,J. (1976) Disturbances in voluntary recruitment order of low and high frequency motor units on blockades of proprioceptive afferent activity, Acta.Physiol.Scand., 96, 207-216.
2) Masakado,Y., Kamen,G. and De Luca,C.J. (1991) Effects of percutaneous stimulation on motor unit firing behavior in man, Exp.Brain Res., 86, 426-432.
3) Yona,M., Muro,M., Seki,H., Tadano,C., Nagata,A. and Sunamoto,H. (1993) Relation between temperature and the recruitment order of motor units during isometric voluntary contractions, XIV ISB-Proceedings, 1498-1499.
4) Yona,M., Muro,M., Seki,H. and Tadano,C. (1995) Changes of threshold force of motor units during decreasing of skin temperature, XIV ISB-Proceedings, 1026-1027.

# LOWER-LIMB EMG ACTIVITIES IN A PATIENT WITH SPINAL CORD INJURY DURING WALKING WITH THE AID OF THE WEIGHT-BEARING CONTROL ORTHOSIS

**Narumi Kojima*, Kimitaka Nakazawa**, Shin-ichiroh Yamamoto**, Hideo Yano****

* Department of Exercise, Sports, and Health Sciences, School of Education, The University of Tokyo, Tokyo, Japan

** Research Institute, National Rehabilitation Center for the Disabled, Tokorozawa, Japan

**INTRODUCTION** Even a paraplegic patient can step or walk under conditions of being helped by some external settings. The purpose of our study is examining electromyographic (EMG) signals in a lower limb of a paraplegic patient due to spinal cord injury during orthotic gait.

**METHODS** We examined surface EMG activities in the right leg of a paraplegic patient due to spinal cord injury (Th.7 level complete paraplegia) walking with the aid of the Weight-Bearing Control Orthosis (WBCO) , a newly-developed orthotic walking device for the paraplegic.

**RESULTS** Phasic EMG discharges which synchronized the stepping cycles were observed in lower limb muscles such as the rectus femoris (RF), the tibialis anterior, and so on. In RF, The EMG showed a biphasic feature in each gait cycle; there were two separate discharges in a cycle: the "greater" one and the "smaller" one. We calculated the muscle length and the acceleration of the length of the examined muscles using a musculoskeletal model. The smaller discharges always occured at the phase in which the acceleration of the stretching of the RF was at it's peak. The TA and other muscles also had the component of the two discharges, which synchronized with those of the RF.

**DISCUSSION** The mechanism underlying the phasic discharges observed in the paralyzed lower limb muscles of the patient has not yet been clear. The fact that the smaller discharge in RF always occured when the muscle was most dynamically stretched may indicate that the smaller discharge originate the stretch reflex in RF. Excitatory signals might be transmitted to other muscles such as the TA through some newral networks. For the origin of the greater discharges, we can suggest two major candidates. Some mechanical shock may cause the cocontractive discharges for the first place. Second, load stimulus on the contralateral foot sole might be involved in the greater discharge because it occurs during the stance phase of the contralateral leg. Further investigations will be required in order to tell the origins of both of the greater and the smaller discharges.

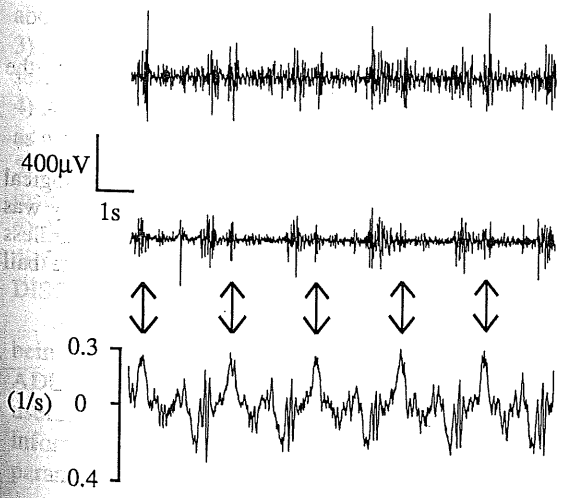



Fig.1 Raw EMG of the TA (top), and the RF (middle) , and muscle length change of the RF per unit time period (bottom) during orthotic gate. All data were recorded for the right leg. Arrows indicate the synchronicity between the "smaller discharge" and the peak of the muscle length change velocity.

# EFFECTS OF PSYCHOLOGICAL PRESSURE ON BALL-STROKING MOVEMENTS

Naoko MATSUMOTO[1], Motoko YANASE-FUJIWARA[2], Tatsuyuki OHTSUKI[3]
[1]Research Institute of Sports Science, ASICS Corporation, Kobe, Japan
[2]Department of Physical Education, Nara Women's University, Nara, Japan
[3]Department of Life Sciences, The University of Tokyo, Tokyo, Japan

## INTRODUCTION

There must be many people who have experienced the psychological pressure when participating in a sports competition. In this case, movements are different from those free from psychological pressure. In order for the players to react the situation appropriately, it is very important to know what is happening in their own body under psychological pressure. There are some investigations about the effects of psychological pressure during rest(1-4), but the effects during the exercise still remain unclear. The purpose of this study was to investigate the effects of psychological pressure on ball-stroking movements in terms of electromyogram and heart rate, by establishing an experimental situation where the subject experienced psychological pressure.

## METHODS

Nine subjects, kept seated in the chair, were instructed to stroke a tennis ball rolling down along a guide rail, using a specially made experimental paddle with their right hand and to put the ball into the target box. Two conditions were given to the subject. In condition PT(pressure trial), subjects were strongly ordered never to fail to put the ball into the target box. In condition non-PT, no such order was given. EMGs were recorded from four muscles in the right upper limb: flexor carpi radialis (FCR), extensor carpi radialis longus (ECR), pectoralis major (PM) and teres minor (TM). Heart rate signals and the ball impact signals were recorded simultaneously with EMGs.

## RESULTS

Heart rate in PT was significantly greater than in non-PT ($p<0.001$). The rate of success was higher in PT than in non-PT for seven out of nine subjects. As shown in Figure 1, the mean EMGs were greater in PT than in non-PT. This tendency was especially clear immediately after the impact. The similar phenomena were also seen in other muscles and subjects.

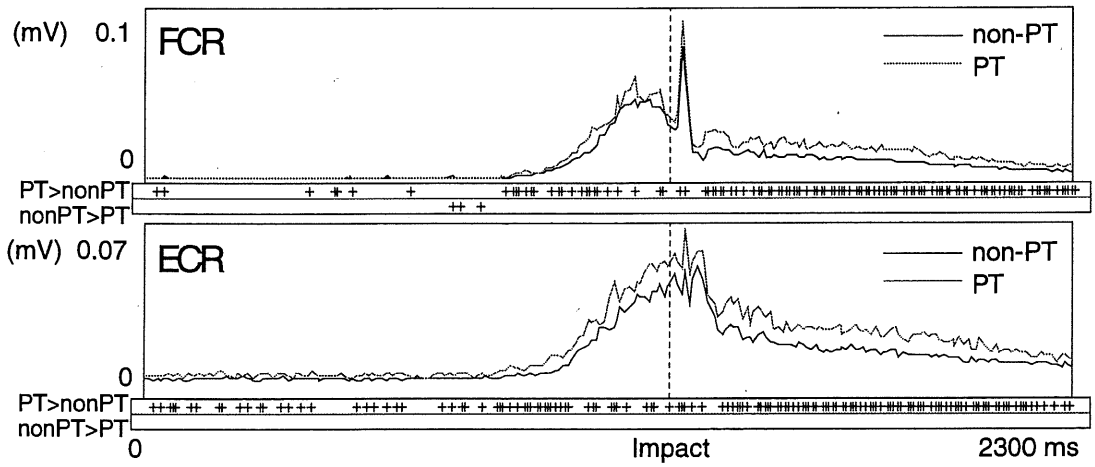



Figure 1: The averaged EMGs obtained from every 10 ms and summarized results of the statistical test for the difference between conditions. + indicates significant difference between conditions PT and non-PT ($p<0.01$).

## DISCUSSION

The fact that the heart rate was greater in PT than in non-PT indicates that subjects felt high psychological pressure in PT as reported by other investigators (1-4). EMG activities were higher in PT and this tendency was especially clear after the impact. Thus it is possible that the psychological pressure itself increases EMG regardless of time phases, but this EMG increase is transiently inhibited during preparatory period for stroking the ball immediately before impact.

## REFERENCES

1. Blanchard, B.E. J. Sports Med. 19: 305-308, 1979.
2. Shapiro, P. A. et al. Arch. Gen. Psychiatry 50: 275-279, 1993.
3. Taggart, P. et al. Brit. Med. J. 4: 130-134, 1969.
4. Videman, T. et al. Med. Sci. Sports 11(3): 266-269, 1979.

# PICTURING ACTIVITIES OF DAILY LIVING

W.H. Groeneveld, K. Kiani, M.P.J.M. van Riel

Dept. of Biomedical Physics and Technology, Erasmus University

Rotterdam, the Netherlands

## INTRODUCTION

Numerical output of measurement systems sometimes requires additional effort to transform the collected data into a graphical representation in order to make this data better interpretable. This is certainly the case when large amounts of kinematic data are acquired through long-term (>8 h) ambulatory monitoring (AM). Collected data-sets usually contain signals from multiple sensors which, even if sampled at a moderate frequency, produce files that are several megabytes large. Interpretation of these time series from a computer screen or printout, is either a tedious or impossible job and certainly makes it unsuitable for clinical practice. In our efforts in developing a clinical system for Ambulatory Monitoring of Motor Activities (AMMA system) the emphasis lies on reduction of necessary operations and procedures. In this, transformation of the time series into a set of understandable and easy interpretable (graphical) formats plays an important role. The system is meant as a supporting tool in Rehabilitation, Geriatrics and Orthopaedics. Monitoring of processes like restoration of function, assessment of (residual) abilities, the effect of (surgical) intervention etc. are the major applications. The potential of AM lies in the fact that collection of data can be done (over several hours) at work, at home or outside on the streets with the subject being unrestrained and (feeling) unobserved.

## METHODS

The AMMA system consists of a number of body-mounted sensors and a digital recording device (8 channels, 8 Mbytes storage capacity) carried by the subject of interest. The sensors measure the acceleration of both upper limbs and of the trunk in both the sagittal and frontal plane. If desired, other variables such as EMG, ambient light level and peak-sound level can be recorded on additional channels. The functions of the recorder (sample rate, number of active channels, communications etc. ) are set from a software shell on a PC which also handles downloading of the data after a recording-session. Software allows visual inspection of the acquired signals and additional processing. Processing modules that are implemented are filtering, simple mathematics and classification. The classification module is capable of recognising four basic categories of activities of daily living (ADL) i.e. walking, standing, sitting and lying from the four accelerometer signals as well as some ADL sub-classes. Software simulated Artificial Neural Networks (ANN) are used to detect the occurrence of a certain activity from the signals. In situations where the signals are difficult to interpret, feature extraction is used to enhance the signal waveforms in order to make them more accurately recognisable by the ANN. The output of the classification process provides numerical information about the time of occurrence of a certain activity. In the following step, the output of the processing modules are used to calculate temporal parameters and to create graphical representations.

## RESULTS

Several types of graphs can be generated on demand in so-called 'Activity Profiles' e.g.:

1) A standard X-t graph of selected parts of the sensor signals.
2) A scatter plot of math-processed accelerometer data from both legs which is used to quickly obtain information about the symmetry of gait (Fig 1).
3) An X-t graph of the recognised ADL-classes over a selected time-slice of the recording.
4) A pie or bar graph of the distribution of the ADL-classes as a percentage of the total recording time.
5) A histogram of the duration of a specific ADL-class (for instance walking) over the total recording time.
6) A 3D graph of the changes in average stride time over different periods of consecutive walking (fig.2).

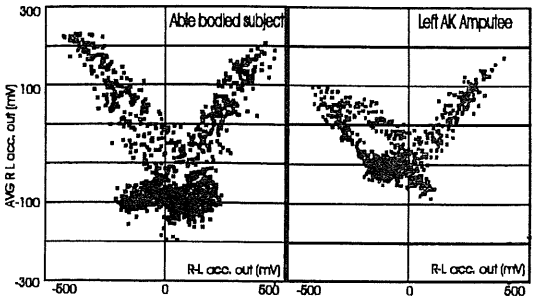


Fig. 1: Example of a scatter plot from accelerometer data

## DISCUSSION

Although the clinical AMMA system is continuously being optimised and we are expanding its features (more ADL-classes or derived parameters) it has proven its usefulness in a number of studies. The system can help the clinician in faster and easier obtaining additional information about the abilities of patients, their compliance with a treatment/training protocol, changes of gait parameters in time (tiredness), effects of intervention, effectiveness of gait, etc. .

# 3D kinematic and kinetic analysis of the biomechanical effects on the walking pattern among patients with drop-foot using the peroneal nerve stimulator

Voigt M. and Sinkjaer T.
Center for Sensory-Motor Interaction, Aalborg University, Denmark. Email: mv@miba.auc.dk

## INTRODUCTION

Paralysis of the anterior tibial and the peroneal muscles causing 'drop-foot' or 'foot drop' is a common consequence of stroke and sclerosis. The drop-foot causes problems with balance and foot clearance during walking. In 1961 Liberson et al. developed the peroneal nerve stimulator (PS) for correction of the drop-foot and different versions of the PS has been widely applied since. The improvement of the walking performance with the PS reported in the literature varies from none (Granat et al. 1996) to a 15% increases in walking speed accompanied with a 37% decrease in walking effort (Burridge et al. 1995). The purpose of the present study was to evaluate how the PS affects the joint loads and the mechanical work outputs from the joints of the lower limbs with focus on possible harmful effects in connection with long term use.

## MATERIAL AND METHODS

Hemiplegic patients suffering from drop-foot who had used their PS regularly during the last two years participated in the study. Thirteen patients were examined. In five cases it turned out that the setting of the PS was wrong and these patients were excluded giving six male and two female patients: age 59.1±8.7 years, height: 1.73±0.08m and weight: 76.1±14.8kg (mean±SD). The 3D movement analysis was carried out according to Davis et al. (1991). The movements were recorded with a four camera MacReflex system at 60 frames $s^{-1}$. Reaction forces were measured with two AMTI OR6 force plates. The patients walked at their preferred speed over the force plates while they where were using the PS and without using it. After a series of practice trials a minimum of 10 trials were recorded in each situation. The analog signal were synchronised with the video signals and sampled at 1kHz and processed off-line. Joint power and work were calculated according to Eng and Winter (1995).

## RESULTS

A large both inter trial and inter subject variability in the walking patterns among the patients was seen. However, common deviations from a normal walking patterns were present e.g.: 1) the total work loads on the lower extremity joints were the same as for normal persons walking about 100% faster (1.6 $ms^{-1}$, Eng and Winter 1995), 2) the relative work load on both ankle joints was 7% higher while the hip joints were unloaded with the same amount and 3) the hip flexor moments were exaggerated during the stance phase causing an increased work absorption by the hip flexors (fig. 1, top). When the PS was turned on seven subjects showed significant increases in walking speed: mean 7.5% (range 3.9-13.5%) from 0.79 to 0.86$ms^{-1}$ on average. The last subject showed no change. The increase was generated by increases in both stride length and frequency. The dorsiflexion during toe off at the affected side was significantly increased. The contralateral leg performed always significantly more work ($p<0.05$) than the affected leg. When the stimulator was turned on the increase in walking speed was reflected as a general increase in both joint moments and joint powers, of both legs with a slight over weight to the contralateral side. However, no significant changes in the load on the lower extremity joints were found.

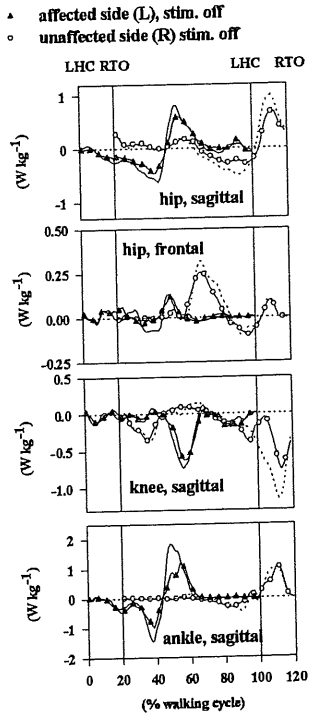



*Fig.1. Average joint power (N=8). The solid and broken curves corresponds to the joint power with the stimulator turned on. LHC,RHC = left and right heel contact times. LTO,RTO = toe off times.*

## DISCUSSION

No potential harmful long term effects of the application of the PS on hemiplegic patients suffering from drop-foot was found. In the present group of patients the use of the PS increased the walking speed significantly and this increase was accompanied with slight increases in joint moments and joint work. The injury causing the drop-foot had given general changes in the walking patterns that kept elevated levels of work loads on the joints while the preferred walking speed was decreased to about 50% of the preferred speed chosen by normal subjects. Therefore the load changes caused by the injury was far more strenuous to the musculo-skeletal system than the changes caused by the peroneal nerve stimulator.

## ACKNOWLEDGEMENTS

To The Danish National Research Foundation.

## REFERENCES

Burridge et al. (1995) Proceedings 5th Vienna International Workshop functional electrostimulation, pp. 367-370.
Granat et al. (1996) Arch Phys Med Rehabil 77:19-24.
Davies et al. (1991) Hum. Mov. Sci. 10:575-587.
Liberson et al. (1961) Arch Phys Med Rehabil 42:101-105
Eng and Winter (1995) J. Biomechanics 28: 753-757.

# THE BIOMECHANICS OF PARTIAL FOOT AMPUTEE GAIT.

M. Dillon[1] and T. M. Bach[2]
[1]North West Hospital, Melbourne, Australia
[2]National Centre for Prosthetics and Orthotics, La Trobe University, Melbourne, Australia.

## INTRODUCTION.

There has been very little quantitative research into the biomechanics of partial foot amputee gait or the biomechanics of prosthetic devices designed to replace foot function. Socket designs have been based on static force analyses which have made unsubstantiated assumptions about external forces applied to the prosthesis (Condie, 1970, Condie and Stills, 1988). The purpose of this study was to provide a basic biomechanical description of partial forefoot amputee gait and to identify the functional losses associated with more proximal levels of partial foot amputation.

## METHOD.

Nine partial forefoot amputees and four normal subjects participated. Amputees used a variety of ankle-foot orthoses and EVA insoles, some with toe fillers to replace or compensate for lost function. The procedure for testing involved a two hour session which included a medical history and physical examination.

Gait analysis was carried out using a VICON motion analysis system and a Kistler 6-component force platform. Data was collected for the affected limb. Kinetic and kinematic analysis were confined to the sagittal plane. Amputees were categorized into a metatarsophalangeal (MP) group (n=4) and a transmetatarsal (TM) group (n=5) based on the level of amputation. Kinematic and kinetic patterns were averaged within each group.

## RESULTS AND DISCUSSION.

Ankle power patterns are illustrated in Figure 1. This figure shows that ankle power was substantially reduced in the MP group and almost negligible in the TM group. A significant relationship was found between residual foot length (expressed as a proportion of intact foot length) and peak plantarflexor power generation (r=.814). This finding was consistent with the prosthetic axiom that function is correlated with residuum length. However the average power curves suggested that if the metatarsal heads were involved, further decreases in foot length did not substantially alter peak power generation.

Substantial changes in moment and power patterns were observed at the hip and knee of the affected limb. These changes were similar to the "hip type" gait observed in transtibial amputees (Winter and Sienko, 1988) and may reflect an attempt to compensate for lost plantarflexor power production.

The results showed a significant relationship between residual foot length and excursion of the centre of pressure (r=.898). Even though transmetatarsal amputees were fitted with orthoses designed to increase the toe lever, these devices appeared ineffective in enabling forces to be transmitted through the forefoot. Static biomechanical models of the partial foot prosthesis assume forefoot loading (Condie, 1970; Condie and Stills, 1988) and therefore seem questionable in light of the finding that ground reaction forces act closer to the ankle axis.

## CONCLUSION

Concerns have been expressed by some authors about the restriction of ankle motion with the use of some prosthetic devices (Condie and Stills, 1988 ; Fillauer, 1976). While ankle motion provides a more cosmetic roll-over during stance phase, the most important reason for retaining ankle ROM is to enable plantarflexor power production. Since power production was negligible in the TM group, it seems unnecessary to preserve ankle ROM when amputation occurs above the metatarsophalangeal joint.

While it is often assumed that function is proportional to the length of the residuum, it appears from this data that ankle power is negligible once the metatarsophalangeal joint has been affected. When the MP joint is affected, surgical decisions should be based on the viability of the residuum and the likely necessity of surgical revision of the amputation rather than the need to retain function.

## REFERENCES

Condie, D. (1970) Prosthetic and Orthotic Practice. Edward Arnold. London. p149.

Condie, D. and Stills, M. L. (1988) In, Murdoch, G. (ed) Amputation Surgery and Lower Limb Prosthetics. Blackwell Scientific. p105-113.

Fillauer, (1976) Orth. and Prosth. 30: 9.

Winter, D. A. and Sienko, S. E. (1988) J. Biomech. 21: 361.

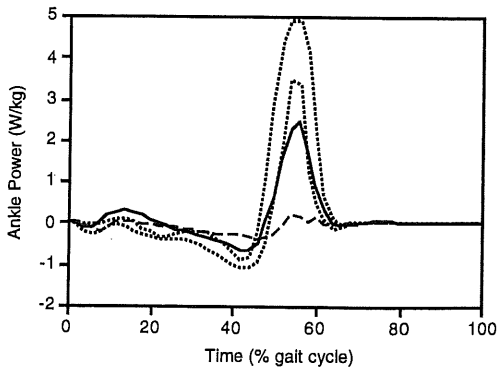



Figure 1. Ankle power. MP and TM groups are shown by solid and broken lines respectively. Dotted lines represent plus and minus one standard deviation around the mean for the normal group.

# GEOMETRICAL ASSESSMENT OF THE RANGE OF MOTION OF LOWER LIMB JOINTS

H. Kuno[1], N. Suzuki[2], K. Akataki[3], M. Watakabe[3], K. Mita[3] and M. Ito[4]
[1]*Chubu University, Kasugai, Japan*
[2]*Izu Rehabilitation and Welfare Center for Children with Disabilities, Shizuoka, Japan*
[3]*Institute for Developmental Research, Aichi Human Service Center, Kasugai, Japan*
[4]*Bio-Mimetic Control Research Center, Riken, Nagoya, Japan*

## INTRODUCTION

The range of motion (ROM) of joint is one of important factors in the biomechanics of human movement together with the muscle force and the neural coordination. The ROM of each joint has been usually assessed with the biarticular muscles relaxed. However, the ROM of hip and knee joint should be determined by a two-dimensional way because their motion is restricted reciprocally while the associated biarticular muscles are fully extended. The aim of this study is to investigate geometrically the ROM of lower limb joints in consideration of the effects of biarticular muscles.

## GEOMETRICAL DESCRIPTION

The interrelationships in the motion between hip and knee joint can be noted by a hexagon on the orthogonal coordinates, whose vertices are related to hip and knee joint angles in six different positions of lower limb (Fig.1). Two sides of the hexagon, $H_f$ and $H_e$ represent the maximal flexion and extension of hip joint, respectively, and $K_f$ and $K_e$ do those of knee joint, respectively. Note that the biarticular muscles, i.e., hamstrings and rectus femoris muscle, are relaxed in the above positions, whereas they are fully extended in the positions characterized by the sides $H_a$ and $R_e$. Thus, the sides, $H_a$ and $R_e$ is concerned with the length of each biarticular muscle.

## MEASUREMENTS

The measurements were performed on right lower limb from 35 normal boys without any histories of trauma and diseases (mean age: 11.3 yr, range: 8.3 to 14.0 yr). A cinematographic analysis was employed for measuring the joint angles in six supine or prone positions as shown in Fig.1. The black round markers were placed at the skin surface on great trochanter, lateral condyle of femur and tibia, and lateral malleolous of ankle. The hip joint angle was defined by the inclination of the longitudinal axis of the thigh relative to the horizontal plane. The knee joint angle was made by the angle between the longitudinal axes of the thigh and leg.

## RESULTS AND DISCUSSION

The side $H_f$, $H_a$ and $R_e$ of the hexagon in elder boys (13 yr) were shifted inside compared with those in younger boys (8 yr), however, the ROM of the knee joint did not change (Fig.2). The results suggest the reduction in the maximal flexion of hip joint and the shortening in hamstrings and rectus femoris muscle due to aging even in children. The age-dependent changes may be caused by imbalance in growth between bone and muscle. The present geometrical method is considered to be useful for systematic and quantitative evaluation of the ROM of the lower limb joints.

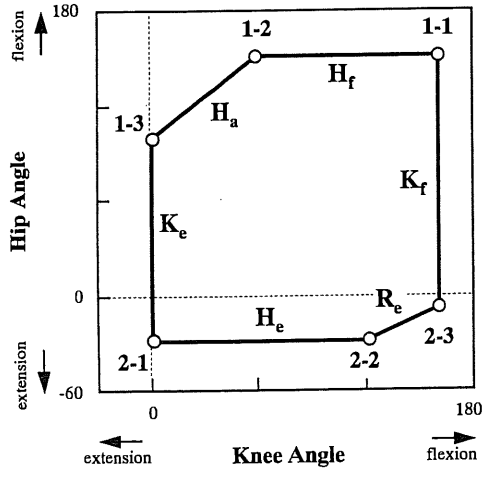



Fig.1 The hexagon expressing the interrelationship between hip and knee joint mobilities on the orthogonal coordinates.

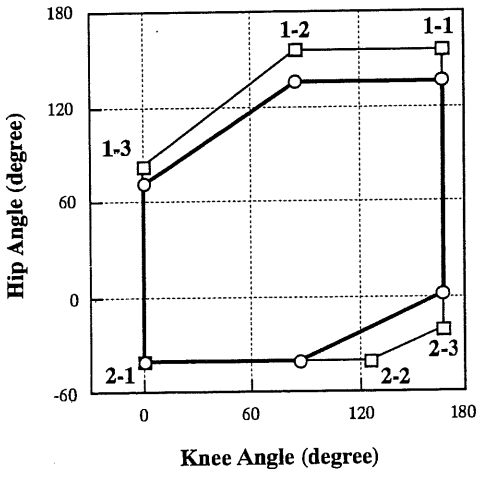



Fig.2 Mean values in normal eight-year-old boys (square) and thirteen-year-old boys (circle).

# STUDY OF ANTERIOR DISPLACEMENT OF THE TIBIA IN MOTION.

Tsutomu Fukui, Sung Hyek Kim
Showa University, Fujigaoka Rehabilitation Hospital, Yokohama, Japan
Yoshikatsu Kuroki, Yujiro Mori, Akihiko Fujimoto and Hiroo Kanai.
Showa University, Fujigaoka Hospital, Yokohama, Japan

## INTRODUCTION

In the research literature on biomechanical analysis, there are some studies designed to examine the performance of anterior cruciate ligament deficient knees. Patients with ACL deficient knees tended to flex their knee in various tasks such as walking, landing and cutting. Researchers have focused upon the movements of deficient knees, however, it is important to note whether these so called quadriceps avoidance movements are the results of insufficient ACL or the cause of giving way.

We have already investigated that anterior-posterior displacement of tibia relative to the femur were related to the center of gravity of whole body in ankle strategy reverse-pendulum task. Also we found support moment, the sum of extension moment of ankle, knee and hip, have coefficient correlation to perpendicular component of ground reaction forces, so in reverse-pendulum task support moment was almost constant.

The purpose of this study were to quantify the anterior displacement of the tibia in voluntary free-pendulum task and to verify the inappropriate movements for rehabilitation of ACL reconstruction.

## METHODS

Subjects were a group of 12 patients with ACL reconstructed knee, a group of 2 patients with ACL deficient knee, a group of 19 healthy volunteers. After a subject's knee joint had been fixed at 30 degrees flexion by a brace, then he was asked to sway his body forward and backward. This movement was measured by three dimensional motion analysis system, a force plate and knee motion analyzer.

## RESULTS

Anterior translation of the tibia corresponded to backward movement of the COG in ankle strategy. In some subjects who moved by hip strategy, anterior translation of their tibia corresponded to forward movements of COG. In all of them anterior translation of tibia had significantly correlated with knee extension moments.

## DISCUSSION

Ankle and hip moments were mutual substitutive as same results with previous studies and support moments were almost constant in this task. In ankle strategy ankle extension moment increased by forward movement and in hip strategy hip extension moment increased by backward movement. Knee extension moment ,the rest of support moment, related to anterior-posterior translation of the tibia.

These results showed that the load of ACL was increased by decrease of ankle extension moment at forward movement and decrease of hip extension moment at backward movement.

# THE ANGULAR DAMPING COEFFICIENT OF ACL DEFICIENT KNEE JOINT WITH AND WITHOUT A FUNCTIONAL KNEE BRACE

M. Lamontagne, P. Johnson, B. Diallo, and J.P. Desrochers
School of Human Kinetics and Dept. of Anatomy and Neurobiology, University of Ottawa, Ottawa, On. Canada K1N 6N5

## INTRODUCTION

Passive elasticity and viscosity of human joints have been considered to be a negligible influence on joint mechanics. This is probably a suitable approximation for slow movements in the midrange of joint motion, however, for more rapid movements and motions approaching the mechanical limits of the joint this assumption is questionable. In the circumstances of damaged intra-articular and peri-articular tissues (i.e. injured ligaments and capsule), the net passive visco-elastic moment of the joint may change. Does a functional knee brace rectify this passive visco-elastic moment pattern of the ACL deficient knee during daily or sport activities?

The aim of the study is to measure the net viscoelastic moment of force (angular damping coefficient) of both the normal and the ACL deficient knee with and without a functional knee brace (Legend Brace: Smith & Nephew Donjoy Inc.).

## METHODS

Seven ACL deficient male and ten males with normal knee joint volunteered for this study. For the normal knee subjects, five subjects were tested on the left leg and five with the right leg under both braced and unbraced conditions. The viscoelastic moment of the knee joint (angular damping coefficient) ($C_k(\theta^*)$) was measured at 0, 45, 60, 90, and 105 degrees of knee flexion using the suspension method based on the small oscillation theory [1-2]. To ensure that the leg musculature was inactive during the testing, the myoelectric activity of the quadriceps and hamstring muscles were monitored with active surface EMG (ME 3000). Both force and angle data were collected via microcomputer (Compaq 386) The data from the braced and unbraced conditions for both the injured and normal subjects were compared to find differences, if any, in the viscous and elastic function of the knee during the preceding conditions. The detailed computational procedures can be found in McFaull [3].

## RESULTS AND DISCUSSION

The angular damping coefficient ($C_k(\theta^*)$) was a polynomial function of the knee flexion angle. For $C_k(\theta^*)$, comparable data in the literature have been obtained for normal subjects[2-3]. The values for the $C_k(\theta^*)$ were lowest at 45 and 60 degrees of knee flexion for the normal and ACL subjects in the unbraced conditions (Figure 1). However, the values for $C_k(\theta^*)$ at 10° and 105° of knee flexion were significantly higher than those at 45 and 60° for the unbraced condition. An ANOVA (2x5) was performed to test for significance between groups for both conditions. No significant differences ($p>0.05$) were found in the $C_k(\theta^*)$ between the ACL-deficient subjects and the normal subjects for either the braced or unbraced conditions. A significant difference ($p<0.0001$) was found between braced and unbraced conditions for all five angles. The use of the functional knee brace increases significantly the angular damping coefficient irrespectively of the degree of knee joint injury. This may be due to mechanical impedance of the functional knee brace or the restriction of the muscle-tendon system to reach their full elongation caused by brace straps surrounding the limb. The increase of angular damping and joint stiffness would require an increase of the mechanical work of the knee joint musculo-skeletal system.

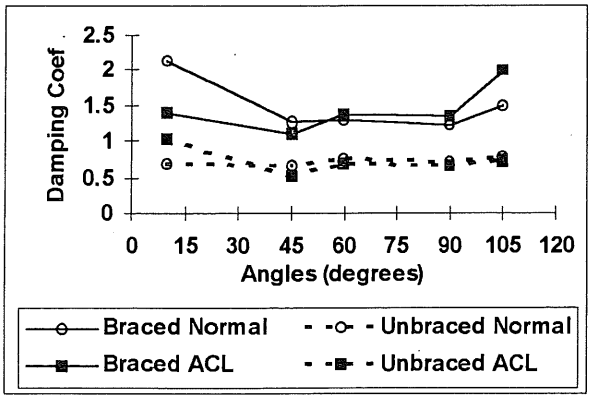



**Figure 1.** Damping Coefficient as Function of Knee angle

## REFERENCES

1. Lamontagne, M. and S. McFaull. Proceedings of the Second North American Conference of Biomechanics, Chicago, August, 1992.
2. McFaull, S. and M. Lamontagne. (1993). The Angular Damping Coefficient Of The *In Vivo* Human Knee Joint. The XVth International Congress of Biomechanics, Paris, p.848-849.
3. McFaull, SR (1993) Unpublished Master Thesis, University of Ottawa, Ottawa, Canada.

# KINEMATIC ANALYSIS OF THE HIP JOINT MOTION IN BASEBALL PITCHING

Kouji Miyashita, Hirokazu Kobayashi, and Kiyoshi Yokoe
Institute of Sports Medicine and Science, Usaka, Agui-cho, Aichi pref.,470-22 Japan

## INTRODUCTION

The forward step takes a part to increase the ball speed for baseball pitching. Some of the baseball pitchers who have injured their shoulders had a limited range of motion (ROM) in the hip join of the stride foot. The purpose of this study was to examine the range of the hip joint movement among baseball pitchers.

## METHODS

5 semiprofessional baseball pitchers who had no injury history were subjects for this study (age:24.4±2.9years, height:1.80±0.03m, weight:77.4±4.1kg). All subjects were right-handed, and over-hand style pitchers.

The pitching motions were filmed with two synchronized high-speed video cameras (200 frames/s) from the rear and right side of the pitcher's mound. Four landmarks (A: right hip joint center, B: left hip joint center, C: left patella center, D: pubic symphysis) of the best pitch in three fastball pitchings were digitized and analyzed by three dimensional DLT method from the stride foot contact to 0.5 sec after. The center of A and B was calculated and indicated as E. The angle between line A-B (cross axis of the pelvic girdle) and line B-C (femur) was calculated by personal computer, which indicates the hip adduction motion (HA). In the same way the angle between line E-D (vertical axis of the pelvic girdle) and line B-C (femur) was calculated, which indicates the hip flexion motion (PFA). These are shown in Figure 1.

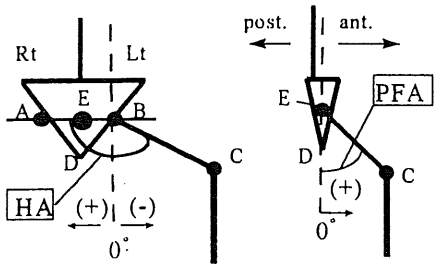



Figure 1. HA&PFA

## RESULTS

Figure 2 shows the average of the changes of HA and PFA. HA was -30.8±9.5° and PFA was 35.1±8.9 at the stride foot contact (SFC). At the time the right shoulder was positioned at the maximum external rotation (MER), HA was -5.5±5.5°, PFA was 30.3±13.4°. HA was 9.0±3.0°, PFA was 22.2±7.2° at the ball release (REL). HA transited 65° for 0.5s and 42° (65% of 65°) appeared on the Late cocking and the Acceleration phase. PFA transited 59° on the Follow-through phase most of the time.

HA was 28.5±7.5° at that time PFA was at it's maximum (80.9±17.5°), which appeared on the Follow-through phase.

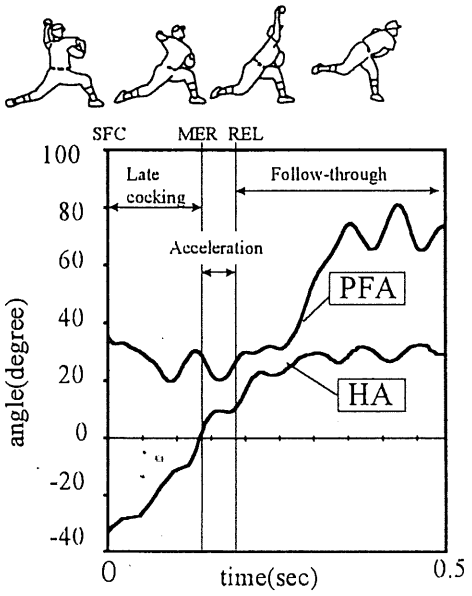



Figure 2.The changes of HA & PFA

## DISCUSSION

Some of the pitchers were injured in their shoulders at the Follow-through phase, whose ROM of the hip joint adduction were limited to less than 10° with 90° flexion (Miyashita,1996). The results show that the angle of hip joint required at the Follow-though phase is about 30° adduction with 80° flexion, and this should be used as an index of ROM of the hip joint in a static position for baseball pitching without injury.

We guess that when the hip joint motion is limited, the kinetic chain would be interrupted and the upper extremity would be injured, since baseball pitching motion is the kinetic chain of the whole body.

# EFFECT OF GAMMA-IRRADIATION ON MICROHARDNESS OF BONE

M. Ramrakhiani
Department of Physics, Rani Durgavati University, Jabalpur, India

## INTRODUCTION

The studies of mechanical behaviour of bones are important for its microstructure. Various combination of techniques have also been reported, however, due to complexity of bone material, the subject is still not well understood. High energy radiation produce free radicals by ionization, whose subsequent interaction with each other or with free charges induces changes in the mechanical behaviour of bone. Presently static as well as dynamic indentation techniques have been used to study the variation in the microhardness of dry bones due to gamma irradiation.

## METHODS

The test samples of approximate size 1x1 $cm^2$ from flat portion of outer compact layer of skull bone were irradiated with gamma rays from cobalt-60 with irradiation dose ranging from 5000 rads to 25 megarads. These were subjected to Vicker indentation test with a few gram load and ball indentations with loads of the order of kilograms. Dynamic indentations were carried out with impact energies of the order of $10^{-2}$ joules. The deformation produced due to indentation was measured and studied microscopically and also by multiple beam interferometry and hardness numbers Hv, Hs and Hd were computed by the formulae.

## RESULTS

Investigation reveals that upto 6000 rads there is negligible effect of irradiation. The variation of hardness numbers with the irradiation dose is shown in Fig. 1. Interferometric investigation shows greater recovery in the depth of indents on irradiated bones.

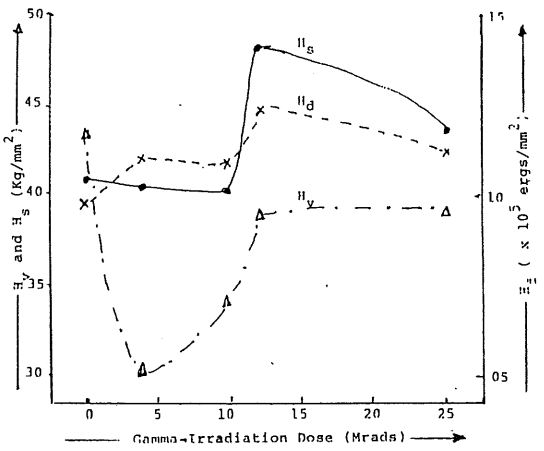



Fig. 1 - Variation of hardness with Y - irradiation

## DISCUSSION

Cobalt-60 emits photons of energy 1.17 MeV which knock out electrons from the atoms of the target, giving wakes of ionization. A number of traces are produced till photon looses energy (Fig. 2). In bones, electrons are knocked out from protein molecules comprising collagen fibres, which travel within the molecule and trapped at trapping sites producing free radicals. These radicals either interact with each other causing cross-links or capture free charges producing rupture, depending on the availability of radicals and free charges. This may change the micromechanical behaviour of the bone.

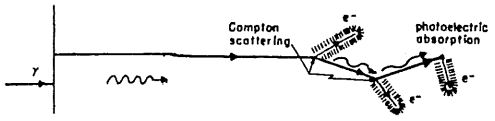



Fig. 2 - Ionization produced by Y - ray photon

The ionization produced by gamma rays varies from layer to layer. It is seen from the figure 1 that as the gamma - irradiation dose is increased, static hardness decreases indicating rupture. The reduction is more in case of Hv, showing prominent rupture at superficial layers, probably due to low radical density, since the gamma - ray energy is absorbed at a little depth. At doses above 10 megarads cross linking becomes prominent, which is greater at deeper layers where radical density becomes high to form cross links. Increased dynamic hardness at all doses shows that weak cross links between distant radicals are also formed which are not broken by momentary forces, but can not withstand prolonged stress. Maximum cross linking occurs due to 10-15 megarads dose at a few micron deep layers. At still higher doses, reduced hardness may be attributed to sufficient availability of free charges.

# Finite Element Analysis of the Stress Distribution within the Seated Buttock and Cushion

Choe-Jan Chiang, You-Li Chou,Hsu-Lung Wang, Shu-Zon Lou
Institute of Biomedical Engineering National Cheng Kung University Tainan, Taiwan

## INTRODCCTION

Persons who spend most of their time seated in wheelchair such as S.C.I.(Spinal Cord Injury) are sensitive to formation of pressure sores, also known as decubitus. Pressure sore is a great concern for our society. It is also a major challenge in the areas of rehabilitation and biomedical engineering at present. Surgeons find that tissue breakdown begins inside the body and works its ways outward to the skin. Therefore, the pressure between the interface and inner will be investigated. In reality, measuring the inner pressure of soft tissue is extremely difficult. However, finite element method(F.E.M.) is a very useful tool to solve the problem. Ultimately, we hope to optimize F.E.A. parameters of material and geometry to integrate with CAD/CAM and design a better custom contoured seated cushion(CCSC) for helping the pressure sores groups to improve their living quality.

## METHOD

Geometry: the geometry model includes the buttock and cushion. Cushions are used from PU foam and its dimension is 40cm $\times$ 40cm $\times$ 7.8cm. The buttock model was established by Chow and Odell model, which gives a nearly axisymmetrical pressure distribution The simulation is performed for one half of the buttock area. So, the study is based on a typical adult with a hip width of 40cm. Bone's material properties are taken from paper average value which E is 17Gpa and $\nu$ is 0.31.

Soft tissues: At present we consider it as linear and homogenous and isotropic and time independent material property. The experimentally determined relationships between force and displacement for soft tissue with contour gage. The linear young's modulus is determined by incorporated a function with least square. For soft tissue, it is nearly incompressible material and $\nu$ is assumed to be 0.49.

Applied force: To maintain a manageable model, a simple form of loading was simplified. For sitting condition with arms and feet support, semibuttock was applied a single concentrated force 29.2 % of the total weight pointing down on the center of the axisymmetry models. Average 10 normal subjects weight is 58.3Kg and its force is about 167N.

Boundary condition: the buttocks and cushion model are 3D problems, and was assumed that only a portion of the right buttock. Axisymmetric model is 2D on the F.E.A.. Therefore it was defined a cross-section rotating about Y-axis and X direction constrained. On the bottom of the cushion, for positioning on a fixed rigid flat, the Y-direction is contained. In addition, contact elements were used on the boundary between the buttock and cushion. It is constraint equation and was produced by means by 2D node to surface. It is also nonlinear analysis.

Solution: assembly nonlinear equations of elements were interactive with full Newton-Raphson method to solve the node displacements and stresses. This model is static structure problem and on a load step was divided into 5-substeps to interactive. Large deformation was considered because of large compressible material for the foam cushion. It belongs to geometry nonlinear static analysis of large strain.

## RESULTS AND CONCLUSION

Displacements and stresses were calculated with the program ANSYS program for both the flat and contour cushion. Comparing the FEA model with experimental measurement, in flat cushion the maximum cushion displacements and buttock-cushion interface pressure of computational results are 4.8cm and 21kPa respectively and the experimental results are about 4.4cm and 10kPa.. So the results are reasonable.

The influence of interface pressure due to different cushion geometry and material property is also studied. The results have the same tendency with experimental clinical data. Contoured and softer cushion shows the lower interface stresses because of the better pressure distribution. The highest interface stress of the buttock is on its lowest point for flat and contour cushion and high internal stresses of soft tissues are near the ischial turberosity. The load transferring is from the bone to the cushion and we find that stress is getting more and more smaller. This is the same as experimental results.

## REFERENCE

1. Chung, K. C., Tissue contour and interface pressure on wheelchair cushions , Ph.D. diss., University of Virginia, 1987.

# THE EFFECT OF TASK-RELATED TRAINING OF SIT-TO-STAND IN A GROUP OF CHRONIC STROKE PATIENTS

J.H. Carr, C.K. Monger and V. Fowler
School of Physiotherapy, The University of Sydney, Australia

## INTRODUCTION

Each year approximately 40,000 Australians suffer a stroke leading in many cases to permanent movement disability. The ability to stand up from a seated position (STS) has been found to be critical to independent mobility in everyday life and lack of independence in this action has been identified as a major factor in increasing the risk of institutionalisation (Branch et al., 1987). The purpose of this study was to examine the effect of task-specific training of STS in individuals at least one year post-stroke. One of the difficulties in demonstrating effectiveness of training in the acute stage following stroke is that of differentiating between "spontaneous recovery" and improvement in performance related to the type and amount of training.

## METHODS

Six hemiplegic subjects, aged from 56 to 73 years participated in the study. Time since stroke ranged from 1 to 8.5. There were 2 females and 4 males, 3 left and 3 right hemiplegias. The study involved a pre-post-test design with subjects participating in a 3 week training program based on the critical biomechanical features of STS including strengthening of the lower limb extensors (Shepherd & Carr, 1994). Subjects were videotaped from the hemiplegic side as they stood up with each foot on a force plate. Functional performance of STS was scored on the Motor Assessment Scale (MAS) (Carr et al., 1985). Walking speed and hand grip strength were also measured. To compare the group means pre- and post-training, Wilcox procedure was used for the ordinal MAS data, while paired sample t-tests were used for all other variables. The significance level was set at $p<0.05$.

## RESULTS

Five of the 6 subjects improved their performance on the MAS, with 2 subjects reaching the highest point on the scale and 3 reaching the second highest point. Mean MAS score was significantly higher at post-test than at the pre-test ($Z = -2.0226$, $p=0.043$). There was a significant decrease in the mean time from thighs-off to the peak vertical force of the affected leg after training ($t=-2.953$, $df=5$, $p=0.032$) indicating that on average the peak vertical force through the affected leg occurred closer to the critical time of thighs-off after training. While there was no signficant increase in peak vertical force produced through the affected leg for the group as a whole, inspection of single subject data revealed that 3 subjects had clear increases, greater than 7.5% body weight. Pre- and post-test measures of movement duration and trunk segment flexion amplitude and velocity were not significantly different. Mean walking speed was significantly greater post training ($t=4.179$, $df=5$, $p=0.009$). There was no signifcant change in grip strength, which was not specifically trained, at post-test.

## DISCUSSION

The results of this study have shown that a group of chronic stroke subjects can improve their performance of STS after a three week period of task-specific training. The fact that the timing of peak vertical force through the affected leg moved closer to thighs-off, where it occurs in healthy subjects, suggests that the force produced by this leg was being utilised more effectively in propelling the body mass into the upright position. The pattern of results suggest that improved timing and coordination may have played a larger role in the improvement in functional performance of STS, than did a strengthening effect. Interestingly, it appears that the early response to strength training in healthy subjects is an increase in skill or "learning" at the neural level rather than hypertrophy (Hakkinen & Komi, 1983).

## REFERENCES

Branch, L.G., Meyers, A.R. (1987) Assessing physical function in the elderly. *Clinical Geriat Med* **3**: 29-51.

Carr, J.H., Shepherd, R.B., Lynne, D. et al. (1985) Investigation of a new motor assessment scale for stroke patients. *Phys Ther* **65**:175-193.

Hakkinen, K., Komi, P.V. (1983) Electromyographic changes during strength training and retraining. *Med Sc Sports Ex* **15**: 455-460.

Shepherd, R.B., Carr, J.H. (1994) Reflections on physiotherapy and the emerging science of movement rehabilitation. *Aust J Physiother* 40th Jubilee: 39-47.

# MAXIMAL VOLUNTARY VENTILATION DURING WATER IMMERSION

M. Kimura* and A. Nagata**

* Graduate School of Human Sciences, Waseda University, Saitama, Japan

** School of Human Sciences, Waseda University, Saitama, Japan

## INTRODUCTION

Headout water immersion might induce a decrease of maximal respiratory function during exercise. The purpose of this study was to determine the relationships between the level of immersion and the maximal respiratory functions during hyperventilation. Then, the maximal voluntary ventilation (MVV) was designed to measure during standing on land, immersion to the xiphoid, and to the neck.

## METHODS

The subjects used in this experiment were 5 male volunteers (24.4±1.0yrs, 167.8±3.2cm, and 62.1 ±4.9kg). Pulmonary functions and MVV were measured under three conditions: standing on land (STL), immersion to the xiphoid (WIX), and to the neck (WIN). Measurement of forced vital capacity (FVC), forced expiratory volume in one second ($FEV_{1.0}$), peak flow (PEF), maximal expiratory flow at 75% ($\dot{V}_{75}$), 50% ($\dot{V}_{50}$), and 25% ($\dot{V}_{25}$) of FVC, and MVV were measured by Autospirometer ST-460 (Fukuda Denshi). And then, tidal volume (TV-MVV) and respiratory frequency (RR-MVV) were also calculated during MVV measurement, that were perfonmed for 12s with free breathing frequency and tidal volume.

## RESULTS

Comparing with STL condition, there were significant decreases in all parameters of pulmonary functions during immersion to the xiphoid. Furthermore, during immersion to the neck, FVC, $FEV_{1.0}$, and $\dot{V}_{25}$ were significantly lower than STL. MVV and TV-MVV were significantly decreased, whereas RR-MVV tended to decrease during immersion.

**Table** MVV, TV-MVV and RR-MVV during standing on land and immersion to the xiphoid and to the neck

n = 5

| | standing on land | immersion to the xiphoid | immersion to the neck |
|---|---|---|---|
| MVV (ℓ/min) | 131.1± 13.9 | 115.7± 10.6 (−11.6±2.9%) | 108.7± 11.8 (−17.0 ±4.5%) |
| | | ** | ** |
| TV-MVV (ℓ) | 2.04± 0.97 | 1.78± 0.82 (−12.4±5.2%) | 1.56± 0.68 (−22.1 ±14.2%) |
| | | ** | * |
| RR-MVV (f) | 75.5± 26.7 | 77.0± 29.4 (+1.2 ±7.0%) | 82.0± 29.6 (+9.4 ±15.8%) |

Values are mean ± SD. *: $p < 0.05$ **: $p < 0.01$ "(mean ±SD%)" refers to % change calculated as "(in water −on land) ⁄ on land × 100".

## DISCUSSION

Prefaut et al. reported that a small airway in lung was closed during immersion, and this small airway closure was caused by redistribution of blood into the thorax. The blood redistribution should promote engorgement of blood vessels in the peribronchial sheath, and would decrease airway diameter. The decrease of $\dot{V}_{25}$/PEF in our study might be caused by the reduction of small airway diameter generated by the redistribution of blood. And then, Koubenec et al. reported that blood redistribution into the thorax should promote a low compliance by the stiffness of lung tissue, which make it more difficult to compress alveoli. The stiffness might be caused by the decrease in recoil of lung tissue. Probably, the low compliance would decrease the expiratory functions indicated by the decrease of FVC and $FEV_{1.0}$.

There was a high correlation between $FEV_{1.0}$ and MVV in the present study ($r=0.85$, $p<0.001$). This relationship suggested that the decrease of expiratory functions such as FVC and $FEV_{1.0}$ should promote the low MVV. And then, the decrease of MVV and TV-MVV were more remarkable than that of $FEV_{1.0}$ during immersion. The air-trapping phenomenon and the increase of RR-MVV might be occurred. The air-trapping might happen during high frequency ventilation just like MVV. The decrease of expiratory functions, and the additional effect of the air-trapping would decrease MVV and TV-MVV more remarkably. On the other hand, RR-MVV might increase to prevent MVV from decreasing when TV-MVV was decreased by immersion. This reason might be that the decrease of tidal volume and the increase of respiratory frequency during hyperventilation mght diminish the ventilatory efficiency by the increase of dead space occupying in total ventilatory volume.

# ANALYSIS OF THE INTENSITY OF THE PRONE BACK EXERCISE TO STRENGTHEN THE BACK MUSCLES

A. Plamondon, C. Marceau, S. Stainton, P. Desjardins
Laurentian University, Sudbury, Ontario, canada

## .INTRODUCTION

The exact role of exercise in the management of low back pain still remains questionable despite it's widespread use. Rationales for exercise in the management of back pain include relieving pain, strenghthening weak muscles, regaining range of motion, improving posture and mobility. The clinical consensus is that exercise is an important factor to the success of back pain rehabilitation. However, little is really known about the efficiency of the back exercises to strenghten the back muscle. The purpose of this study is to use an inverse dynamic rigid-body modeling approach combined with electromyographic (EMG) data to examine the ability of the prone back exercise to strengthen the trunk extensor muscles.

## METHOD

Five male subjects in good physical condition, with no previous history of low back injury volunteered for this study(age: 21-23; average mass 76.5 ±6.9 kg; mean height 1.78 ±.08 m). Each subject performed two sets of six exercises. These consisted of three maximal isometric voluntary contractions and three back exercises. The subject in the maximal voluntary contractions (**MVC**s) was lying prone on a table and was positioned so that the hips were near the edge of the table top and the upper body was suspended off the end of the table. The arms were positioned parallel to the trunk and the legs and thighs were secured to the table with straps. The trunk was resting on a mobile support which fixed the starting angle of the exercise. An extensiometer was used to measure the static force of the subject. This extensiometer was installed to a fixed base at one extremity, and the other extremity was attached to the subject via a belt placed in the upper back at the level of the xiphoid process. At the signal of the experimenter, the subject raised a few millimeters so that he did not contact the support and applied a maximal isometric force for six seconds. Three positions from the horizontal were experimented: 0°, 30° and 60° and two repetitions per position were required, for a total of six MVCs'. Between each repetition, the subject was placed in the horizontal position in order to maximally benefit from the resting period of 2 minutes (minimum). Subjects' starting positions in the dynamic prone back extensions were similar to those required in the maximal isometric contractions: 0°, 30° and 60°. The trunk was resting on the mobile support which fixed the starting angle. At the signal of the experimenter, the subject raised his trunk in the horizontal position, holded a second and returned to the mobile support. Two repetitions per position were required, for a total of six dynamic prone back exercises. A peak motion measurement system were used to collect the 2D positions of the anatomical markers on the subject during the exercises. A dynamic multi-segment model was used to estimate the joint reaction moment (**JRM**) at L5/S1. In addition, muscular activity of the lumbar erector spinae and rectus abdominis was collected using bipolar surface electrodes. EMG signals were transformed using a root-mean-square processing method with a time interval size of 100 msec. The maximum EMG values during the dynamic exercise were normalized relative to the maximal values observed during the MVC.

## RESULTS AND DISCUSSION

The MVCs obtained about the L5/S1 joint were in average 162 (SD=21) Nm , 236 (35) Nm and 361 (61) Nm in respectively the 0°, 30° and 60° position. If the mass of the upper body is taken into consideration the MVCs increase in average to 262 (35) Nm, 331 (46) Nm and 427 (68) Nm. The maximum joint reaction moment averaged 108 (12) Nm during the dynamic back exercises and did not differ between the positions because it occurred approximately when the trunk was horizontal. The maximum intensity (100 * JRM/MVC) of the prone back exercise was 67% of the MVC in average. On the other hand, the maximum level reached by the erectors spinae muscles averaged 36% (12) of the maximum. The intensity to start the back exercise differed from 64% (3) to 38% (4) and to 13% (1) respectively for the 0°, 30° and 60° position. From these results, the prone back exercise is more efficient near the horizontal and this position should be maintained longer in order to strengthen the back.

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# FEATURE OF POWER SPECTRUM OF PATHOLOGICAL TREMOR

Hitoshi MAKABE and Kazuyoshi SAKAMOTO
Dept. of Communications and Systems,
The University of Electro-Communications, Tokyo , Japan

## INTRODUCTION

Physiological tremor is defined as invisible vibration of body part. Pathological tremor is physiological tremor for patient. The mechanism for physiological tremor has been studied so that the main factors are known to be both spinal reflex system and central nervous system(Sakamoto et al.), while the study of pathological tremor has proceed to medical diagnosis and the central nervous system has been considered to play an important role for the mechanism. In the study, the feature of power spectrum of pathological tremor could be evaluated chiefly for the aged patients with Parkinsonian disease and cerebellum disease as compared with normal aged persons. The judgment of stages of the symptoms investigated.

## METHODS

The pathological tremor for right index finger are detected by accelerator sensor (MT-3T,Nihon-Kohden). In the measurement, subject continues to gaze the target at pointed end of the index finger to hold the finger horizontal direction. This tremor is called postural tremor. The measurement time is one minute. Subjects are 18 aged patients for mean age of 67 years and 10 aged normal persons for mean age of 66 years. The pathological tremor during loading 50g weight is also measured. The accelerator components obtained are analyzed to obtain the power spectrum by auto-regression model(i.e., AR model). The peak frequency and the peak power which is the power spectrum at the peak frequency are evaluated.

## RESULTS AND DISCUSSION

The results for Parkinsonian aged patients and the normal aged persons are described. The relation between peak power and peak frequency for both the patients and the normal persons is showed in Fig.1. The peak frequencies at two peaks for the patients are significantly lower than that for the normal ,and the peak powers for the patients are significantly larger than that of the normal. When the stage of disease proceed to be wrong, the peak power increases significantly, but the peak frequency does not show the reduction. Especially, the tendency is remarkable for the peak power at first frequency around 10 Hz. As for the results in the condition of weigh loading, the first peak frequency and its peak power are useful index to discriminate the value for the normal persons.

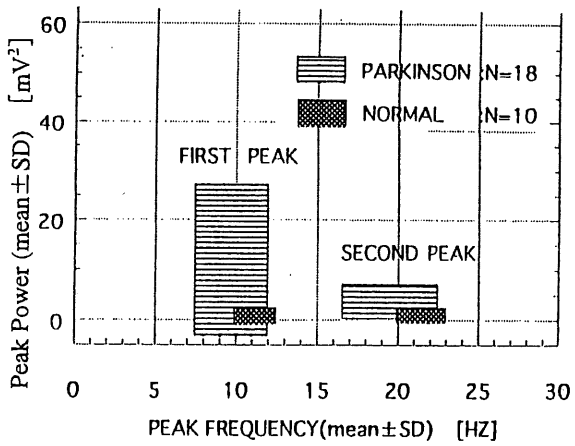



Fig. 1 Relation between power spectrum and peak frequency

## CONCLUSION

Pathological tremor for the aged patients and physiological tremor for the normal aged ones are measured. The power spectra are obtained by AR model calculation, so that peak frequencies and their power spectrum (peak power) are evaluated. The difference for both subject groups could be discriminated with the use of peak frequency and peak power. It is found that the peak power at first peak frequency around 10 Hz is useful index of the judgment of the stages of patients.

## REFERENCE

Sakamoto,K.,Itakura,N.,Nishida,K.,and Zhou L. : Study of Function of Fingers by Physiological Tremor. J.Thermal Biology,18,665-669,1993.

# THE EFFECT OF MAGNITUDE AND DURATION OF PRESSURE ON CEREBRAL CORTEX IN THE RAT

Ming-Shih Tsai [1, 2], You-Li Chou [1], Guan-Liang Chang [1] and Ching-Liang Shen [3]
[1] Institute of Biomedical Engineering, National Cheng Kung University, Tainan, Taiwan
[2] Department of Neurosurgery, Provincial Tainan Hospital, Tainan, Taiwan
[3] Department of Anatomy, College of Medicine, National Cheng Kung University, Tainan, Taiwan

## INTRODUCTION

This biomechanical study was aimed to investigate and establish the pressure-time-damage relationship on cerebral cortex using a rat model. Additional cortex injuries induced by traction during brain surgery have been an important clinical issue. Unfortunately, very few research studies have been conducted to provide basic information and clinical guidance for the acceptable loading in the operative procedure. The ultimate goal of this research is to enhance the biomechanical understanding of pressure magnitude and duration factors on cortex injury, and to provide clinical guidance with alarm systems during the surgery for minimal damage of cerebral cortex .

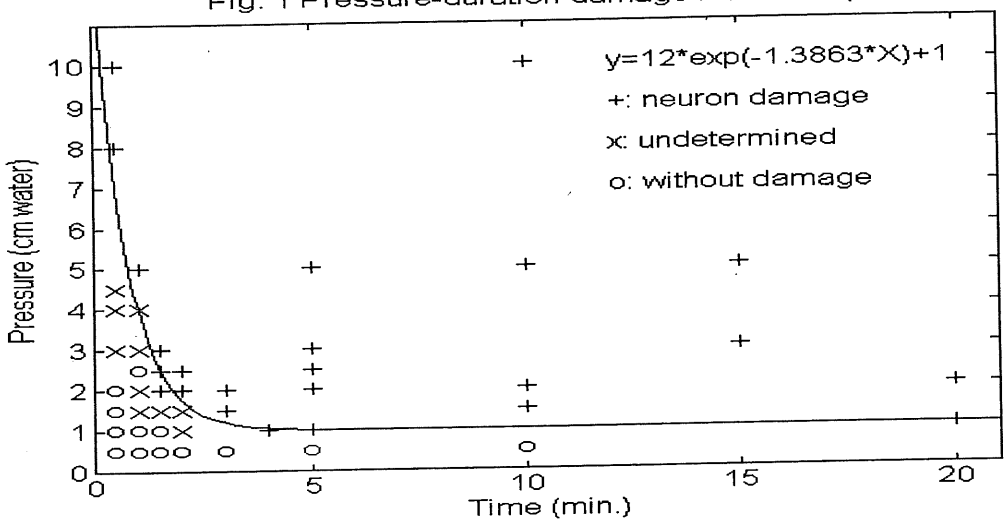
## METHODS

Thirty rats with weights ranging from 300 to 500 grams were conducted using a modified central venous pressure device, which was fixed on a stereotactic frame. Hydrostatic pressure loadings controlled by the pressure device were applied to bilateral cortex of the rats after the craniotomy. The pressures ranging from 0.5 to 10 $cmH_2O$ with 1or 0.5 cm increment and the loading durations from 0.5 to 20 minutes with 0.5 or 1 minute increment were applied to the rats with room temperature at 25 ℃. The rats were scarified at three to five days after the experiment, and the brains were removed for histopathological evaluations using light microscopy. The damage of cortical neurons from the histological examination were analyzed and correlated with the applied pressures and the corresponding load durations. The tolerable and damaged ranges for cortical neurons were quantitatively determined from the varying combination of pressures and durations.

## RESULTS

The results of pressure-duration-damage relationship for cortical neurons are shown in Fig.1. Using Least-Square-Error Fitting method, the exponential curve ( i.e. $Y = 12e^{-1.3862944X} + 1$) is established to quantitatively characterize the tissue damage and tolerable regions in terms of pressure-duration loading.

Fig. 1 Pressure-duration-damage relationship




## DISCUSSION

The major finding demonstrates that the area (Fig.1) above the curve with pressure higher than 1 $cmH_2O$ and to the right with time longer than 30 seconds is highly susceptible to cortex neuron damage. In the acute stage of clinical course, an increased intracranial pressure up to 20 mmHg (approx. 2.7 $cmH_2O$) signals an risky condition. The information generated from this study seems to be clinical relevance and particularly useful in applying retractor on brain cortex to reduce damage during surgery. This may also provide criteria for design of an alarm system equipped with surgical instruments for pressure monitoring clinically. Further biomechanical studies of shear force effect, stress-strain distribution and the corresponding damaging region at the cortex interface are warrented for the clinical application.

# THE EFFECTS OF CLIPLESS FLOAT PEDAL ON THE ARTICULATIONS OF THE LEG IN CYCLING

F. Maronneaud, F. Barbier, J.C. Angué
Laboratoire d'Automatique et de Mécanique Industrielle et Humaine
URA CNRS 1775, Université de Valenciennes, France

## INTRODUCTION

The study deals with the characterization of differences between clipless fixed and clipless float pedals in cycling. Opto-electronics systems and force platforms allow quantitative analyses of movement but do not provide any explanation about the constraints they generate. These last information, articular stress, are unobservable but may be calculated via a previous biomechanical modelling of the body. The analysis of these efforts allows to evaluate the constraints generated by the two kinds of pedal on the joints.

## METHOD

A bicycle used in cycling sport competition and a home trainer constitute the experimental device. The bicycle is fixed by its front fork and this device allows to simulate the seated position of a cyclist on the road. The articular efforts are computed from an iterative code (Barbier 1994). The inputs of this program are the positions of anatomical points and the external efforts and their application points. The position of the leg were measured simultaneously with the pedal loads. The experimental device was presented and evaluated in (Maronneaud 1997). Anthropometric parameters specified for each segment were the lenght, moment of inertia (Hanavan 1964), mass and center of gravity location (Dempster 1955). In addition, quintic splines are used to smooth the inputs of the program. Experimental data were collected from the right leg of a male subject during cycling at 90 rpm and 200 W at a sampling rate of 50 Hz for 30s. Prior data collection, subjects cycled at least 15 min at 85 rpm and a reduced power level of about 120 W to familiarize themselves with each pedals design. Each curve is averaged over 15 cycles.

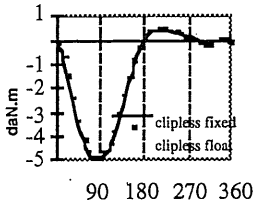



Fig 1 : The effective transmission from the shoe to the pedal

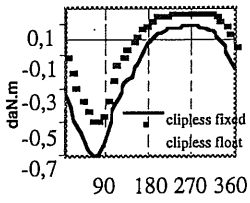



Fig 2 : The longitudinal torque of the ankle

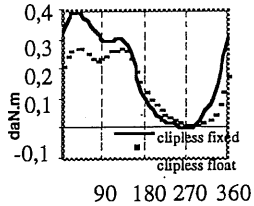



Fig 3 : The longitudinal torque of the knee

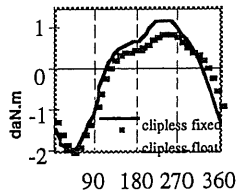



Fig 4 : The longitudinal torque of the hip

## RESULTS

On the one hand, the quite similar curves of figure 1 state that there is no loss of effective mechanical torque transmission from the shoe to the pedal when floating pedal systems are used. On the other hand, the study of the articular variables states that the longitudinal torque of each articulation is weaker when the floating pedal is used. The longitudinal torques were indicators of the axial moment realized at the ankle, the knee and the hip. Limitation of longitudinal torques reduce trauma to the leg during pedalling. When floating pedal systems are used, the decreased magnitude for the ankle (Figure 2) and the hip (Figure 4) were respectively 15 % and 13 %. With clipless float pedals, for the knee (Figure 3), differences are more significant, the decreased magnitude is 30 %. In fact, from an anatomic point of view, during the cycler's flexion-extension movement, there exists a compulsory rotation of the tibia under the femur (Kapandji 1980). Observing the longitudinal torque of the knee shows that the floating pedal allows a significant reduction of this torque.

## CONCLUSION

The cycler's biomechanical modelling leads to the calculation of articular efforts on the whole of the inferior limb (hip, knee, ankle). The analysis of the articular variables of the ankle, the knee, and the hip brings to the fore that the floating pedal allows a significant reduction of the longitudinal torque of the leg. Clipless float pedals quantifiably reduce main axial moment from the inferior members without compromising power transmitted to the bike.

## REFERENCES

Barbier F. (1994). Modélisation biomécanique du corps humain et analyse de la marche normale et pathologique-appplication à la rééducation-, thèse de doctorat n°94-16, Université de Valenciennes.

Dempster W.T. (1955). Space requirement of the seated operator, Wright Patterson Air Force Base, WADC-TR, p55-159.

Hanavan I.A. (1964). A mathematical Model of the human body, A.M.R.L.-TR, 64-102, Aero Medical Research Laboratories, Wright Patterson, A.F. Base OHIO.

Kapandji I.A. (1980). Physiologie articulaire : schémas commentés de mécanique humaine, Fascicule II, membre inférieur, ISBN 2 224-00060-X, cinquième tirage, Maloine S.A. Editeur, 27 rue de l'école de médecine, Paris, 1980.

Maronneaud F., Barbier F., Angué J.C. (1997). Experimental device for the analysis of the cycler's movement, International Society of Biomechanics in Sport, in soumisssion, Denton, Texas, USA, July 1997.

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# CYCLING KINEMATICS AT THREE POWER OUTPUTS - BILATERAL DIFFERENCES

R.J. Neal[¶], P.E. Martin[‡], T.D. Royer[‡], P.F. Vint[‡], S. McLean[¶] and D.A. Aitken[¶]
[¶]Department of Human Movement Studies, The University of Queensland, QLD, AUSTRALIA
[‡]Exercise Science Research Institute, Arizona State University. Tempe, AZ, USA

## INTRODUCTION

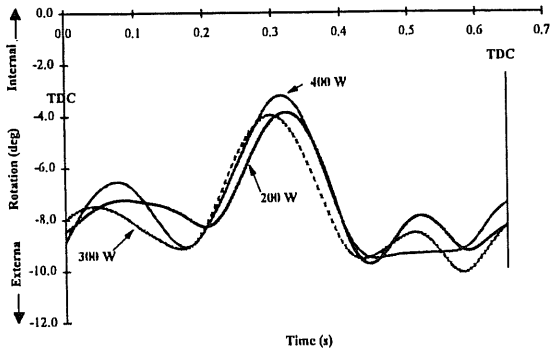
Overuse injuries are commonplace amongst highly trained athletes involved in sports that require repeated application of submaximal loads over thousands of cycles (e.g., long distance running, cross-country skiing and road cycling). One injury that is common to competitive and to a lesser extent, recreational cyclists, is knee pain (*chondromalacia patella*). This pain is believed to be associated with the repeated application of loads typical of a normal pedalling cycle and usually occurs in only one of the two lower limbs (Hull & Ruby, 1996). Some cyclists never develop knee pain whilst others continually battle with it during their competitive careers. While it is probable that the loads cause the tissue damage and/or inflammatory response, the kinematics at the joints must reflect the loading patterns. One movement believed to be associated with knee joint pain is internal/external knee joint rotation. While numerous studies have examined two-dimensional cycling technique, no studies have been published that describe the three-dimensional (3D) motion of the knee and ankle joints during cycling. As a first step in attempting to identify a link between pain and joint motion, the 3D kinematics of pain-free, elite cyclists must be quantified to provide the necessary baseline data for comparison. Thus, the purposes of this research were to describe the knee joint motion of a group of pain-free cyclists, as they rode at various power outputs and to compare the kinematics of their left and right limbs.

## METHOD

Fifteen elite male and female subjects (A-grade level or higher) rode an Excalibur bicycle ergometer, with a setup that duplicated their road bikes, at three power outputs (200 W, 300 W & 400 W) while the motion of external markers placed on their lower limbs was digitised at 200 Hz using a four-camera Motion Analysis Corporation system. Eight pedal cycles were captured for each power output and the 3D coordinates were generated. These coordinates were used in a spatial model that allowed calculation of clinical rotations using Grood and Suntay's (1983) joint coordinate system. Dependent measures of medio-lateral knee displacement, flexion-extension and internal-external range of motion at the knee and ankle, were calculated for each pedal cycle. Two-way (leg X power) ANOVAs were used to evaluate the effects of the independent variables. An $\alpha$ of .05 was chosen as the probability level for differences.

## RESULTS AND DISCUSSION

Figure 1 illustrates typical knee internal/external knee rotation during pedal cycle, at the three power outputs. Flexion-extension of the knee was typically around 80 deg., consistent with previous research (Broker & Gregor, 1996) while the range of motion for internal-external rotation was variable among subjects, ranging from only 5 deg in one subject up to a maximum of 25 deg in another. Significant asymmetry ($p<.01$) was found for medio-lateral knee displacement, with subjects displaying greater range of motion in one knee than the other. There was a significant increase ($p<.01$) in internal-external knee rotation and a trend ($p=.06$) for increased knee flexion-extension as a function of power output. No differences were found between limbs or across power outputs for the three ankle joint rotations.




These results show that bilateral asymmetry of medio-lateral knee joint displacement is typical of elite cyclists and is evident despite the fact that these subjects were free of pain. Power output has a modulating effect on the internal-external knee joint range of motion and flexion-extension range of motion with the likelihood that the increased pedal forces required to produce the high power output, also caused increased knee joint motion.

Figure 1. Typical internal/external right knee joint rotation (S14)

## REFERENCES

Hull, M., & Ruby, P. (1996). Preventing overuse knee injuries. **High tech cycling**. Human Kinetics.
Broker, J., & Gregor, R. (1996). Cycling biomechanics. **High tech cycling**. Human Kinetics.
Grood, E. & Suntay, W. (1983) A joint coordinate system for the clinical description of three dimensional motions: Application to the knee. **J. Biomech. Engng 105**:136-144.

# EFFECTS OF FOOTWEAR ON POSTURE CONTROL OF RUNNING TARGET SHOOTERS

J. T. Viitasalo*, P. Era**, K. Norvapalo*, H. Mononen*, K. Mononen* and M. Salonen*
* Research Institute for Olympic Sports, Jyväskylä, Finland, ** Metitur Ltd, Jyväskylä, Finland

## INTRODUCTION

In many shooting events special shoes are used in order to have better contact with the floor and to improve posture control. Scientific research reports on the effects of footwear on balance and on shooting itself are rare, especially in running target shooting (Olympic rifle), in which an air rifle is used for shooting a moving target at a distance of 10 m. The target runs the 2 m long shooting area in 5 s (slow run) or in 2.5 s (fast run). The present study was designed to compare, three various footwear conditions, body swaying while standing with eyes opened and closed, body swaying during aiming, area of oscillation around the target, and shooting score.

## METHODS

Six Finnish running target shooters representing high international and national levels volunteered for the study. Their mean (± SD) age, height and weight were 22±4 years, 1.80±0.09 m and 82±14 kg, respectively. All the tests were performed in exactly the same way for three conditions; the subjects wore their own shooting shoes (SS), conventional jogging shoes (CJS) and only socks (S). In each of the three conditions the subjects were first asked to stand still for 30 s with their eyes opened (EO) and fixed on the motionless target, and then for 30 s with their eyes closed (EC) while keeping their arms by their sides. These were followed by 30 slow (target speed 0.4 m*s-1) and 30 fast (0.8 m*s-1) run shots (15 to the left and 15 to the right). The subjects used their own air rifles. A force platform and a special computer program were used to calculate deviation of the center of pressure (COP) in x (target running direction) and z (perpendicular to the target running direction) directions. While shooting, these parameters were calculated only for the aiming phase (while the target was followed at a horizontal level) ending at trigger squeeze. The oscillation in aiming at the target area was registered by a computerized Noptel ST-2000 optical training and analysis system, and the deviation in horizontal (x) and vertical (y) directions were calculated during the aiming. One way analysis of variance was used to compare the three footwear conditions.

## RESULTS

The mean shooting scores for the three footwear conditions (SS, CJS and S) for the slow run were 9.82±0.76, 9.88±0.78 and 9.98±0.56, respectively, and for the fast run 9.20±2.21, 9.41±1.05 and 9.46±0.99, respectively. Swaying of the body COP in x-direction while standing with eyes opened and closed were 0.87±0.2 mm and 0.82±0.17 mm for the SS condition, 1.02±0.28 mm and 0.94±0.31 mm for the CJS condition, and 1.01±0.44 mm and 0.78±0.30 mm for the S condition, respectively, and in z-direction 2.67±0.84 mm and 3.36±0.95 mm for the SS condition, 2.62±0.92 mm and 3.17±0.82 mm for the CJS condition, and 2.12±0.65 mm and 2.62±0.92 mm for the S condition, respectively. During aiming of the slow runs the swaying of COP in x-direction was 2.54±0.71 mm, 2.35±0.57 mm and 2.44±0.66 mm, and during aiming of the fast runs 1.92±0.74 mm, 2.10±0.62 mm and 2.08±0.50 mm for the SS,CJS and S conditions, respectively. In z-direction the swaying of COP during the slow run was 1.89±0.60 mm, 2.04±0.74 mm and 1.83±0.58 mm, and during aiming of the fast runs 1.86±0.33 mm, 1.83±0.29mm and 1.78±0.31 mm for the SS,CJS and S conditions, respectively. Deviations in the aiming point in the horizontal (x) direction were 7.48±2.06 mm / 16.32±4.86 mm, 6.99±1.99 mm / 14.27±3.03 mm and 7.87±2.22 mm / 16.25±5.16 mm , and in the vertical direction (y) 6.77±1.95 mm / 13.31±5.03 mm, 6.87±1.63 mm / 16.57±8.68 mm and 7.01±1.73 mm / 15.68±6.37 mm during the slow/fast runs for the SS, CJS and S conditions, respectively. The analysis of variance did not reveal any statistically significant ($p<.05$) differences between the three footwear conditions.

## DISCUSSION

In our previous studies the current variables have been found to be very sensitive for differences between shooters of high international caliber. However, in the current study we did not find any significant effect of footwear on parameters describing posture control and shooting score in running target shooting at the group level, though for some individual shooters a trend for differences according to expectations was found. This may suggest, that footwear is not very important in posture control of running target shooting, or that the shooters were able to control and compensate for the effects of the differences in ground contact between the three footwear conditions on parameters describing posture control.

# FORCE AFFECTING BETWEEN BUTT AND SHOULDER DURING AIMING PHASE IN RUNNING TARGET SHOOTING

H.V. Mononen and J.T. Viitasalo
Research Institute for Olympic Sports, Jyväskylä, Finland

## INTRODUCTION

In hunting a firm grip on the rifle is essential to absorb the recoil of the rifle while triggering. In Olympic rifle (running target shooting, air rifle) recoil is not an issue. On the other hand, a firm grip has thought to ensure aiming stability. Coaches often advise pulling the rifle firmly against the shoulder and keeping the force level steady during the aiming phase. Even though this seems logical in order to achieve better control of the rifle it has not been discussed in scientific literature. This report presents force parameters which describe the rifle grip and their association with aiming.

## METHODS

The subject group consisted of 8 international level running target shooters (G1), 8 national/regional level rifle shooters (G2) and 8 hunters (G3, no systematic training for competitions). The subjects mean (± S.D.) age, height and weight were 22±4, 37±12 and 47±11 years, 1.80±0.09, 1.78±0.08 and 1.83±0.04 m and 82±12, 82±12 and 82±8 kg, respectively. Subjects performed 30 slow run (target speed 0.4 m/s, 10 m distance) shots in series of 10 shots with 5 minute rest intervals between the series. Three specially constructed transducers (for different rifle butt structures) were used to measure force between rifle butt and shoulder. The force signals were on-line digitized (12 bit, with 100 Hz sampling frequency) and stored for further analyses. Off-line analyses included calibration (voltage related to physical values), signal conditioning (3 point moving average), marking (recording of different alteration points in the force signals) and calculation of various time and force variables. The aiming trajectories were registered with a computerized Noptel ST-2000 optical analysis system with a 100 Hz sampling frequency. The aiming deviations in vertical ($Y_{SD}$) and horizontal ($X_{SD}$) directions were calculated during the aiming phase. Maintenance of the butt force during the aiming phase is expressed as percentual drop ($\%F_D$) and it was calculated as percentual difference of $F_{Bmax}$ (maximal force during aiming) to $F_{Bshot}$ (force level at shot) . Analysis of variance (ANOVA, One-way) was used to compare force variables between the groups and Pearson correlations were used to study the interrelationships between variables.

## RESULTS

The mean shooting scores (± S.D.) were 9.5±0.9 (G1), 8.8±0.6 (G2) and 5.8±0.8 (G3), mean values for $F_{Bmax}$ were 129.0±47.6, 158.9±77.9 and 63.3±28.2 N and mean values of $\%F_D$ were 14.4±12.9, 29.6±12.1 and 35.1±13.6 %, respectively. Within all subjects $F_{Bmax}$ showed statistically significant correlations to score ($R=.62, p<0.005$), $Y_{SD}$ ($R=-.54$, $p<0.05$) and $X_{SD}$ ($R=-.59$, $p<0.01$), but this could not be found in group level. The analysis of variance showed statistically significant difference ($F=5.51$, $p<.05$) between groups in $\%F_D$ and the T-test showed that G1 differed statistically significantly from G2 ($p<0.05$) and G3 ($p<0.01$). While combining G1 and G2 statistically significant positive correlations were observed between $\%F_D$ and $Y_{SD}$ ($R=.69$, $p<0.05$) and $X_{SD}$ ($R=.79$, $p<0.01$), but no statistically significant correlations were found between $\%F_D$ and scores ($R=-.33$, $p=.298$). In Addition the correlations between score and $Y_{SD}$ and $X_{SD}$ did not reach the level of significance ($R=-.62$, $p=.054$ and $R=-.50$, $P=.144$). On the other hand, within all subjects correlations between score and $Y_{SD}$ and $X_{SD}$ were highly statistically significant ($R=-.64$, $p<0.005$ and $R=-.71$, $p<0.001$, respectively). Also, no statistically significant differences were found within groups in force variables while comparing performances where the target was moving to left and to right.

## DISCUSSION

In running target shooting aiming time is restricted and a moving target is makes aiming even more difficult. Therefore accurate aiming is a difficult task. The whole performance can be affected by posture before and during aiming phase, success in lifting the rifle, rifle handling during the aiming phase and trigger control. A stable force level between butt and shoulder during the aiming phase showed to have a positive affect on aiming for experienced shooters. Even though the $\%F_D$ did not highly correlate with the shooting score, the high correlation with $Y_{SD}$ and $X_{SD}$ can be interpreted as enabling a chance for a good score. Different variables showed that national level shooters acted rather similarly to international level shooters. This could be based on roughly the same amount of training (see the mean age) and goal oriented coaching. On the contrary, the training background of hunters might not be sufficient to ensure a good overall performance in this specific shooting task. Therefore the current relations between $F_{Bmax}$ and aiming variables should be interpreted carefully. In addition, performance can be constructed with different performance strategies. Even in G1 shapes of force-time curves differed in the early phase of aiming. Furthermore, the use of the front hand, ratio of force produced with both hands, and relative positions of upper extremities and shoulder line during the aiming phase should be taken into consideration.

# POSTURE CONTROL IN COMPETITIVE AND NOVICE SHOOTERS IN RUNNING TARGET RIFLE SHOOTING

K. Norvapalo*, P. Era**, H. Mononen*, K. Mononen*, M. Salonen* and J.T. Viitasalo*
* Research Institute for Olympic Sports, Jyväskylä, Finland ** Metitur Ltd, Jyväskylä, Finland

## INTRODUCTION

In many kinds of sports events the role of balance is essential. Good whole body stabilization is one of the prerequisites for successful shooting performance (Era et al 1996). When shooting at a moving target, balance is highly important (Mason 1987). The purpose of this study was to analyse the differences in posture control between highly trained top-level shooters, national level shooters and untrained novice shooters .

## METHODS

The subjects (n=24) were Finnish male rifle shooters and running target rifle shooters. They were divided into three groups; group A consisted of internationally top-level running target rifle shooters, group B of national level rifle shooters and group C of untrained novice shooters with little shooting experience. Their mean (± S.D.) age, height and weight were 22±4, 37±12 and 47±11 years, 1.80±0.09, 1.78±0.08 and 1.83±0.04 m and 82±14, 82±12 and 82±8 kg. The subjects shot 30 slow, running target rifle shots at a shooting distance of 10 m. The Noptel ST 2000 optical analysis system measured the horizontal (x) and vertical (y) deviation (amplitude) of the rifle's oscillation on the target during shooting. Subjects stood on a force platform during shooting. The mediolateral -x (parallel to the direction of the target motion) and anteroposterior direction -z ( perpendicular to the target motion direction) coordinate values were detected by the force platform with a sampling frequency of 100 Hz. A special computer program was used to calculate the total displacement (mm) and velocity ($mm*s^{-1}$) of the center of pressure (COP). The deviation and velocity of COP were calculated for the aiming phase which started when the lift of the rifle was ready (the rifle pointed at target area) and ended at trigger squeeze. The aiming time ranged from 2 to 4 seconds in slow shots. Also the swaying (x- and z- direction) of the COP with eyes opened (30s) and closed (30s) in normal standing position was measured on the force platform before shooting. The statistical significance of the differences between the groups was calculated using oneway analysis of variance (ANOVA).

## RESULTS

The amplitude of body swaying in normal standing position (eyes opened) at the z-direction was smallest among groups A and B ($F=8.60$, $p<0.05$). Swaying differences at the x- direction were not significant between the groups. Swaying velocities in up-right standing position were greater with eyes closed than eyes opened in z-direction ($p<0.05$) among all groups . The amplitude of COP during the aiming phase was biggest in group C, both at the z-direction ($F=19.72$, $p<0.05$) and at the x-direction ($F=10.67$, $p<0.05$). The top-level and the national level shooters had the lowest movement velocities of COP during the aiming phase in both directions; z ($F=12.97$, $p<0.05$) and x ($F=8.42$, $p<0.05$). Also the deviation of the rifle's oscillation on the target was most stable in group A, followed by group B and finally group C, both at the horizontal-x ($F= 12.16$, $p<0.05$) and vertical-y direction ($F=18.21$, $p<0.05$).

## DISCUSSION

The current study showed that the posture control and the handling of the rifle during aiming was steadiest among the top-level running target rifle shooters and most unstable among novice shooters. Body swaying as indicated by the COP in normal standing position was much smaller among the top-level shooters compared to the novice shooters. This result in the running target shooting is similar to the observations of Aalto et al 1990 in normal rifle and pistol shooting. The current results also showed that the national level shooters were rather close to top-level shooters in the analysed parameters. The reason for this could be that the study was done in training conditions and the shooting was done only in slow target speed. Also, the average age of group B was higher and therefore the total amount of training was roughly at the same level in both groups. The force platform method and the optical laser-rifle system was shown to differentiate groups of shooters with different training backrounds in a logical way.

## REFERENCES

Aalto, H., Pyykkö, I., Ilmarinen, R., Kähkönen, E and Starck, J. Postural stability in shooters. Otol. Rhinol. Laryngol, 52, p. 232-238 1990.

Era, P., Konttinen, N., Mehto, P., Saarela, P., Lyytinen, H.: Postural stability and skilled performance - A study on top - level and naive rifle shooters. J. Biomechanics, Vol 29, No 3, pp. 301-306, 1996.

Mason, B. R., Burton, G. R.: Ground reaction forces of elite Australian trap shooters. In Jonsson, B. (eds.). Biomechanics X - B: proceeding of the Tenth International Congress of Biomechanics, 1987, p. 797-800.

# POSTURE CONTROL IN RUNNING TARGET SHOOTING

J. T. Viitasalo*, P. Era**, K. Norvapalo*, H. Mononen*, K. Mononen* and M. Salonen*
* Research Institute for Olympic Sports, Jyväskylä, Finland ** Metitur Ltd, Jyväskylä, Finland

## INTRODUCTION

In running target shooting the role of balance is considered important. Good balance is thought to give a foundation to perform stable vertical body rotations while aiming at a moving target during a 2 m horizontal run at a distance of 10 m. The present study was designed to investigate interrelationships between body swaying while standing eyes opened and closed, body swaying during aiming, swaying of the aiming point around the target and shooting score.

## METHODS

22 subjects, 6 national and international level running target shooters (group A), 8 regional level running target shooters (group B) and 8 moose hunters (group C) with no systematic shooting training background, volunteered for the study. Their mean (± S.D.) age, height and weight were 22±4, 37±12 and 47±11 years, 1.80±0.09, 1.78±0.08 and 1.83±0.04 m and 82±14, 82±12 and 82±8 kg, respectively. Each subject was asked to stand motionless for 30 s with eyes opened and fixed on a motionless target, and for 30 s with eyes closed while keeping their arms by their sides. This was followed by 30 slow run shots (target speed 0.4 m*s-1). A force platform and a special computer program were used to calculate the deviations (amplitude) of the center of pressure (COP) at the mediolateral (x) and anteroposterior (z) directions in relation to the target direction of movement (= x). While shooting, these parameters were only calculated for the aiming phase. The shooting scores, oscillation in aiming and deviations in horizontal (x) and vertical (y) directions were registered by a computerized Noptel ST-2000 optical training and analysis system. Pearson's product-moment correlations were calculated between the parameters.

## RESULTS

The mean shooting scores were 9.5±0.8, 8.8±0.6 and 5.8±0.8 for the A, B and C groups, respectively. When all the subjects were analyzed as one group, the shooting score was significantly correlated with the oscillation of the gun around the target (x-direction r=-.636, p<.01; y-direction r=-.712, p<.001), with amplitude of the COP swaying during the aiming (x-direction r=-.620, p<.01; z-direction r=-.819, p<.001) and with amplitude of the COP swaying while standing eyes opened (x-direction r=-.630, p<.01; z-direction r=-.692, p<.001) and eyes closed (z-direction r=-.517 p<.05) ; for the higher shooting scores the oscillation of the gun, swaying of the COP while aiming and while standing were smaller than for the higher shooting scores. Oscillation of the gun in the y-direction was positively correlated with swaying of the COP in the z-direction (r=.536, p<.05) and in the x-direction (r=.534, p<.05) while aiming. Also swaying of the COP during aiming in z-direction was positively correlated with the z-direction swaying of the COP while standing with eyes opened (r=.828, p<.001) as well as while standing the eyes closed ( r=.739, p<.001). When tests were repeated for all the three groups separately and while keeping the groups A and B as one group, the only significant relationships were for the group C between swaying of the COP in the z-direction during standing with eyes opened and during the aiming (r=.775, p<.05), and between swaying of the COP in the z-direction during standing with eyes closed and during aiming (r=.830, p<.05).

## DISCUSSION

The current results showed that for shooters with a heterogeneous background in systematic shooting training, the ability to maintain good body balance (minimal swaying) while standing with eyes opened and closed also predicts a good ability to control body posture during the aiming phase in running target shooting. It seems, that good posture control during aiming, especially in the anteroposterior direction, is connected with minimal swaying of the aim point around the target in the vertical direction. Minimal swaying of the aim point on the target both, in horizontal and vertical directions, seems to predict good scores. These relationships were not found among the trained groups (A, B and their combined group) which may show that variables other than body balance (swaying) are important for a good score. The low correlation coefficients may also have been influenced by small variations in the observed variables among the trained subjects. Among the less trained C-group the positive correlation between body swaying while standing and during aiming may mean that good body balance is important for minimal swaying during aiming. The fact that their scores were not related with swaying of the COP may mean, that among less trained shooters other variables such as unstable trigger squeezes may be more important.

## POSTURE STABILITY OF SHOOTERS

Wen-Lan Wu, Fong-Chin Su, Wen-Dung Lee, Chia-Cheng Chang
Institute of Biomedical Engineering, National Cheng Kung University, Tainan, TAIWAN

### INTRODUCTION

Sensory information about posture is important for balance control. The main sources for this information are the vestibular organs, visual system, proprioceptive system and exteroceptive pressoreceptors. The contribution of each sensory modality may also be changed by the help of specific balance training. In the study of Aalto et al. (1990), the results indicated that the shooters used an increased amount of proprioceptive and vestibular cues to stabilize their posture. In many sports events the role of balance is obvious. The stability in shooting is more important than that of other sport events. The top-level shooters could stabilize their posture during the last seconds just before the shot in order to increase the accuracy of shot (Era, 1995). Thus the ability of good instant posture control is necessary for shooters. The purpose of this study is to find out if there are differences in posture stabilizing abilities between shooters and controls, test the sensitivity of three sway parameters in discriminating changes in posture stability, and weigh the value of two factors, eyes and foot positions, for the stability of shooters and controls.

### METHODS

The subject characteristics are listed in Table 1. The test was consisted of four sensory conditions : (1) comfortable stance eyes open, (2) comfortable stance eyes closed, (3) narrow stance eyes open, and (4) narrow stance eyes closed

Table 1 The basic data of subject

| | subject numbers | age (y/o) | height (cm) | shooting experiences (years) |
|---|---|---|---|---|
| shooters (male) | 6 | 17-30 (21.17) | 167-182 (173.33) | 2-8 (4.8) |
| shooters (female) | 5 | 16-18 (16.8) | 155-169 (161.8) | 1-4 (1.6) |
| controls (male) | 8 | 24-28 (25.33) | 168-183 (173.75) | |
| controls (female) | 11 | 21-23 (22) | 162-167 (163.4) | |

A Kistler forceplate system was used to measure the posture stability. While the subject stood on the platform with barefoot, the ground reaction forces were sampled at 100 Hz for 15 seconds for each trial. Then, the center of pressure (COP) on the platform could be determined. In this study, two parameters related with sway velocity and one parameter related with location of COP were adopted to describe the postural stability. They were the maximum sway velocity ($V_{max}$), mean sway velocity ($V_{mean}$), and dispersion index (Dis). The dispersion index was defined as the standard deviation of the COP dispersion. Finally, two-sample t-test was used to evaluate the statistical significance of the differences between shooters and controls. Paired t-test was used to compare the means between different sensory conditions within each group.

### RESULTS

The results revealed that the sensitivity of two measures, $V_{max}$ and $V_{mean}$, in discriminating the stability between the all sensory conditions were high ($p<.01$) except for the comfortable stance-eye closed condition (male, $V_{mean}$). But the sensitivity with Dis measure was poor ($p>.05$) except for the narrow stance-eye closed condition (female) ($p<.01$). Also the shooters had significantly reduced body sway measured by sway velocity and Dis when compared to the control subjects in most sensory conditions. The stability of controls could be contributed by eyes and foot position, since significant differences occurred not only on different positions but also on different eye conditions ($p<.05$). However, the stability of shooters could be contributed only by foot position ($p<.05$).

### DISCUSSION

This study revealed that the shooters had significantly better stability than untrained control subjects. The same findings also had been shown on former study (Aalto,1990). Based on the results of this study, both the maximum sway velocity and the mean sway velocity had good concurrent validity for stability test, but not for the dispersion index. The visual cues in shooting could be compensated by the other sensory organs. On the other hand, the standing positions played an important role for postural control both in shooters and controls. In general, the postural testing presented in this study could give valuable information about individual shooters and might be used in training to improve the shooters' capabilities (using vestibulospinal pathways). It also could be used for selecting shooters with high capability.

### REFERENCES

1. Aalto H, et al., Otol. Rhinol. Laryngol. 1990; 52:232-8.
2. Era P, Konttinen N, J. Biomechanics. 1996;29:3:301-6.

# AUTOMATIC METHOD FOR CYCLE EXTRACTION AND SEGMENTATION IN HUMAN GAIT KINEMATIC DATA

**Emmanuel CORDIER, INRIA Rhône-Alpes**

**Grenoble, France**

## 1. INTRODUCTION

The goal of the proposed method is the automatic detection of heel contact (HC) from the kinematic registration of human walk, on levels or on inclines, with or without shoes.

This method allows us to obviate the use of footswitches, does not require to know the slope angle, and does not require the often unreliable calculation of velocities and accelerations.

The immediate application of this method is in the automatic extraction of cycles (from HC 0% to HC 100%) in continuous human locomotion data. The method can also be generalized to cut each cycle to obtain segments delimited by others important events classically used in human gait study.

## 2. METHOD

We start with the position data registered from retro-reflective markers attached on five anatomical landmarks on the subject's skin: external malleolus (ankle), fibula head (knee1), lateral epicondyle of femur (knee2), great trochanter (hip) and trochiter (shoulder). From this data we can obtain the raw knee and hip angle data signals (KN and HI) which we subsequently pass through a very light smoothing filter. Then the angles are centered on zero.

We then construct a composite signal (fig. 1) as in the following

$$\text{CS} = k, \text{ if } (\text{KN} + c \times \text{HI}) > t,$$
$$\text{CS} = 0, \text{ otherwise.}$$

where $k$ is an arbitrary constant, $c$ is a coefficient and $t$ is a threshold value expressed as a percentage of the maximum value of $(\text{KN} + c \times \text{HI})$.

We thus obtain a signal consisting of thin rectangles (fig. 2) whose descending sides are said to coincide with the calculated heel-contact (CHC). In order to validate the method, we can calculate the average of the difference in the time intervals (MI, mean interval), between the CHC and visually determined heel-contact instants (VHC).

## 3. RESULTS

The outlined method was used on the treadmill slope walking data of a single subject in various conditions: variable slopes from -12° to +12° with 1° step, wearing soft and very hard shoes. It was found that $t = 75\%$ and $c = 1.20$ give the best overall values of MI. Depending on the conditions of walk, the typical MI values were between 13ms and 75ms (mean 29ms) corresponding to 0.9% to 5.0% (mean1.9%) of the average cycle period.

If we plot MI as $c$ varies from zero to a large value (for example 15) we find that the error asymptotically converges to a value of 150ms, in the worth case (fig. 3). Thus even without an optimally chosen coefficient $c$ the method seems to work, more or less accurately. However, for the fixed best value $c = 1.2$ we tend to get spurious or missing CHC for threshold $t < 70\%$ or $t > 80\%$,.

## 4. DISCUSSION

The robustness of the method needs to be tested by applying it on the data taken from several subjects as well as from overground walk (as opposed to treadmill walk).

A true validation will require the comparison of the CHC identified by this method to that obtained from force platform data. This is on future project.

It will be interesting to see if the method corresponds to an equivalent mathematical treatment and if the values of $c$ and $t$ depend on the body segment lengths used for the calculation of angles.

The method helps us in precisely detecting the moment of heel contact uniquely from kinematic data (hip and knee angles). It does not need the use of any sophisticated device but simply small markers attached onto the subjects. The calculations are fast and simple. We can thus use it for biomechanics experiments conducted inside or outside the laboratory.

Finally, the method appears to be easily generalizable. Segmentation with other linear combinations of the joint angles can be used for detecting other characteristic instants of a walk cycle.

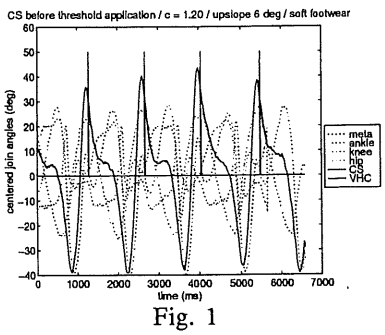



Fig. 1

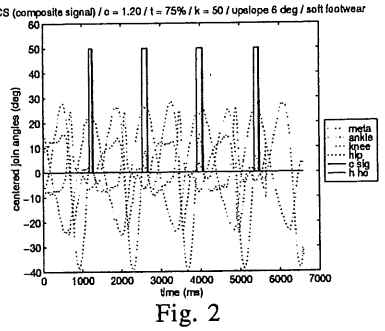



Fig. 2

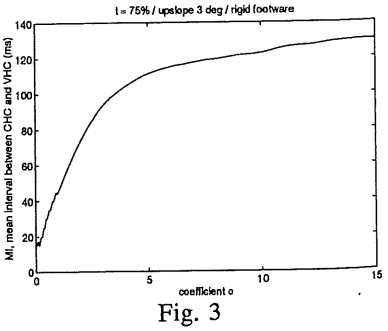



Fig. 3

## REFERENCE

1. Wall J.C., Crosbie J.: "Accuracy and reliability of temporal gait measurement". Gait & Posture, Vol. 4 (1996), p.293-296.

# MECHANICAL ENERGY CONTRIBUTION OF THE SEGMENTS TO HUMAN LOCOMOTION.

S.C.Corrêa[1,2], A.C. Amadio[2], U. Glitsch[3], W. Baumann[3]
1 - School of Physical Education, Federal University of Rio de Janeiro, UFRJ, Brazil.
2 - School of Physical Education, São Paulo University, USP, Brazil.
3 - Institute of Biomechanics, German Sport University, Cologne, Germany.

## INTRODUCTION

The mechanical energy contribution of the several segments to the total mechanical energy of the body varies according to the form of locomotion - walk or running - and to the velocity of running. Due to different procedures of energy calculation the results presented in the literature are hardly comparable (Corrêa, 1996). The aim of this research is to compare the curves of potential, kinetic and rotational energy of the lower and upper extremity in walking at 1.5 m/s and running at 3.0 and 4.0 m/s.

## METHODS

Six male subjects were filmed with two video-cameras (Sony-50Hz) while walking at 1.5 m/s and running at 3.0 and 4.0 m/s on the treadmill. From the kinematics we performed a 3D analysis after a manual digitizing process. The analysis was based on a 13 segment model, represented by 17 markers placed on: ear, and on right and left shoulder, elbow, wrist, hip, knee, ankle, heel and front foot extremity. Positions of segmental centers of gravity, segmental weights, and moments of inertia were estimated on the basis of tables devised by Dempster (1955); the segmental lengths were estimated as a percent of body height (Drillis & Contini, 1966), both as revised by Winter (1979). From the different forms of mechanical energy, we have calculated the potential, kinetic and rotational energy at each instant of time for each segment, using basically the equations described by Zatsiorsky *et al.* (1987).

## RESULTS AND DISCUSSION

Our results for the kinetic energy of each segment differ from the described in the literature because they have been calculated in relation to the center of mass of the subject while the other authors (Winter *et al.*, 1976) considered the absolute velocity of the segments. When comparing the energy curves of the lower and upper extremities for walking and running we observe: a) there are no differences in the energy variations for the upper extremity: b) for the lower extremity the variation is 10% greater in running; c) the contribution of the variation in the lower extremity is 4 times the contribution of the upper extremity in walking and 6 times in running. Observing the segments of the lower extremity we noted: a) for both forms of locomotion the greater contribution for the energy variation in thigh is the potential energy, and for leg and foot is the kinetic energy; b) in walking only the thigh shows two peaks in the total energy variation while in running both thigh and leg show this pattern; c) in walking the contribution of energy variation of thigh and leg for the total variation of the lower limb occurs in an equal proportion while in running the contribution of leg is greater. The contribution of the variation of thigh is lower in running because there is a greater increase in the kinetic energy than in the potential and as the curves are in opposition they partly cancel their increases.

When comparing the variation of the energy curves from the lower and upper extremities in function of the running velocity we observe: a) there are no differences in the energy variations for the upper extremity: b) for the lower extremity the variation is 100% greater in the greater velocity; c) the contribution of the variation in the lower extremity is 10 - 20 times the contribution of the upper extremity. Observing the segments of the lower extremity we noted at the greater velocity: a) the pattern of the thigh's curve is altered, because the potential energy remains the same but occurs a great increase in the kinetic energy, with the total energy variation decreasing; b) there is an increase of 100% in the total energy variation of the leg, due to increases mainly in the variation of the kinetic energy.

We can conclude that for the lower extremity, with the passage from walking to running and with the increase of the velocity of running, the contribution of the energy variation of the thigh decreases with a great increase in the energy variation of the leg.

## REFERENCES

Corrêa, S.C. (1996) Methodological approach for the determination of mechanical energy: application in the biomechanics of human locomotion. Unpublished Ph.D. Dissertation, School of Physical Education, São Paulo University, Brazil.

Winter, D. (1979) Biomechanics of human movement, New York, John Wiley.

Winter, D.; Quanbury, A.O.; Reimer G.D. (1976) Analysis of instantaneous energy of normal gait. J. Biomech. 9, 253-257.

Zatsiorsky, V.; Aleshinsky, S. Y.; Jakunin N. A. (1987) Biomechanische Grundlagen der Ausdauer, Berlim, Sportverlag.

# UNCERTAINTY IN 3-D JOINT MOMENTS ASSOCIATED WITH HUMAN GAIT

Brian L. Davis, A.S. de Vasconcellos and Thomas M. Lundin
Department of Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, OH, USA

**INTRODUCTION**: Errors in human gait analysis arise due to the inherent uncertainty of the data collection process. For example, typical errors associated with ground reaction data are 1%, typical marker coordinate error is 5mm, and typical anthropometric data errors range from 5-10%. Gait analysis is used in a variety of research and clinical settings -- a search of the literature revealed nearly 200 publications between 1992 and 1996 which used these methods. While uncertainties associated with two dimensional gait analyses have been examined (Davis, 1992; Andrews and Mish, 1996), there has not been an effort to examine the uncertainties associated with 3-D gait analyses. Thus the purpose of this investigation was to determine the uncertainties associated with 3-D joint moments in human gait analyses using the inverse dynamics approach described by Vaughan et al. 1991.

**METHODS**: Two methods were used to determine errors associated with the calculation of an independent variable: 1. Known errors can be added or subtracted to the inverse dynamics equations which would result in the largest error possible for joint moments ("worst case" scenario); or 2. An uncertainty analysis (Equation 1, Davis, 1995).

1. $$W_R = [\ (\delta R/\delta x_1\ w_1)^2 + (\delta R/\delta x_2\ w_2)^2 + \ldots + (\delta R/\delta x_n\ w_n)^2\ ]^{1/2}$$

$W_R$ refers to the uncertainty associated with the dependent variable R. The uncertainty is due to the inherent uncertainties ($w_n$) of the independent variables ($x_n$). For the present study 3-D kinematic , kinetic and anthropometric data were used from a 64.7 kg male walking at approximately 0.99 m/s and joint moment uncertainties were calculated in three dimensions during one gait cycle (heel-strike to heel-strike) of the right leg. The typical errors associated with the independent variables described above were used in the uncertainty analysis.

**RESULTS**: While three dimensional joint moment uncertainties for the right ankle, knee and hip joints were calculated in this study, only those of the hip are presented here (Figure 1). The results show that as a result of typical errors associated with the collection of gait data the uncertainties of hip joint moments are large (but less than the "worst case" scenario). The average uncertainties across the entire stance phase (time = 0 to 0.8 seconds) are: internal/external rotation - 15.3 Nm; flexion/extension - 18.1 Nm; and abduction/adduction - 16.4 Nm. The independent variables which contribute the most to the uncertainty of the joint moments are marker position, anthropometric data and center of pressure location (Figure 2) while ground reactions, segment mass and moment of inertia, and segment velocity and acceleration contribute minimally to joint moment uncertainty (all less than 0.5 Nm).

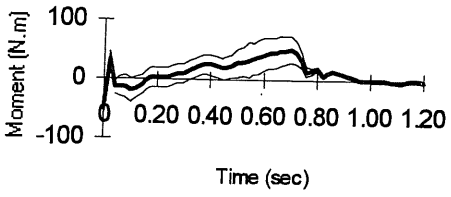



Figure 1. Hip flexion moment. The bold line is the calculated joint moment. The thin lines represent bands of uncertainty.

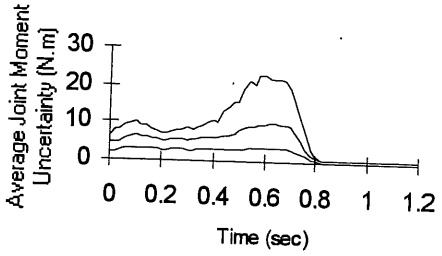



Figure 2. Average of ankle, knee and hip joint uncertainties due to marker position (top), anthropometric data (middle) and center of pressure (bottom).

**DISCUSSION**: The results of this study show that the joint moment uncertainties associated with the inverse dynamics approach are large in comparison to the calculated joint moments values. For example the peak hip flexion moment measured was 40 Nm while the uncertainty was 20 Nm: a 50% error. Not only can the moments be over or underestimated, the moments may also be calculated as being in the wrong direction. This error may be attributed to typical errors in data collection techniques. The question then arises as to how can (and have) investigators find significant results in gait studies when inherent uncertainties are so large. For example, Prodromos et al., 1985, showed that patients who exhibited peak knee adduction moments of less than 4.00 (%body weight x body height) benefited substantially more from a high tibial osteotomy surgical procedure than those patients who had peak knee adduction moments greater than 4.00 %BWxBH. A possible explanation as to how Prodromos et al, found significant differences despite such large uncertainty in joint moments is that they averaged across 12 gait cycles for each subject. By applying equation 1 to the calculation of the average of 12 trials it can be seen that increasing the number of trials will decrease the uncertainty of hip adduction moment (averaged across the stance phase) from 16.4 Nm to 4.7 Nm. In conclusion, this study examined the uncertainties in 3-D joint moments and found that even in the sagittal plane, where moments are large, the errors can be as substantial as fifty percent. These uncertainties need to be considered when performing gait analyses and selecting the number of trials.

**REFERENCES**: 1. Andrews and Mish, J. Biomech 29:651-4, 1996. **2.** Davis, Conference Proceedings of the European Society of Biomechanics 276, 1995. **3.** Prodromos et al., J. Bone and Joint Surg. 67:1188-94, 1985. **4.** Vaughan et al., Dynamics of Human Gait, Human Kinetics Publishers, 1992.

# MOMENT-BASED PARAMETERIZATION OF CYCLOGRAMS OF SLOPE-WALKING

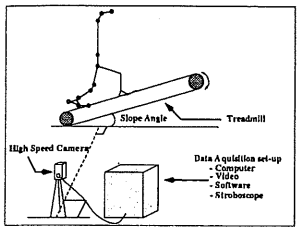
Ambarish Goswami and Emmanuel Cordier
INRIA Rhône-Alpes, Grenoble, France

## 1 MOTIVATION

The problem under study is the moment-based parameterization of a series of gradually evolving cyclograms. The cyclograms correspond to natural human walk on a variable-inclination treadmill with slope varying from $-13°$ to $+13°$ at each $1°$ interval. Parameterization of cyclograms has several useful applications including the diagnosis of pathological gait. For this study we consider the planar hip angle/knee angle cyclograms although the described method is fairly general in nature and may be used without restriction for any closed curve such as the phase diagrams and the moment-angle diagrams. Moreover, the method can be extended to higher dimensions and thus we can treat cyclograms involving three or more joint angles.

Moment-based identification of physical objects is a topic of active research in the field of pattern recognition. We perform the necessary adaptations of these techniques to use them in biomechanics.

## 2 METHODS

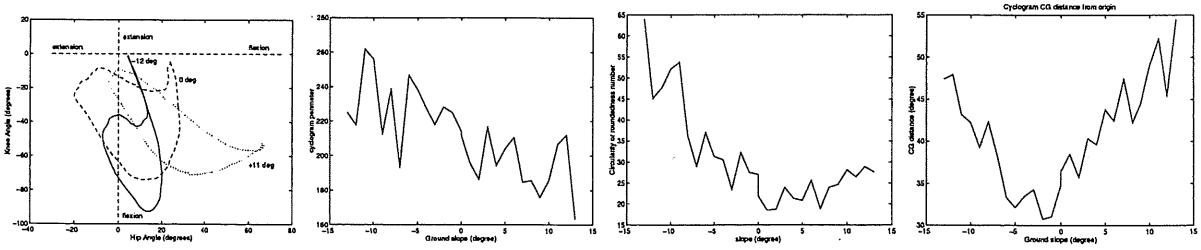



We collected sagittal plane position data of retro-reflective markers attached to the traditionally used bony landmarks of two healthy treadmill-habituated male subjects with a video camera. The treadmill speed was adjusted so that the subjects could walk with their natural gaits. After each run, consisting of several datasets from the same slope, the treadmill inclination was changed without any pre-specified order to prevent any anticipatory gait by the subjects. The marker position data were subsequently smoothed using a Butterworth filter and the inter-segmental angles were extracted from them.

We consider that the closed-loop cyclograms are physically analogous to thin wire loops with uniform mass distribution. The analogy is justified in [1]. The two-dimensional moments of order $(p+q)$ of a curve of length $L$ is expressed as [3], $M_{pq} = \int_0^L x^p y^q dl$, where the curve is represented in the $xy$-plane. Since, in practice, the cyclograms are constructed from discrete data points, the above equations become summations rather than integrals. It can be shown that for the cyclograms the perimeter $P = M_{00}$, the position of the center of gravity $x_{CG} = \frac{M_{10}}{M_{00}}$, $y_{CG} = \frac{M_{01}}{M_{00}}$, and the orientation are obtained from $M_{20}, M_{11}$ and $M_{02}$. Many other parameters can be calculated as well from the moments[1].

## 3 RESULTS

The evolution of several gait features were studied for parameterization. A representative of the results are shown in the following set of figures. The first figure exhibits the substantial change in the forms of the




hip-knee cyclograms. The cyclograms in the figure correspond to $-12°$, level walk, and $+11°$ slopes. The results are similar to those found by [4]. The cyclogram perimeter (in degree) has a clear decreasing trend as can be seen from the second figure. The next figure shows the circularity factor or the roundedness criterion [2] as a function of slope. The fourth figure characterizes the evolution of the distance of the CG of the cyclograms with the ground slope.

## 4 DISCUSSION

This study provides us with a straightforward route to the parameterization of slope walking gait. We may start, for instance, by fitting the above curves with linear, piecewise linear or quadratic functions. Please note that the significant fluctuations in the plots are partly due to the fact that the walk cycles were not extracted with any signal processing technique but were obtained visually from smoothed data. Appropriate segmentation techniques are expected to improve the results.

# References

[1] A. Goswami and E. Cordier. *Moment-based classification of evolving cyclograms of human walk on gradually changing slopes.* INRIA Research Report, (to be published), 1997.

[2] C. Hershler and M. Milner. Angle-angle diagrams in the assessment of locomotion. *American Journal of Physical Medicine*, 59(3):109–125, 1980.

[3] X. Jiang and H. Bunke. Simple and fast computation of moments. *Pattern Recognition*, 24(8):801–806, 1991.

[4] J. Wall and J. Charteris. The temporal and angular kinematics of uphill and downhill walking. *unpublished.*

# PENDULAR EFFICIENCY AT VARIOUS SPEEDS OF WALKING IN ELDERLY WOMEN

M. Kaneko, H. Fujiwara, and T. Fuchimoto
Osaka University of Health and Sport Sciences, Osaka, Japan

## INTRODUCTION

Locomotion is of particular importance in elderly people to maintain their voluntary and active social interactions in daily life. In the elderly population, the speed of walking significantly correlates with various aspects of physical fitness (Kaneko et al, 1991). From the perspective of energetics, Cavagna et al (1976) explained the mechanism of human walking using a pendulum model, in which the potential energy at the body's center of mass (*Ep*) is transformed into kinetic energy (*Ek*) and vice versa. It was also shown that the most efficient walking speed increases with age during the growth period from 2 to 12 years old (Cavagna et al, 1983). The present study examined0 the pendular motion characteristics of walking at various speeds in elderly women.

## METHOD

The subjects were 37 elderly women consisting of two age groups; G70 aged 65-74 years (n=19) and G80 aged 75-85 years (n=18). Twenty-one younger women aged 18-25 years (G20) were compared with the two elderly groups. The subjects walked on a 10m wooden walkway where 3 force plates (KISTLER) were implanted midway to record the ground reaction forces. A newly designed electrical pacemaker (CAT EYE) was also set along the walkway at a 0.9 m high, with which the subject's walking pace was directed by a moving light-band. Walking was performed at 7 different speeds, consisting of 5 fixed speeds (40, 55, 70, 80, and 100 m/min), a self-selected speed and the maximal speed. In addition to kinematic data collected by two video cameras, the mechanical energy was computed from the force recordings. The work to accelerate the body forward due to *Ek* changes (*Wf*) and the work to lift the body due to Ep changes (*Wv*) were calculated from vertical and forward ground reaction forces, respectively. The external work (*Wext*) was obtained by adding increments in the curves of total energy (*Etot=Ek + Ep*) . The pendular efficiency (*PEF;* called '% recovery' by Cavagna) was also calculated from the equation below. Note that *PEF* is 100% when Wext is zero.

$$PEF\,(\%) = [(Wf + Wv - Wext) / (Wf + Wv)] \times 100$$

## RESULTS

The external power (*Wext*) attributable to muscular activity increased exponentially with increasing speed (r= 0.984 - 0.997). As seen in Fig.1, *Wext* of G70 (solid line) tended to be less than that of G20 (dotted line). *PEF* (lower panel in Fig.1) changed with speed in a concave fashion, exhibiting a maximum at an intermediate speed. The maximal *PEF* values of the elderly were not different from tose in the younger group of G20 (63-64%). However, the walking speeds at which *PEF* reaches a maximum ('optimum speed'; *V*-opt) were found to be lower in the elderly groups; 74 m/min in both G70 and G80, whereas 77 m/min in G20 (see thin arrow). The *V*-opt relative to the maximal speed (% *V*-max) were higher in the elderly (70% in G70, 77% in G80) than in G20 (61%), because of lower *V*-max in the elderly. Interestingly, these *V*-opt appeared to be somewhat lower but similar to the self-selected normal walking speed indicated by the thick arrow. As the walking speed shifted away from *V*-opt, *PEF* decreased more drastically in G70 and G80 than in G20, suggesting that the efficient walking speed range is much narrower in the elderly than in the younger subjects. At the speed above *V*-opt, subjects in G70 and G80 walked with shorter step length and at higher step rate than those in G20. At speeds below *V*-opt, however, the elderly tended to walk with longer step length and at lower step rate on the contrary. These characteristics of step length and frequency in elderly subjects may account for the reduction in pendular efficiency at speeds other than the optimum.

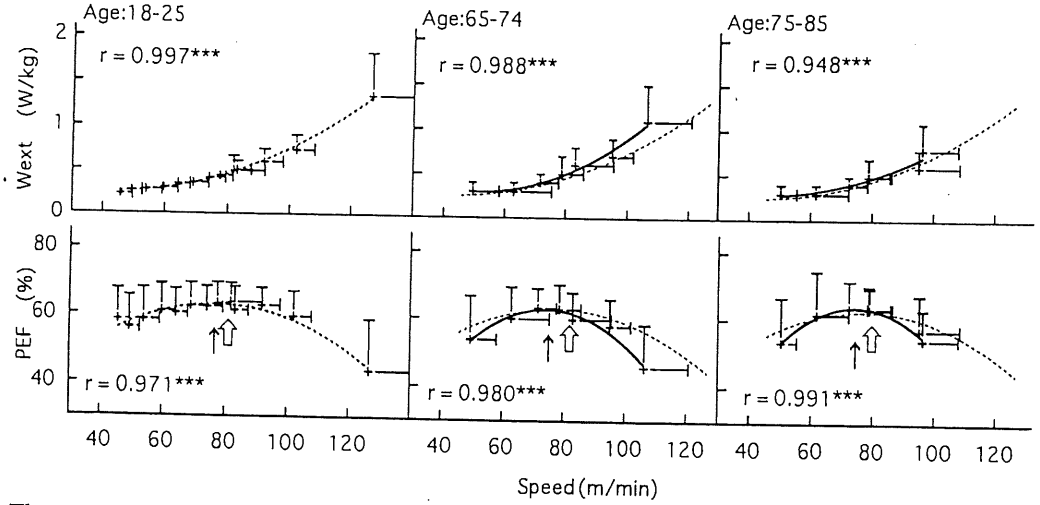



Fig. 1. The external power (*Wext*) and pendular efficiency (*PEF*) in relation to walking speed. The thin and thick arrows indicate the optimum and the normal walking speed, respectively.

# PLANTAR PRESSURE AND DYNAMIC SIMILARITY
## (Spontaneously chosen walking velocity *vs* normalised velocity)

Pierre Moretto ◊; Patrick Pelayo ◊ And Mario A. Lafortune ‡
◊ Faculté des Sciences du Sport et de l'Education Physique, Lille II, France
‡ Nike Sport Research Laboratory, Beaverton, Oregon USA.

**Introduction:**

Dimensionless analysis of gait pattern has been used to evaluate patient (Wagenaar and Beek, 1992) and seems to be of interest for gait analysis (Zatsiorsky et al., 1994). However, the application of this method is linked to the observation of pattern similarities between different subjects. Many authors consider that this similarity can be observed during walking at spontaneously chosen velocity (SCV) (Alexander, 1989). Nevertheless other studies showed that SCV may depended on both the context of the study and the physical fitness level of the subjects (Yamasaki et sato, 1990; Sato et al., 1991). The aim of this study was to compare space and time parameters and in-shoe plantar pressures during walking at spontaneously chosen vs normalised velocity.

**Procedures**

Fifteen male subjects (mean age: 24 years; height: 1.73 m; leg length: 0.91 m; body mass: 71.37 kg) performed 2 tests where the walking velocity was imposed by a bank of lights. In the first test ($T_{SCV}$), the spontaneously chosen velocity (SCV) corresponded to a fraction of Froude number (Nfr) equal to 0.41. In the second test ($T_{NV}$), the velocity (NV) was normalised and determined for each subject from the leg length (L) and the same Nfr fraction ($NV=0.41(g.L)^{1/2}$). Step Frequency (SF) was recorded by Parotec-System (P-S by Kraemer, Germany) and the Step Length (SL) was calculated from the ratio of the velocity and SF. The heel and forefoot peak pressure were recorded using the Parotec-System and computed for two groups of 4 and 8 transducers, respectively. Intrasubject coefficient of variability (CoVi) were calculated over 16 steps in each test. Mean, standard deviation and coefficient of variability (CoV) were determined and the differences between the tests have been studied. In order to verify the proportionality of the different parameters across subjects, the different ratio were computed for all pair of subjects. Anthropometric characteristics, space and time parameters, and plantar pressure ratio were computed as follows: L1/L2, $(M1/M2)^{1/3}$, $(SF2/SF1)^2$, P1/P2 with L (leg length); M (Body Mass); SF (step Frequency); P (plantar peak pressure) of two subjects (1 and 2). A two factors ANOVA was conducted to compare the ratio between the two tests ($P<0.05$). A Student "t" test was used to compare all mean values recorded during the two tests ($P<0.05$).

**Results:**

Velocity (1.21±0,15 vs 1.24±0.03 $m.s^{-1}$), SL (0.68±0.04 vs 0.66±0.10 m), SF (1.84±0.07 vs 1.81±0.10 $step.s^{-1}$) and FFPP (796.7±141.9 vs 801.23±107 kPa) mean values were not significantly different between $T_{SCV}$ and $T_{NV}$. HPP were 18% lower at $T_{NV}$ (357.8±88.23 kPa) than at $T_{SCV}$ (422.34±97 kPa). HPP and FFPP CoVi were lower in $T_{SCV}$ than in $T_{NV}$. No significant difference has been noticed for SF and SL CoVi. CoV is never significantly affected by the conditions of the tests. In $T_{SCV}$, the analysis of the ratios showed that SF and FFPP are proportional to body mass. Nevertheless, SL and HPP were not proportional to anthropometric parameters. In $T_{NV}$, all the parameters, except the HPP, were proportional to body mass .

**Discussion and conclusion:**

The results suggest that walking velocity proportional to the length of limb should be determined and imposed in order to observed proportional gait patterns. NV could be imposed in order to determine plantar pressure mean value references for rehabilitation. Indeed, HPP mean values were overestimated by 18% at SCV and DeVita and Bates (1988) showed that a 5% difference can be biomechanically meaningful.

However, CoVi is lower at spontaneously chosen walking speed. Plantar pressure CoVi is linked to balance (Hennig and Rosenbaum, 1991) and suggests that SCV should be used to study equilibration process.

**References:**

Alexander R. McN.. Physiol. Rev., 69, 1199-1227, 1989.
DeVita P. and Bates B.T.. Hum. Mov. Sci., 7, 73-85, 1988.
Hennig E.M. and Rosenbaum D.. Foot & Ankle, 11(5), 306-311, 1991.
Sato H. , Sako H., Mukae H., Sato A. and Takahashi T.. J. Hum. Ergol. (Tokyo), 20, 85-88, 1991.
Wagenaar R.C. and Beek W.J.. J. Biomech., 25, 1007-1015, 1992.
Yamasaki M. and Sato H.., J. Anthrop. Soc. Nippon, 98, 385-400, 1990.
Zatsiorsky V.M., Werner S.L. and Kaimin M.A.., J. Sp. Med. & Phys. Fit., 34(2), 109-134, (1994).

**Acknowledgement:**

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# THREE-DIMENSIONAL KINEMATICS OF NORMAL WALKING GAIT USING A FIVE SEGMENT MODEL

U. Rattanaprasert, R. Smith, W. Gilleard
Faculty of Health Sciences, The University of Sydney, AUSTRALIA

## INTRODUCTION

It is thought that there is a cause and effect relationship between abnormal motion of the foot and injury or pain symptoms found in more cephalad joints and segments. However, these relationships have rarely been investigated objectively. Kadaba et al. (1990)[1] proposed a marker system for three-dimensional (3D) analysis of the hip, knee and ankle joints and pelvis of normal gait but examined only two degrees of freedom for the angular motion of the foot. The simultaneous measurement of the 3D motion of the forefoot, rearfoot, leg, thigh, pelvis, and trunk during normal gait would provide useful additional information in this area as well as in the area of coordination through the calculation of cross-correlations. The objective of this study was to quantify the 3D, three angular degrees of freedom kinematics of the foot, leg, thigh, pelvis, and trunk in normal walking.

## METHODS

Ten healthy subjects (male, n=5, female, n=5) with 29 retro-reflective markers fitted on their right lower limbs and torso, performed five successful trials of walking at a self-selected speed on a 10-m walkway. An eight-camera Expert Vision Advanced (EVa) Motion Analysis System™ was used to record 3D movements of the segments and synchronised force platform variables at 60 Hz. The 3D trajectories of the markers were measured and calculated by the EVa Motion Analysis System™. Kinematic and force platform records were used to determine the events of the walking cycle. Relative angles between the adjacent segments were then computed according to the International Society of Biomechanics conventions[2]. The three degrees of freedom of the angular excursions of rearfoot-forefoot, leg-rearfoot, thigh-leg, pelvis-thigh, and trunk-pelvis for each subject were time-normalised and averaged. The mean and 95% Confidence Interval (CI) for ten subjects were then calculated.

## RESULTS AND DISCUSSION

The measurements were reliable as the 95% CI of each angular motion was relatively small when compared with the range of motion of the joint. Further such comparisons of the 95% CI of each relative angular motion led to two more observations. The flexion/extension motion at each joint between the two segments, excepted trunk-pelvis, was less variable between subjects than it was for ad/abduction and internal/external rotation. The variability within subjects was small when compared with the differences between subjects. The subjects showed the same patterns of motion but different ranges of motion. For the mean and standard deviation of the range of motions for each joint, the rearfoot-forefoot dorsi/plantarflexion (df/pf) was 9.52°±1.67°, inversion/eversion (inv/ev) 5.46°±3.5° and ad/abduction (add/abd) 11.79°±4.51°; the leg-rearfoot df/pf was 18.52°±0.80°, inv/ev 11.38°±3.52° and add/abd 7.22°±4.54°; the thigh-leg flexion/extension (fx/ex) was 54.04°±0.99°, add/abd 10.01°±0.74° and internal/external (int/ext) rotation 8.61°±1.29°; the pelvis-thigh fx/ex was 42.76°±4.01°, add/abd 11.39°±2.23° and int/ext rotation 11.11°±1.79°; the trunk-pelvis fx/ex was 2.82°±4.29°, left/right (lt/rt) lateral flexion 11.80°±2.38° and rt/lt axial rotation 14.24°±1.56°.

The results which have been obtained simultaneously in this study were similar to movement patterns described in a range of other studies[1,3-5]. The 3D forefoot motion relative to the rearfoot demonstrated some functional aspects of the foot. During stance phase, the dorsiflexion of the forefoot was consistent with the bending of a beam under stress. It is in this period that the ankle moment is generated. The forefoot plantarflexed relative to the rearfoot immediately prior to toe-off, a period when large plantarflexion moments would be required. It is theorised that the dorsiflexion of the hallux shortens the plantar fascia under the arch of the foot at this time to provide the necessary tension (the windlass effect). The results of this study provide a database of 3D, three angular degrees of freedom kinematics of the foot, leg, thigh, pelvis, and trunk during normal gait for further research in investigating how the pathological foot affects the rest of the body during walking.

## REFERENCES

1. Kadaba, M.P., Ramakrishnan, H.K. and Wootten, M.E. (1990). Measurement of lower extremity kinematics during level walking. *J. Orthop. Res.* **8**, 1-10.
2. Wu, G. and Cavanagh, P.R. (1995). ISB Recommendations for standardization in the reporting of kinematic data. *J. Biomechanics* **28**, 1257-1261.
3. Crosbie, J., Vachalathiti, V. and Smith, R. (1996). Age, gender and speed effects on spinal kinematics during walking. *Gait Posture* In Press.
4. Sutherland, D.H., Olshen, R., Cooper, L., & Woo, S.L-Y. (1980). The development of mature gait. *J. Bone Jt Surg.* **62**, 336-353.
5. Winter, D.A. (1983). Biomechanical patterns in normal walking. *J. Mot. Behav.* **15**, 302-330.

# A DETAILED ANALYSIS OF MOTION OCCURRING IN THE TALOCALCANEAL JOINT DURING NORMAL WALKING

P. Westblad, A. Lundberg, I. G. Winson, K. Halvorsen, T. Hashimoto.
From the Dept of Orthopaedics, Karolinska Institute at Huddinge Hospital, Stockholm, Sweden.

We have undertaken a detailed analysis of motion occuring in the subtalar (talocalcaneal joint) during walking using a skeletally fixed markers and an optoelectronic system of kinematic observation. We present detailed results indicating the motion utilised during normal walking and the alingment of the axis of motion during the various stages of the stance phase of gait.

## MATERIALS AND METHODS

Three healthy normal volunteers were studied. Using local anaesthetic infiltration 2.4mm Hoffman external fixateur pins were inserted into the tibia, the fibula, the talus and the calcaneum. Their rigidity of fixation was assessed manually and the pins placement checked radiographically. Each pin then had a total of four 10 mm reflecting markers attached. Great care was taken to ensure that each pin's markers did not interfere with other markers, and pin placement was critically controlled to prevent interferrence with anatomical structures which would restrict joint motion. None of the subjects experienced any discomfort during walking and there was no visible limp. A Kistler force plate was used to compare ground reaction force patterns while walking with pins inserted, to step cycles collected with the same subjects before pin insertion. Steps with grossly different force distribution were not used in the analysis.

Studies were then undertaken in standing and walking using a three camera optoelectronic system (MacReflex) of recording motion. Subsequent analysis of data made use of kinematic computer software to show the amount of motion, the direction of motion and the axis around which motion occured.

## RESULTS

In each of the subjects the amount of motion utilised in the talocalcaneal joint was low (Figure). Maximum motion into eversion, compared with a standing neutral as reference, occurred from heel strike to foot flat. Maximum motion into inversion occurred from loss of heel contact to toe off. No clear pattern of external internal rotation could be observed (Figure).

## CONCLUSION

The presented method can be used to accurately analyse the small amounts of motion which occur in individual joints of the ankle complex during normal walking.

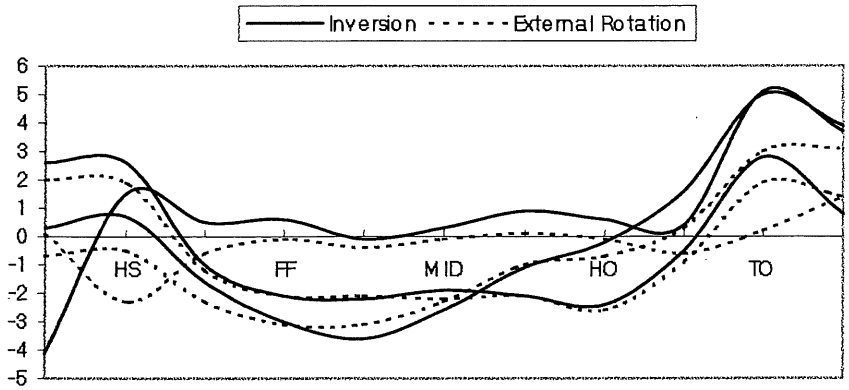



Figure. Curves showing inversion (+) and eversion (-) (solid lines) in the talocalcaneal joint in three subjects during normal walking. Internal (+) and external (-) rotation is indicated by dashed lines.

# A NEW QUANTITATIVE APPROACH FOR ARCHERY STABILITY ANALYSIS

Tzyy-Yuang Shiang, Cheng-Jen Tseng
National College of Physical Education and Sports, Taiwan

## INTRODUCTION

The aiming techniques of archery are different from that of shooting. At the present, experience and feel are the most frequently used terminology in archery coaching. A more scientific coaching approach is needed to enhance archery performance. Stability has been determined to be the most important factor in the aiming of the target. In the previous studies [Manson, 1986, Wong, 1990], the stability was determined by the movement of the locus of the center of mass projected on the ground which is caused by body sway. Subsequent studies show that the aiming stability has multiple contributing factors such as the upper body muscle stability, psychological factors, muscle fatigue and external forces in addition to body sway. [Baker, 1994, Tseng, 1996] Of the various contributing factors, the upper body muscle stability is determined to be a major factor. In this study, the authors employed a laser pointer on the bow to quantify the aiming stability in order to enhance archery performance.

## METHODS

The subjects for this study were 10 male experienced archers, with a mean age of 20 years, and with a mean archery history of 4.8 years. All subjects have been elected as members of National Archery Team during the test. The best performance of each subject score 1200 at least. Each subject was asked to shoot 12 arrows during the data collection. Body sway stability was measured using force plate. Briefly, a subject stands on the force plate when he pull the bow and string during aiming process, the force plate records the path of center of mass projected on the ground in the 2 seconds prior to arrow release. The center of the projected locus was calculated and the mean radius of the area which is the mean distance between each point and the center was then computed. This mean radius was used to determine the stability of the body sway. A laser pointer was employed on the bow, the laser spot was point on a separate target during the aiming process, and the path of the laser spot can be recorded using high speed video camera (60 frames/sec) to represent the movement locus of the bow. The mean radius of laser spot area was calculated using the motion analysis system (PEAK Performance System) according to the same approach which determine the body sway stability.

## RESULTS

The mean radius of the aiming locus, the mean radius of the body sway trajectory and archery performance (score) for 10 subjects are shown in Table 1. And linear regression analysis was used to determine the relationship among the aiming stability, body sway stability and archery performance, Table 1 also illustrates Pearson Correlation coefficient of archery performance and both stability radius. In order to compare different pattern of aiming locus, Figure 1 demonstrates 2 typical aiming locus patterns.

Table 1: The mean score, radius of aiming locus and body sway locus, and Pearson correlation coefficients (★ significant, $p < 0.05$)

| Subject | Mean Score | Radius A (cm) | Radius B (cm) | Correlate A vs S | Correlate B vs S |
|---|---|---|---|---|---|
| 1 | 9.25 | 23.68 | 0.43 | -0.79 ★ | -0.47 |
| 2 | 9.25 | 7.64 | 0.67 | -0.70 ★ | -0.06 |
| 3 | 9.00 | 9.11 | 0.65 | -0.34 | 0.54 |
| 4 | 9.41 | 5.66 | 0.27 | -0.69 ★ | 0.18 |
| 5 | 9.16 | 5.66 | 0.86 | -0.79 ★ | 0.15 |
| 6 | 9.08 | 5.87 | 0.79 | -0.61 ★ | -0.44 |
| 7 | 9.67 | 7.71 | 0.48 | -0.64 ★ | 0.14 |
| 8 | 8.33 | 14.69 | 0.61 | -0.67 ★ | -0.25 |
| 9 | 9.50 | 10.58 | 0.67 | -0.64 ★ | -0.01 |
| 10 | 9.41 | 7.07 | 0.43 | -0.73 ★ | 0.34 |
| Total | 9.20 | 11.33 | 0.63 | -0.31 ★ | -0.03 |

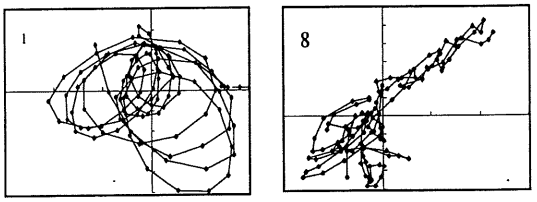



Figure 1: The aiming locus patterns of subject 1 and 8

## DISCUSSION

The correlation between aiming stability and performance for 90% of subjects are significant, and none of the subjects has significant correlation between performance and body sway. This result seems to indicate that the aiming stability is the key factor affect the archery performance. In addition to the size of aiming locus, the aiming locus pattern is also a useful index to determine the performance. Interestingly, the aiming locus area of subject 1 is larger than subject 8, but the performance of subject 1 is better. This discrepancy may be due to problems related to locus pattern which is demonstrated in Figure 1. Subject 1 tends to have a circular movement locus and converge to the center of target, and subject 8 moves the aiming point from one area to another which may miss the target. Better archers keep the patterns more consistent to increase stability. As a conclusion this new quantitative approach can be applied to other activities involving stability.

## REFERENCE

Baker,J., Mclean,B. (1994) Bow movement in archery. *Pelops* (Melbourne-Aust.) 9, 17-21.

Manson,B.R., Pelgrim,P.P. (1986) Body stability and performance in archery. *Excel*(Canberra-Aust.)3(2), 17-20.

Tseng,C.J., Shiang,T.Y. (1996) The laser application for testing archery stability. *1996 International Sports Training Science Symposium.* 146-156.

Wong,C.F., Tsen,T.K. (1990) The analysis on the instantaneous stability of archers: a synopsis. *Asian J. of Physical Education.* 13(1)56-57.

# COMPUTER SIMULATION IN BIOMECHANICAL TESTING OF TENNIS RACKETS

U. Glitsch, C. Detlefs, W. Baumann
Institute of Biomechanics, German Sport University Cologne, Germany

## INTRODUCTION

The contact phase during a tennis stroke which is very complex and short is the most important incident in tennis playing. Many factors such as mechanical properties of the ball, strings, racket frame, grip firmness, and the mechanical properties of the player's arm are influencing the rebound of the ball. These interactions are yet not sufficiently understood. In this context computer simulation may assist with the improvement in understanding of the impact phase in the tennis stroke. The aim of this study was to investigate the geometrical mass distribution of different tennis rackets and its influence on the impact using computer simulation methods.

## METHODS

A planar computer model used in this study was based on a multiple pendulum consisting of upper and lower arm, hand and racket as described by Casolo and Ruggieri (1991). The arm segments were linked by hinge joints and the racket was firmly clamped to the hand. The preimpact angular velocity of the arm was set to $\omega_0$ =17.45 rad/s which is approximately equivalent to a velocity of 19 m/s of the center of the string area. The ball's trajectory was adjusted to hit the racket perpendicular to the string surface with a velocity of 20 m/s. The geometrical mass distributions of 7 modern rackets (less than 3 years old) and 2 old rackets (more than 10 years old) were determined experimentally and served as input variables of the mathematical model. The visco-elastic properties of the racket, strings and ball were combined in an overall spring-damper system (D= 18 kN/m; R=10 N·s/m) according to Leigh and Lu (1992). All the model setup and calculations of the impact kinetics were performed with the DADS software package (CADSI Inc.).

## RESULTS AND DISCUSSION

It was not very surprising that the masses of the investigated tennis rackets varied considerably from 0.272 to 0.376 kg, and that the older rackets had 0.06 kg more mass on average. The locations of the center of gravity showed no definite tendency of the modern rackets, they ranged from 47.4 to 56.0 % of the racket length. When hitting the ball in the head center rebound velocities of the ball from 29.4 to 31.0 m/s were found. Surprisingly, one of the old ones achieved this maximum. But on the other hand, nearly the same rebound velocity of 29.7 m/s was achieved by two rackets with more than a 0.1 kg difference in mass. Even in the group of modern rackets, differences of 0.07 kg could be found and no significant difference in rebound velocity occurred. Increased rebound velocities of 3 % on average were calculated in off-center impact points of 5 cm towards the handle. The maximum impact forces at the strings reached about 800 N whereas with respect to the center of percussion -60 to 110 N were applied at the grip. Hitting the rackets at the border of the string area increased the impact forced up to 350 N. Generally, the modern rackets had the center of percussion nearer to the string center or even sometimes beyond this point. This caused a general reduction of the impact forces at the grip.

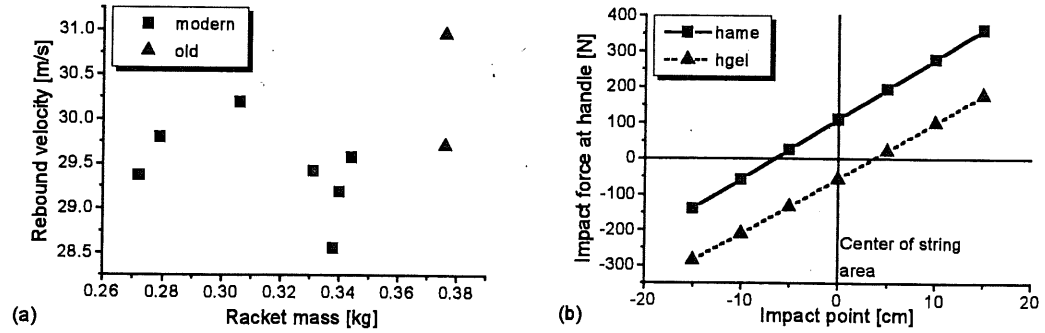



Fig.1: (a) Rebound velocity of the ball vs. racket mass; (b) Impact force at the handle vs. impact point on the longitudinal axis of the string area of two rackets.

## CONCLUSIONS

Basically, the investigated modern rackets showed a slight tendency to less mass, without reducing the rebound velocity. It is supposed that there still exist remarkable potentialities for further optimization of the mass distribution. That means not only the maximization of the rebound velocity, but also the reduction of impact loads on the player's arm.

## REFERENCES

Casolo, F.; Ruggieri, G. (1991): Dynamic Analysis of the ball-racket impact in the game of tennis. Instituto degli Azionamenti Meccanici, Politecnico di Milano, Meccanica **26**, 67-72

Leigh, D. C.; Lu, W.-Y. (1992): Dynamics of the interactions between ball, string and racket in tennis. Int. J. of Sport Biomech. **8**, 181-206

# ABOUT THE DYNAMIC BEHAVIOUR OF THE SKI-BINDING-BOOT-SYSTEM DURING TURNS IN ALPINE SKIING FOCUSED ON THE USE OF BINDING PLATES

Niessen Wolfgang, Müller Erich, Raschner Christian, Schwameder Hermann
Institute for Sport Science, University Salzburg, Austria

The influence of binding plates on the ski-binding-boot-system is actually discussed in alpine skiing. Objects of the discussions are the change of the lever arms as well as the damping charactaristics of binding plates. The influence of binding plates on the dynamic behaviour of the whole ski-binding-boot-system should be investigated in this paper. In order to describe the dynamic behaviour the influence of binding plates on the natural frequencies of the ski and on the transmission track of the shorttime acting forces from the ski to the skier had to be determined.

Three different ski-binding-boot-systems were compared with each other: system A without a binding plate, a 203 cm long Atomic ARC RS ski and an ESS VAR 614 binding, system B with an ESS Powerflex plate and the system C with an ESS Superpower plate. The systems B and C had the same ski and the same binding as system A. The height between the running surface of the ski and the bottom of the boot was 45 mm for system A, 55mm for system B and 65 mm for system C. A highly skilled alpine skier had to cover a standardizised course in which turns causing clattering were provoked. He had to perform three runs with each of the three systems. The ski boot (*Koflach*) was prepared with two three-dimensional acceleration sensors (*ERNST*, range ± 50g) at the shaft and the sole of the boot recording the shorttime acting forces (see fig. 1). At this locations it is possible to determine the influence of the analyzed plates on damping as well as the influence on the range of rotation regarding the longitudinal axes of the ski. Eight one-dimensional acceleration sensors were distributed along the ski. This made it possible to identify natural frequencies of the ski being excited during skiing. The natural frequencies were additionally checked in the laboratory.

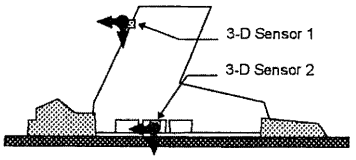


Fig. 1: Position of the sensors

The whole raw data of the acceleration values, which was measured at the relevant turns, was taken through FFT analysis. The dominant clattering frequency was always easy to ascertain at the acceleration sensors being located at the boot. The range was between 25 and 28 Hz. The next step was to quantify the $C_k$ coefficients, which were calculated at the clattering frequency, and to draw comparisons. The mean values of the Fourier coefficients at the clattering frequency for the different systems analyzed are shown in Fig. 2 (Ch 2 for sole lateral, Ch 3 for sole normal - always reffering to the ski CS, Ch 5 for shaft lateral and Ch 8 for shaft normal). Using binding plates the acceleration values at the shaft of the boot become obviously smaller in lateral and normal direction. This clear reduction is also recognizable at the values of the sole in lateral direction. It can be said that both plates reduce the lateral accelerations at turns with clattering parts. In normal direction of the sole it is conspicuous that only for the *Powerflex plate* slightly reduced acceleration values can be determined. As proved in earlier research (NIESSEN) the choice of the screw locations for the fixation of the binding has a great influence on the natural frequencies of the ski. In this project the screw locations were identical respectively similar. The differences between the three systems used lie mainly in the different damping characteristics of the materials of the binding plates. After low pass filtering frequencies about 30 Hz could be determined in some runs. The systems had no significant differences in the event frequency.

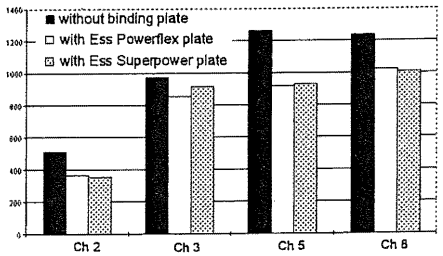


Fig. 2: mean values of the $C_k$ coefficients

It seems that the shorttime acting forces decrease at turns with clattering parts when binding plates are used. If the measuring channels were examined one by one, a higher decrease was found at the lateral acting shorttime forces than at the forces acting normal to the ski. Hence the material dampening of the mounted plates plays only a small part. This phenomenon can however be explained by the increase of the moment of inertia and by the shrinkage of the edge angle of the ski depending on increasing standing position. In contrast with considerations made earlier about the lengthening of the shank and the combined increase of the lever arms there will now be made considerations about the effect of higher standing positions on edge increase. It can be shown that smaller edge angles are necessary when skiing with binding plates to reach the force balance during edge increasing and turning. Hence it cannot be supposed that the really acting forces during turning increase because of the lengthening of the shank, as the change of the acting moments has also to be taken into consideration. Due to the decreased edging angles the system with binding plate has shorter lever arms from the knee to the resultant force vector in lateral and normal direction referring to the ski. In this case the shorttime acting forces probably produce reduced moments at the knee of the skier and decrease the shorttime balancing movements of the knee during clattering phases.

The event frequency of determined natural frequencies did not favour one of the three systems, although two of them had mounted plates with damping materials between the ski and the binding. This strengthens the thesis that the screw locations have a greater influence on the natural frequencies than the materials used for binding plates.

The authors wish to thank particularly the ATOMIC AUSTRIA GmbH for their support.

# BIOMECHANICAL ESTIMATE OF MOTION KINETICS UNDER VARYING MOTION INTENSITY IN WEIGHTLESSNESS

A.V.Zinkovsky, V.A.Sholukha, D.G.Arsenjev, A.A.Ivanov
St. Petersburg State Technical University, St. Petersburg, Russia

## INTRODUCTION

Motion control structure and energy expenditures analysis for human locomotions under extreme conditions is seriously obstructed from experimental biomechanics point of view. This analysis is traditionally dealt with by means of mathematical and more exactly computer modelling. Distinguishing of general laws in motion control organization in weightlessness and varying motion intensity requires employment of developed computer model which would allow to carry out synthesis of goal-oriented (with respect to kinematics and kinetics) motions (Zinkovsky et al., 1996).

## METHODS

Assessment of motion intensity in weightlessness is carried out on computer model of human skeletal-muscular apparatus (SMA). The computer model allows structural and parametrical adjustment with respect to concrete individual motion. Motion synthesis via employment of kinematic and dynamic non-stationary constraints allows to vary model elements and joints structure preserving meanwhile (or presetting desired) values of integral characteristics of motion. In particular, in this investigation, alongside with variation of gravity force acceleration $G$, we varied amplitude and form of support reaction impulse preserving however kinetic moment value and/or desired shoulder joint kinematics was preset.

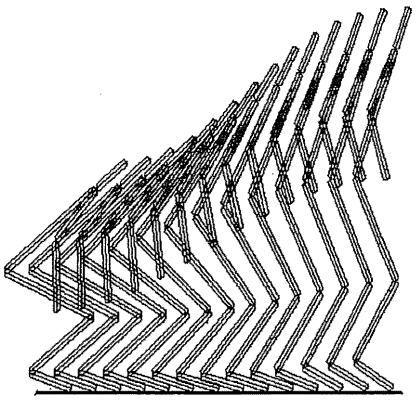


Fig.1. Kinematic scheme of the 7-link model motion.

## RESULTS

In Figs. 1-3 there are presented kinematic scheme of motion, calculated power and vertical components of joint reactions functions for start-stop motion. The motion was performed for 1 and 2 seconds successively under zero gravity ($G$ =0). Feet toes were fixed. Analysis of joint powers (Fig. 2) shows necessity of inter-element moments sign switch. Increase of motion velocity by two times leads on the average to four and eight times higher energy losses and maximum power values correspondingly. For different joints resulting increase can vary from two to ten times. There has also been conducted assessment of necessary intensity of motion under zero-gravity, providing on the average equivalent energy losses in comparison with normal $G$ value.

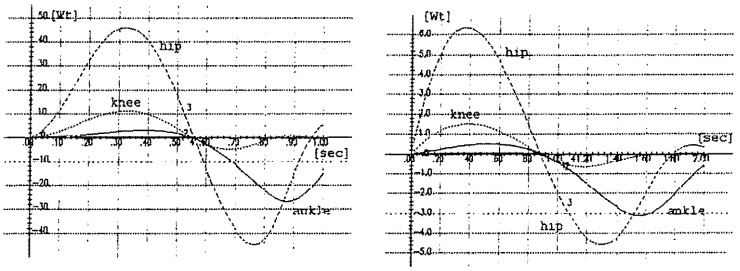



Fig.2. Power distribution over joints.

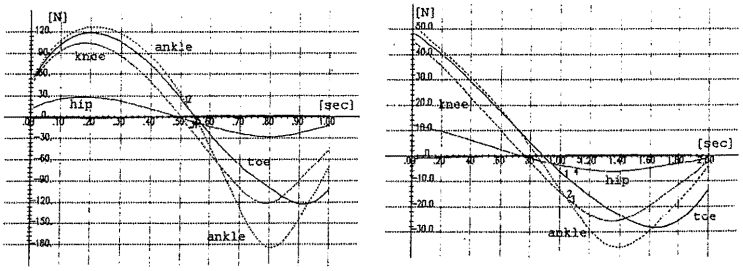



Fig.3. Joints vertical reactions components

## DISCUSSION

Conducted assessment of control structure and energy losses under zero-gravity showed principal changes in forces and moments distribution in SMA. However, from the point of view of biomechanics there can be found motion characteristics providing certain equivalency of motion under zero-gravity to that in normal conditions. This knowledge can be used for construction of special rehabilitation system.

## REFERENCES

Zinkovsky, A.V., Sholuha, V.A., Ivanov, A.A. (1996) Mathematical Modelling and Computer Simulation of Biomechanical Systems. World Sci. Publ. Co. Ltd., London, UK, 212 pp.

# HUMAN MOTOR STRATEGIES EVALUATION IN LONG TERM mGRAVITY EXPOSURE

G. Baroni, G. Ferrigno, G. Andreoni and A. Pedotti
Dipartimento di Bioingegneria, Politecnico di Milano, Milan, Italy
Centro di Bioingegneria, Fnd. Pro Juventute Don Gnocchi, Politecnico di Milano, Milan, Italy

## INTRODUCTION

European Space Agency (ESA) Mission EUROMIR'95 has implied the permanence of an European astronaut and a Russian cosmonaut on board of the Russian space station MIR for 180 days. (3rd September 1995-29th February 1996). Among the in-flight experiments, the technological experiment T4 has been designed in order to evaluate the chance of successful exploitation of an opto-electronic motion analysis technology on board of an orbital module. For this aim, a related scientific program of human movement analysis in weightlessness has been carried out on both subjects. Beside the sessions on ground (Baseline Data Collection pre-flight and post-flight) performed few days before the launch and after landing, 8 sessions in flight have been realised on flight days (FD) 11, FD 19, FD 22, FD 61, FD 69, FD 74, FD 113, FD 150. Among the ten T4 experimental protocol, we focus on the three neurological related voluntary posture perturbation experiment. The investigation has been performed in order to verify and broaden well known results obtained on parabolic flights and short-term mgravity exposures, enriching the analysis with the evaluation of long term motor adaptation process to weightlessness.

## METHOD

Voluntary posture perturbation protocols have been : Axial Movements (AM), Leg Raising (LR) and Head-trunk Co-ordination (HT). A full body marker disposition model has been defined, considering traditionally reliable and easily identifiable anatomical landmarks (temple, zigom, acromion, anterior superior iliac spine, great trochanter, femoral epicondyles, medial and lateral malleolus). Analysis primary output consisted of markers three-dimensional co-ordinates, after on-ground post-processing, identified with an accuracy which has been estimated in 1/1000 of the maximal dimension of the calibrated volume (2000 mm)[1]. Three-dimensional marker co-ordinates have been exploited for protocol related analysis, aiming to the quantification of movement parameters for the evaluation of subject's motor strategies. These processing has included angular analysis (AA) displacement (D) and displacement ratio (DR), roll dispersion (RD) and anchoring index (AI) computation and the principal component analysis (PC) [2].

## RESULTS

Axial movements : task is, starting from erect posture, to bend as fast as possible the trunk 30 degrees forward, to recover the erect position and then to bend the trunk 30 degrees backwards. Evaluation on erect posture has been performed by means of AA, showing that subject A was able to assume an erect posture similar to the one on Earth not before the 11th FD. D and DR and PC analysis confirm that although equilibrium constraint no longer exists in mgravity, prime movement is always accompanied by feed-forward postural adjustments, appearing to be influenced by a unique central control mechanism[2].Leg Raising : subject elevates one leg up to 45 degrees laterally, keep position for few seconds, and recover the initial position. D and AA analysis have allowed to point out significant changes both in anticipatory postural adjustment and in the global LR task execution. Consideration about the role that biomechanical constraints and variables play in the evaluated motor behaviour will be provided[2].Head-Trunk Co-ordination : subject performs rhythmical 15 degrees trunk oscillations on the frontal plane with the explicit task of maintaining the infra-orbital line aligned with the perceived horizontal direction. RD and AI analysis has pointed out that subject are able to actively stabilise the head even after a long period of weightlessness exposure. This results seem to confirm the more significant task related role of active vestibular inputs rather than of static vestibular or proprioceptive information[2].

## AKNOWDLEGEMENT

The author wish to thank the Italian Space Agency for giving them the chance of flying the space qualified motion analyser and of collecting the data presented in this communication.

## REFERENCES

1. Baroni, G., Ferrigno, G., Andreoni, G., Bracciaferri, F. and Pedotti, A. (1996) Accuracy assessment in microgravity human movement analysis, In Proceedings VIth European Symposium on Life Sciences Research in Space. Trondheim, Norway, ESA SP-390, pp.129-134.
2. Ferrigno, G., Baroni, G., Anolli, A. and Pedotti A. (1996) quantitative analysis of movement control in prolonged mgravity. In Proceedings VIth European Symposium on Life Sciences Research in Space. Trondheim, Norway, ESA-SP 390, pp.105-110.

# EFFECT OF ORIENTATION ON HUMAN POSTURE IN NEUTRAL BUOYANCY AND PARABOLIC FLIGHT

Julianne Zimmerman and David L. Akin
University of Maryland Space Systems Laboratory
College Park, Maryland USA

## INTRODUCTION

For more than three decades, manned space operations research and mission preparation activities have been conducted in simulation environments which at best provide limited fidelity representations of the on-orbit microgravity environment they seek to model. The two most commonly employed simulation environments are neutral buoyancy (water immersion, NB) and parabolic flight (PF). It is immediately obvious that neither environment produces the same or even consistently similar cues to those provided by spaceflight. Under NB conditions, the subject's body is balanced in the water such that his or her buoyancy is compensated exactly by his or her weight. The subject ideally has no preferred orientation, and neither sinks nor rises. During PF, an aircraft is flown in a Keplerian trajectory, producing periods of "zero gravity" (0g) alternating with periods of up to twice-Earth gravity accelerations. These experiences are only loosely comparable to spaceflight. Several studies using different experimental approaches, and performed over a period of multiple decades [e.g., 1-4], demonstrated that human subjects evidence asymmetric responses to neutral buoyancy conditions, particularly with respect to orientation. Lackner and colleagues [e.g., 5], among others, have also shown that human subjects experience a wide variety of perceptual effects during exposure to PF, and that these effects arise from both internal and external cues. The two environments very clearly affect subjects in observable ways, but the implications of those effects for the successful simulation of spaceflight conditions has not been explicitly assessed. Those implications must be understood in order to improve the fidelity of spaceflight operations research and astronaut or cosmonaut training exercises which must necessarily be performed in these two environments. The purpose of the present study is to examine the measurable effects of orientation in NB and PF environments on relaxed human posture.

## METHODS

Eight healthy adults (4 men and 4 women) were asked to adopt relaxed postures while "standing" in space shuttle middeck standard-type foot restraints, in NB and during the 0g periods of PF. Subjects were tested in NB in 9 orientations, 3 trials each: Upright; tilted 45° Front, 45° Back, 45° Right, 45° Left; tilted 90° Front, 90° Back, 90° Right, and 90° Left. PF conditions prohibited 90° testing; consequently the PF test protocol included only the Upright and 45° orientations. All NB testing was performed at the bottom of a 25' deep NB facility, using SCUBA certified, experienced test personnel. All PF testing was performed during four flights on the NASA KC-135. Subjects were fully informed of the test protocol prior to and during testing. Reaction loads were recorded for each foot independently and normalized to subjects' masses; hip, knee, and ankle angles were measured from photographs. Trunk and shank angles (defined respectively by hip-shoulder ray and ankle-knee ray angles to the foot restraint plane) were also recorded. ANOVA (analysis of variance) calculations were performed for individual and group comparisons.

## RESULTS

All subjects completed all trials for NB; however, some trials were lost during PF, due to subject motion sickness. Nonetheless, statistically significant differences ($p<.05$) between orientations for each environment (e.g., PF 45F v. 45B) were present, primarily for the ankle angle and Fy. Differences were also found between environments (NB v. PF) and particularly between subjects for both postural angles and reaction loads. Differences between test conditions for each subject (e.g., ES ankle angle, NB v. PF) were likewise marked. Differences between right and left foot reaction loads were sparse for PF and more numerous for NB, but differences between environments (NB v. PF) were widespread.

## DISCUSSION

Postural configurations adopted by test subjects placed in both environments are found to differ in curious ways: statistical analyses showed that between-orientation, between-subject, and between environment discrepancies are all present, but lacking apparent patterns. Within-subject discrepancies are likewise present and erratic. These results support the earlier results [1-4] and further indicate that the human body, individually and in general, perceives and responds uniquely not only to orientation but also to the environments. As previously stated, neither NB nor PF offers particularly good representation of spaceflight conditions, and clearly NB and PF cannot be considered interchangeable. Thus simulations performed in either of these environments cannot be expected *a priori* to accurately represent spaceflight, but must be evaluated carefully for their relevance and potential to mislead both subjects and researchers. Further studies currently in progress will report on comparisons to spaceflight postural angle and reaction load data.

## REFERENCES

(1) Knight Maj LA. 1958. An Approach to the Physiologic Simulation of the Null-Gravity State. *J. Av. Med.* 29:283-286.

(2) Schock GJD. 1960. Perception of the Horizontal and Vertical in Simulated Subgravity Conditions. *US Armed Forces Med. J.* 11(7):786-793.

(3) Schöne H. 1964. On the Role of Gravity in Human Spatial Orientation. *Aerospace Med.* 35 (8): 764-772.

(4) Nelson JG. 1968. Effect of Water Immersion and Body Position upon Perception of the Gravitational Vertical. *Aerospace Med.* 39:806-811.

(5) Lackner JR. 1992 . Spatial Orientation in Weightless Environments. *Perception* 21:803-812.

**Aircraft Control Forces and Electromyographic Activity During Landings in an Aermacchi MB339CB Flight Simulator.**

D. J. Hewson[1], P.J. McNair[2], and R.N. Marshall[3].
1. Aviation Medicine Unit, Royal New Zealand Air Force
2. Neuromuscular Research Unit, Faculty of Health Studies, Auckland Institute of Technology
3. Department of Sports Science, University of Auckland

The magnitude and duration of the force required to control an aircraft in flight can vary greatly, and it may often be necessary to apply force simultaneously to two or more controls. Little information about the magnitude of aircraft control forces is available, although upper limits to the size of the forces that need to be applied are set by the United States Federal Aviation Administration (FAA) and the United States Air Force (USAF). Although these control force specifications exist, few data are available on the quantification of control forces in aircraft. The USAF and FAA control forces were empirically derived from levels found to be acceptable to male test pilots, and in the one USAF study in which control forces were reported, the investigators found control forces in an Iroquois helicopter which exceeded design standards. Some investigations have been focused on the ability of pilots to produce the force limits specified in the design standards. Most of these studies have provided evidence that a number of pilots may not be able to match the specified force. This paper represents one phase of an investigation of aircraft control forces and pilot strength. The focus of this phase of the study was the force required to fly a flight simulator in strength-critical situations, and to measure muscle activity during those tasks. During subsequent investigations control forces and muscle activity during actual flight will be recorded. The subjects were nine experienced Royal New Zealand Air Force (RNZAF) pilots (1066 ± 1010 flying hours), and 12 novice RNZAF pilots under training (134 ± 18 hr). All testing was carried out in the RNZAF Aermacchi MB339CB Flight Simulator at RNZAF Base Ohakea. Subjects flew a using a randomised sortie profile, which included 25 manoeuvres, each performed three times. Landings were performed with five different trim settings: neutral trim; full nose-up elevator trim; full nose-down elevator trim; full left aileron trim; and full right aileron trim. The runaway trim settings were used to simulate strength-critical tasks. Control force data from the simulator was collected using Chart software and a MacLab (sample frequency 400 Hz). Flight variables were recorded from the simulator using RealTime Monitor (sample frequency 8 Hz). Electromyographical (EMG) data was captured using SuperScope (sample frequency 500 Hz) from triceps brachii, biceps brachii, pectorals, deltoid, the wrist extensor muscle group, the wrist flexor muscle group, and quadriceps. EMG data was normalised by calculating the Root Mean Square (RMS) of each signal, and expressing the RMS as a percentage of the RMS during maximum voluntary contraction (MVC). Little muscle activity was present during the manoeuvres; pilots never used more than 50% MVC. Novice pilots used more EMG activity than experienced pilots to control the aircraft during landing in the following muscles: wrist flexors (19% more EMG activity than experienced pilots); wrist extensors (120%); triceps (38%); and deltoid (26%). Experienced pilots displayed more quadriceps activity during landing than did the novice pilots (53%). There were no consistent differences between novice and experienced pilots in EMG activity of biceps or pectorals. Novice pilots displayed more variability in yaw and pitch than experienced pilots during landing (50 and 17% respectively). However, experienced pilots rolled more on the approach to land than novice pilots (25%). Experienced pilots applied larger rudder forces (59%) and smaller elevator forces (20%) during landing than novices. There was little difference in the amount of aileron force applied. The increased EMG activity of novice pilots may have been due to 'white-knuckle' syndrome, a common occurrence among trainee pilots. The increased variability of novice pilots in both pitch and yaw was also predictable. However, the greater roll of experienced pilots was not expected. Familiarity with the roll parameters in which the Aermacchi can land may have been a contributing factor in this finding.

## 3-D QUANTITATIVE ASSESSMENT OF NEUTRAL BODY POSTURE

G.Andreoni[1,2], G.Ferrigno[1,2],G.Baroni[1,2] and A.Pedotti[1,2]
1 - Politecnico di Milano, Dept. of Bioengineering, Milan, Italy
2 - Centro di Bioingegneria, Fnd. Pro Juventute "Don C.Gnocchi" - Politecnico di Milano, Milan, Italy

### INTRODUCTION

In microgravity conditions, when relaxed and unrestrained, the human body assumes a particular position called "neutral body posture".
This posture has been rarely studied and described only by photographs in a few days mission in space.
In this research we present the results of the experiments carried out during the EUROMIR '95 mission and related to the description of the weightlessness posture and its modification during a prolonged period of permanence in microgravity.

### METHODS

The experiments were carried out using the ELITE-S system which was modified (from the commercial system) and qualified for space utilisation. It performs a real time processing of images from 1 to 8 tv-cameras in order to recognise in the field of view the presence of small passive markers; it calculates the 3-D co-ordinates of these markers which can be easily attached to the anatomical landmarks.
On the subject 41 markers were positioned. They could be divided in anatomical markers, which were placed directly in correspondence of anatomical points, and technical markers, which were positioned on sticks pointing at anatomical points that could be extrapolated.
During EUROMIR '95 mission 2 subjects were studied in different experimental sessions. The acquisitions were performed at a sampling rate of 10 Hz for 2 sec.

### RESULTS & DISCUSSION

From the data we calculate the 3-D angles of the joints between the segments measured which is the posture description required.

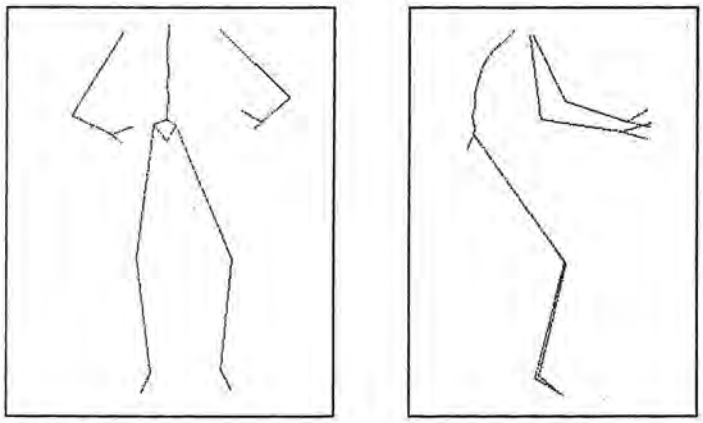


**Figure 1: Stick diagram of neutral body posture in the frontal plane (left) and in the sagittal plane (right).**

In particular the spine conformation showed a pronounced kyphosis; this derives from the equilibrium of compressive and extensive forces respectively corresponding to the actions due to ligaments and intervertebral disks which in microgravity were more filled with water. In fact, spine length respectively increased of about 1% and 4% for the two astronauts.
The results allow for improving the spacecraft design. This is an important objective in particular oriented to a permanently inhabited orbiting station as in program for the next years.

### REFERENCES

Griffin, B.N. (1978): The Influence of Zero-G and Acceleration on the Human Factors of Spacecraft Design. Design Guide, 1-52.
NASA-STD-3000, Vol. 1, Rev. B., 1-78.

### ACKNOWLEDGEMENTS

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# MECHANICAL POWER AT THE ANKLE JOINT DURING STANDING

Håkan Lanshammar and Annica Karlsson
Systems and Control group, Uppsala University, Sweden

## INTRODUCTION

Different measures based on the ground reaction force (F) vector and the centre of pressure (COP) are often used to characterise a persons postural control[1]. The variations in these quantities during standing is the basis for various measures such as the standard deviations and the length of travel of the COP during a specified time duration, often 30 seconds. To get a deeper understanding of the effort needed to maintain balance and for diagnosis of disorders in the neuromuscular system it could be useful to also study the mechanical power that is developed. The method that is presented here assumes that only the ankle joint is active. That this is really the case can be certified using the method described by Karlsson *et al*[1].

## METHODS

The power developed by the muscles can be calculated from $P = T\omega$, where $P$ is the joint power, $T$ is the moment acting at the joint, and $\omega$ is the angular velocity at the joint. Under the assumption that the movement is like an inverted pendulum, with only the ankle joint involved in balance actions, it is straight forward to develop approximate expression for $P$ from equations presented in [1].

In the equations given in the box to the right, $m$ is the body mass, $g$ is the gravitational constant, $a_x$ is the position of COP on the force plate relative to the ankle joint, $\dot{x}$ is the horizontal velocity of the body's centre of mass (COM), $l$ is the distance from the ankle joint to the COM, and $f_x$ is the horizontal force in the sagittal plane. Since $a_x$ and $f_x$ can be measured with a force plate, it is thus possible to estimate the joint power without the need to actually measure the joint angle. Care must taken, though, to make sure that drift in the force plate data does not cause the velocity estimates to assume increasing values with time. This can be accomplished with suitable detrending algorithms.

$$\left.\begin{aligned} T(t) &\approx mga_x(t) \\ \omega &\approx \frac{\dot{x}}{l} = \frac{1}{l}\int_0^t f_x(\tau)d\tau \end{aligned}\right\} \Rightarrow$$

$$P(t) \approx \frac{mg}{l} a_x(t) \int_0^t f_x(\tau)d\tau$$

To test if the suggested method can be of practical use force plate data were collected from 5 normal subjects and one MMC child. Measurements in normal standing (30s) on a force plate were made on each subject. Only the sagittal plane was considered in the analysis.

## RESULTS

In Figure 1 the joint power is plotted versus time for one normal subject and for the MMC child.

Some numeric results are presented in Table 1 for these two subjects. For comparison the standard deviations of the horizontal force $f_x$ and of the COP position, are also presented.

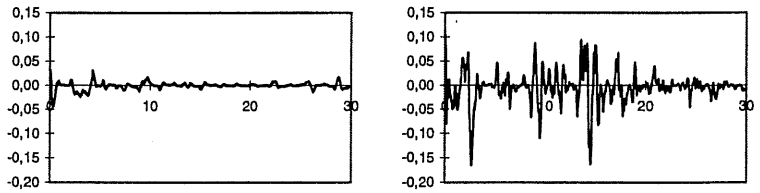



*Figure 1. Joint power (W) versus time (s) for a subject with good balance (left) and for a child with MMC (right).*

| Subject | Power S.D. (W) | Power Max-min (W) | COP S.D. (m) | $F_x$/m S.D. ($ms^{-2}$) |
|---|---|---|---|---|
| Normal | 0,0073 | 0,0765 | 0,032 | 0,011 |
| MMC Child | 0,0293 | 0,3025 | 0,069 | 0,023 |

*Table 1. Results for two test subjects. The difference in power measures are much greater than in COP or $F_x/m$.*

## DISCUSSION

The new method for assessment of postural control has the potential of complementing other measures currently in use. For the method to be relevant it is important that the subject is using the ankle strategy. Otherwise misleading conclusions may be drawn. It is obvious that the new method provides new insight, since it is reasonable to assume that the effort involved in standing is related to the mechanical work done. The formula presented for the joint power also shows how both the centre of pressure and the horizontal force affect the joint power. This is especially interesting since these variables have been previously identified as major input and output variables for the mechanical subsystem during standing[1]. As seen in the example presented, it is also to be expected that differences between normals and patients are amplified in the power variable, since it is quadratic by nature.

## REFERENCES

[1] A. Karlsson and H. Lanshammar. Analysis of postural sway strategies using an inverted pendulum model and force plate data. Gait and Posture, 5/2, 1997 (to appear).

# INCIDENCE OF FALLS RESULTING FROM TRIPS INDUCED IN OLDER ADULTS

Michael J. Pavol *, Tammy M. Owings *, Kevin T. Foley †, Mark D. Grabiner *
*Department of Biomedical Engineering, †Section of Geriatric Medicine
The Cleveland Clinic Foundation, Cleveland, Ohio, U.S.A.

INTRODUCTION. Falls are a significant source of morbidity and mortality in older adults. 11% of falls in community-dwelling older adults result in serious injury[1], with up to 53% of these falls due to tripping[2]. To reduce the occurrence of such falls, there is a need to identify what determines whether a trip is recoverable and those factors that increase an individual's risk of falling. The purpose of the present research was to estimate the proportion of trips that result in falls in healthy older adults and determine the effects of sex, age, and the foot tripped on the incidence of these falls.

METHODS. Trips of 73 healthy older adults, aged 65 to 86 yrs (median 70 yrs), were attempted in a laboratory setting. Subjects were previously examined by a geriatrician and tested for a hip bone mineral density of at least 0.65 g/cm$^2$. Trips were induced using a concealed mechanical obstacle which rose from the floor to a height of 5.1 cm in approximately 160 ms when manually triggered. A decoy "tripping rope" was laid 1.5 m before the obstacle. For safety, subjects were placed in a body harness and connected by a pair of ropes, through a load cell, to a ceiling-mounted track. Unloaded rope lengths were set with the wrists and knees 0.47 m and 0.3 m above the floor respectively. Subjects were instructed to walk normally at a self-selected speed between two specified points and told only that they would be tripped during one such trial. They were instructed to respond to the trip by recovering and continuing to walk. When appropriate, the obstacle was triggered at footstrike to induce a trip by obstructing the contralateral foot during mid-to-late swing. Only one attempt was made to trip each subject. Trip outcomes were graded as recoveries, falls, or rope-assists. In a fall, the subject was fully and continuously supported by the safety ropes. For the other trips, the load cell signal was low-pass filtered at 16 Hz and integrated over the second following the instant at which the obstacle was triggered. Those recoveries with integrated rope forces of less than 5.0% body weight•second were judged to be true recoveries, the rest were considered rope-assisted. To evaluate the effect of the foot obstructed on the outcome of a trip, the dominant foot was determined prior to testing as the preferred stepping foot in preventing a fall forward.

RESULTS. The outcomes of the 56 successful trips are displayed in the table below, both in percentages (%) and number of occurrences (n). Effects of sex and age on the proportion of falls and rope-assists are clearly visible.

| Trip Outcome | Overall (n=56) | | Sex | | | | Age | | | | Foot Obstructed | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | Males (n=17) | | Females (n=39) | | < Median (n=28) | | > Median (n=28) | | Dominant (n=25) | | Non-Dom. (n=31) | |
| | % | n | % | n | % | n | % | n | % | n | % | n | % | n |
| Recovery | 62.5 | 35 | 64.7 | 11 | 61.5 | 24 | 64.3 | 18 | 60.7 | 17 | 60.0 | 15 | 64.5 | 20 |
| Rope-Assist | 21.4 | 12 | 29.4 | 5 | 17.9 | 7 | 10.7 | 3 | 32.1 | 9 | 24.0 | 6 | 19.4 | 6 |
| Fall | 16.1 | 9 | 5.9 | 1 | 20.5 | 8 | 25.0 | 7 | 7.1 | 2 | 16.0 | 4 | 16.1 | 5 |

DISCUSSION. Epidemiological studies have reported the relative number of falls in older adults that occur as a result of trips. However, to date, none have attempted to determine the proportion of trips that actually result in falls, nor the effect of different factors on the relative ability to recover following a trip. Such information is needed to evaluate the extent to which the observed increases in trip-related falls in older adults may be attributed to a greater rate of tripping or to a lesser ability to recover from a trip.

The results indicate that the majority of trips in healthy older adults do not result in falls. However, an exact rate of falling could not be determined due to the ambiguity of the rope-assist outcome. These rope-assist recoveries appear to be characterized by rapid and moderate-to-large trunk flexion following the trip. The resulting force impulses through the rope to the mid-scapular region presumably arrested trunk rotation sufficiently to allow a recovery of dynamic balance. Unfortunately, it is presently impossible to confidently estimate how many of the rope-assisted individuals would have fallen without the aid of the ropes.

There were no sex- or age-related differences in the proportion of recoveries. However, the percentage of falls in the females was significantly higher than in the males, a result that mirrors the relative incidence of trip-related falls in this population[2]. These results imply that the source of this sex difference is intrinsic and unrelated to the frequency or type of trip. Surprisingly, individuals above the median age were less likely to fall upon tripping than those below, a result that contrasts with the reported increase in the prevalence of falls with age[2]. This may reflect an alteration of gait with aging that reduces one's chance of falling following a trip, as well as an increased rate of tripping with age. Finally, there appears to be no influence of the foot tripped on one's ability to recover.

REFERENCES.
1. Tinetti et al. (1988). N Engl J Med 319, 1701-1707.
2. Blake et al. (1988). Age Ageing 17, 365-372.

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# GROUND REACTION-FORCE CHARACTERISTICS OF HEALTHY YOUNG ADULTS IN NEGOTIATING OBSTACLES

R K Begg, W A Sparrow*, N D Lythgo and S Lewis

Centre for Rehabilitation Exercise & Sport Science, Victoria University, Melbourne, Australia. *School of Human Movement, Deakin University, Burwood, Victoria, Australia.

## INTRODUCTION

Traditionally the major focus of human locomotion studies has been straight line, unobstructed gait. It is only recently that researchers have begun to examine the biomechanical characteristics of the adaptive gait strategies used to negotiate uneven terrains that are commonly encountered in the everyday environment (Patla & Rietdyk, 1993). Relatively few of these studies have examined the kinetic characteristics of obstructed gait when avoiding an obstacle by stepping over or across it. Measurement and analysis of the ground reaction forces (GRFs) both before and after obstacle clearance may provide useful information about the control strategies employed by the locomotor system in a perturbed environment. The experiment reported here used a dual force platform procedure to examine the GRFs of both the lead and trail foot as individuals stepped over obstacles adjusted to various percentages of leg length.

## METHOD

Twelve healthy young adults (6 male and 6 female) participated in the study. Initially each subject completed 8 unobstructed walking trials (control trials) along a 10 m walkway. Two AMTI strain gauge force platforms (FPs) embedded near the centre of the walkway recorded the lead (first foot to cross the obstacle) and trail foot GRFs. Subjects then completed a block of obstructed walking trials in which three obstacle height conditions were presented in randomised order. The obstacle consisted of a wooden rod resting on two adjustable stands graduated in mm. The obstacle heights corresponded to 10%, 20% and 30% of a subject's leg length and was positioned between the FPs such that both the lead and trail foot GRFs were recorded. Foot GRFs were sampled at a rate of 200 Hz for 3 s following initial contact with the first FP. The outcome variables examined in this study included: i) obstacle crossing step length, support time and obstacle crossing speed; ii) lead and trail foot vertical and horizontal GRF-time characteristics (peak forces and timing) during selected phases of the gait cycle (weight acceptance, midstance and push-off); and, iii) vertical and horizontal force-impulses. A 3 factor (height, gender, limb) analysis of variance with repeated measures was used to analyse the force-time data.

## RESULTS

No reliable gender effects on the force-time data were found. Obstacle crossing step length increased in obstructed gait ($p < 0.05$), but was not influenced by obstacle height. Crossing speed decreased significantly in obstructed gait and across obstacle heights ($p < 0.001$), and trail foot support time was significantly longer for all obstacle heights ($p < 0.01$). The vertical peak force during the push-off phase differed significantly between the limbs and across all obstacle heights ($p < 0.01$); trail foot data being significantly greater than lead foot data. A high degree of timing invariance was found in both the horizontal and vertical force components despite the requirement to step over an obstacle of varying height. The only consistent difference in any of the timing characteristics was that the maximum braking force of the trail foot occurred later in the normalised step cycle ($p < 0.01$). Significant vertical impulse differences ($p < 0.001$) were found between the lead and trail feet and across obstacle heights. The trail foot also produced greater braking and propulsive forces across obstacle heights.

## DISCUSSION

The results of this study showed that some of the force-time outcome variables measured during obstructed gait are obstacle-height dependent while others remain invariant with changes in obstacle height. The trail foot generated greater vertical and horizontal GRFs during the push-off phase across all obstacle conditions, had lower vertical peak force during mid-stance, and produced greater vertical and horizontal impulses. While maximum propulsive force increased with obstacle height its timing in the normalised step cycle was not influenced by height. The results suggest that the different constraints on the lead and trail foot impose different kinetic demands. The trail limb's role is to arrest the body's forward momentum, provide stable support for the lead limb, and generate sufficient vertical impulse to ensure adequate elevation and forward momentum of the center of mass whereas the lead limb stance control necessitates relatively lower vertical and horizontal impulses due to the fact that center of mass elevation does not need to be as great during trail limb crossing. The results also suggest that selected kinetic variables may be used in a clinical setting to monitor performance and provide indices of normalcy for obstructed and unobstructed gait.

## REFERENCE

Patla, A.E. and Rietdyk, S. (1993) *Gait Posture*, **1**: 45-60.

Prior to the collection of the obstructed gait data subjects were allowed adequate practice trials in order to determine their lead and trail foot. Prior to this analysis these data were normalised (divided by body weight) for each subject.

# MAGNETIC RESONANCE IMAGING IN GAIT ANALYSIS: DIRECT DETERMINATION OF JOINT CENTRES FROM SKIN MARKER LOCATIONS

**K. Kuspert*, D. Scale†, A. Arndt°, G.-P. Brüggemann°, W. Hauger***
* Universtity of Darmstadt, Institute for Mechanics, I. Germany
† University Orthopedic Hospital, Frankfurt/Main, Germany
° German Sport University Cologne, Institute for Athletics and Gymnastics, Germany

## INTRODUCTION

An important aspect of gait analysis is the assessment of the relative position of external marker data and relevant internal anatomical locations such as joint centres, the tibial plateau and reference points on the femur and tibia in order to incorporate individual anatomy into the musculoskeletal model used. An accurate determination of the hip joint centre is especially problematic. The available methods for prediction of hip centre location have several severe inherent limitations. Procedures not based upon radiographs or magnetic resonance images (MRI) have questionable accuracy. According to Bell (1990) three commonly applied methods have an accuracy of 20 - 40 mm with a range of 5 - 65 mm for individuals. If one considers that the lever arms of hip muscles are in the same order of magnitude then an adequate prediction of muscle forces is impossible. Delp (1993) predicted a modification of the moment arm of up to 40% and a change in moment generating capacity of the hip up to 50% when moving the hip joint centre 20 mm. Theoretically, the same problems apply to the knee and ankle joints. Existing radiographic and MRI methods are more accurate but provide only internal landmarks with respect to external skin markers. The problem lies in estimating the distance between the two.

## METHODS

A new method was developed for determining important internal landmarks such as the hip joint centre directly from skin marker positions. The method was based upon the idea of leaving the markers attached to their specific gait analysis locations and making them visible on MRI scans. One inch diameter reflective spheres mounted on plastic plates were used as skin markers. Plastic screws and threads drilled into the spheres allowed them to be removed from the plates which were taped to the subjects' skin. Plastic was chosen because many other materials would have disturbed the MRI scans. After the gait analysis data capture (Motion Analysis System, 120Hz) the spheres were removed and the subjects transported to the MRI unit. A second marker set covered with a fat film and plastic foil was attached to the plates within the MRI unit. The subject's leg was constrained by wedges and an orthopedic L-guide for the feet to maintain an absolutely motionless position. The subject lay upon a foam plate which provided room for the marker attached to the sacrum. Transverse slice images were made on which the fat layer around the spheres and the hip joints appeared as different size circles. A three dimensional reconstruction of the slices provided the centre of the spheres and the hip joints. Other internal locations of interest could easily be obtained from the images.

## RESULTS AND DISCUSSION

The method developed provides a direct measurement of joint centres and other internal structures of interest relative to external markers. The accuracy of determining the centre of a sphere from the MRI scans was approximately ±1 mm. As the movement of the segments was known, the motion of any relevant internal structure of that segment could be easily determined by the relative vectors from the MRI scans. There appeared to be no significant disturbance through skin movement between standing and lying in the MRI unit provided markers were attached to bony external landmarks. The results indicated only minor dislocation relative to the position data provided by the video system. The procedure was therefore, seen as an easily applied and accurate method for determining internal landmarks from external skin marker locations which is imperative for the reduction of errors in high quality gait analysis. Furthermore, MRI systems omit no radiation and provide high resolution images although a drawback is still the cost involved with the MRI technique.

## REFERENCES

Bell, A.L., Pedersen, D.R. and Brand, R.A. (1990). A comaprison of the accuracy of several hip centre location prediction methods. J. Biomechanics, 23(6); 617-621.

Delp, S.L. and Maloney, W. (1993). Effects of hip centre location on the moment-generating capacity of the muscles. J. Biomechanics, 26(4); 485-499.

# A PROCEDURE FOR QUANTITATIVE KINEMATIC ANALYSIS IN CYCLING

R. RODANO[1], R. SQUADRONE[2]
[1] Bioengineering Center (Politecnico di Milano - Pro Juventute Fnd.), Milan, Italy
[2] Associazione Scuola Viva, Rome, Italy

## INTRODUCTION

The analysis of lower limb kinematics in cycling has generally been confined in the sagittal plane by using data obtained with a 2-D analysis. When asymmetries were evaluated this was done by detecting data from one leg at time and subsequently comparing the scores obtained from the two sides in different trials.

The purpose of this paper is to describe a method and a special developed software package capable of doing a complete 3-D kinematic analysis in cycling, including the assessment of kinematic asymmetries by a simultaneous right-left analysis.

## METHOD

The first step in the present methodology is the capture of the motion of the athlete while pedaling on his own bicycle mounted on rollers fitted with a magnetic-operated variable-load device. This operation is carried out by the ELITE system motion analyzer (BTS srl) which, by means of four 100 Hz TV cameras paired off on the two sides of the cyclist, allows automatic 3-D reconstruction of a number of anatomical landmarks marked through the application of small passive reflective markers. Then, the 3-D co-ordinates and the appropriate body segment parameters, stored in data files, are used as input for the computer program CICLO which was written in Matlab (4.2b version for Windows).

A fourteen segment rigid link model was used to represent the cyclist body for the analysis: pelvis, back, torso, neck-head, and right left arms, forearms, thighs, legs and feet. The software package also includes a relational data base written in C for the management of the quantitative and statistical comparison among the computed kinematic and asymmetries indices.

## RESULTS

The program is capable of producing a large amount of data, all of which is not reasonable to report here. In particular it provides a unique array of graphic display to allow the user to instantly diagnose and treat asymmetry problems. When we start to run the program we can see the main menu on the screen with five menu options. 1) *Enter data*: after an automatic identification of the main pedaling cycle events (Top Death Center - TDC, Bottom Death Center - BTC) this routine normalizes the time over all the pedaling cycles recorded. At least fifteen cycles per trial are recorded. 2) *Pre-elaboration*: calculates internal joint rotation centers using markers co-ordinates and the measured anthropometric parameters. 3) *Trajectories*: plots and compares the averaged trajectories of left and right internal joint centers. 4) *Angles*: calculates relative and absolute joint angles and plots average angle patterns as a function of crank angle. Automatically identifies meaningful asymmetries. For each joint, left-angle vs. right-angle plots are also used to evaluate asymmetries. 5) *Post elaboration*: stores the data in the data base. Figure 1 is an example of the knee joint rotation center trajectory plots in the sagittal, frontal and horizontal plane for a subject of this study. The plot compares left and right patterns, the left and right differences for selected kinematic parameters are also reported.

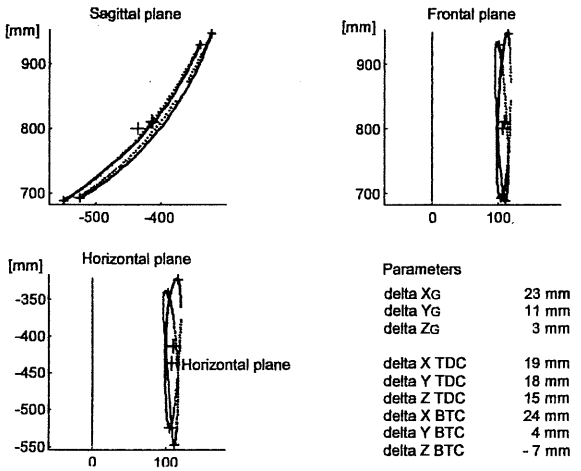



*Figure 1. Knee joint rotation center trajectory plots and parameters report. Thick curves belong to the right knee.*

## DISCUSSION

The method and the user friendly software presented here has been applied to assess kinematic problems of many professional cyclists and it seems to be an useful tool for scientists, trainers and athletes to evaluate biomechanical data during cycling.

The proposed model gives a good representation of the cyclist during its action. In particular the possibility to collect simultaneously data from both sides of the body appears to be very informative about asymmetries characterizing cyclists. Future versions of the software will permit inclusion of external forces acting on the body to allow joint moment and power calculation.

# RELIABILITY OF POOL TESTING: ABILITY OF SWIMMERS TO SWIM AT PREDETERMINED SPEEDS

K. L. Keskinen
Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

## INTRODUCTION

Pool testing, due to its utmost specificity, has become increasingly important in evaluating the conditioning of swimmers during training process. Prominent part of the training have been performed by applying these results in practice as such. However, only little attention has been paid to the reliability and validity of testing as well as to one's ability to execute the test swimming according to given instructions. The present project was thus designed to investigate the test-retest reliability of a commonly used pool testing method. This was done by examining the biomechanical and physiological responses of swimmers to controlled swimming velocities and also by examining the swimmers ability to swim at predetermined speeds. The aim of this paper was to concentrate on the problem: How well the swimmers could follow a standardized pool testing protocol with predetermined swimming speeds?

## METHODS

Thirty swimmers, males and females, from different age categories and performance levels volunteered to serve as subjects (Table 1). The swimmers performed a series of ten 100-m swims twice in three days. Before the start of the first trial the swimmers were asked to estimate their best performance during the swimming. Starting velocity was fixed to a level which was 0.45 $m*s^{-1}$ less than the estimated maximum. The speed was then increased by 0.05 $m*s^{-1}$ until the highest level was reached. During the last swim the swimmers were allowed to go as fast as possible disregarding the predetermined speed. The second exercise was repeated with identical time table as the first one. Swimming velocities were adjusted by successive lights underwater in the bottom of the 50-m pool (40 lamps with 1.25 m intervals). Two lamps at one time were active and the swimmers were informed to follow the lights their heads above the latter lamp so that both of the lamps could be seen while the face was in the water. Rest interval between the 100-m swims was 90 s. Time for the 100-m swims as well as for 50-m laps were taken by hand held watches to calculate mean velocity. The time for starting and finishing five meters as well as for turning ten meters was measured during each of the swims to calculate the mean velocity of 40-m distance in the middle of each pool length. Correlation coefficient and coefficient of linear regression were computed between parameters. T-test between paired observations was used to test the statistical significance of the differences at $p < 0.05$.

Table 1. Age, stature, body mass, body mass index, hydrostatic weight, arm span and personal best in 100-m swim

| | Age (y) | Stature (m) | Mass (kg) | BMI ($kg/m^2$) | HW (kg) | AS (m) | BP100 (s) |
|---|---|---|---|---|---|---|---|
| All (N=30) | 16.7 | 1.70 | 61.8 | 21.3 | 2.64 | 1.75 | 65.2 |
| Males (15) | 16.0 | 1.72 | 62.8 | 21.2 | 3.03 | 1.79 | 62.8 |
| Females (15) | 17.3 | 1.68 | 60.7 | 21.5 | 2.25 | 1.72 | 67.3 |

## RESULTS AND DISCUSSION

Regression analysis (n=299) between the preprogrammed and measured 100-m swimming speeds showed that the swimmers could follow the lights accurately ($R^2 = 0.987$ and $R^2 = 0.986$) in the two series, respectively. Same was true when the velocity was calculated by 80 m distance ($R^2 = 0.984$ and $R^2 = 0.981$). The situation was similar when the measured values of the two exercises were compared between each other ($R^2 = 0.994$ and $R^2 = 0.993$) in the 100-m and 80-m distances, respectively. The differences were all statistically non-significant.

The situation was however changed when the comparisons were made between individual records. Significant ($p < .05$) differences were found between the preprogrammed and measured swimming speeds in 13 and 15 trials out of 30 in 100-m distance, and 23 and 21 out of 30 in 80-m distance in the two cases, respectively. The most inaccurate swims were the first four and number nine in the series of ten. The swimmers themselves reported that they could easily follow the lights in the middle of the pool and no difference between the swimmers' speed and the speed of the lights was observed by looking from the pool side. However, most swimmers experienced notable difficulty in adjusting starting and turning speeds to fit with the steady-paced lights. Especially problematic were the starts, even though the starting was from the water, while most of the swimmers were able to glide underwater faster than the lights and consequently they had to wait for the lights before they could start stroking again.

It was concluded that although some difficulties in starting and turning the successive lights appeared to be a good means to adjust the swimmers' speed in standardized pool testing conditions. The test-retest reliability was considered high even though individual differences were found between the predetermined and measured speeds.

## ACKNOWLEDGEMENTS

The Finnish Ministry of Education supported this project financially. Technical assistance was provided by Mr. Ossi P. Keskinen, M.Sc. from the Department of Biology of Physical Activity.

# SHOULDER LOADS AND FORCE DISTRIBUTION DURING SANDING OF A CEILING

L. Lindbeck[1], D. Karlsson[2], S. Kihlberg[1], K. Kjellberg[1], K. Rabenius[3], B. Stenlund[4], and J. Tollqvist[3].
[1] The National Institute for Working Life, Solna, Sweden, [2] Prosolvia Research & Technology AB, Jönköping, Sweden, [3] The Swedish Construction Industry's Organization for Working Environment, Safety and Health, Stockholm, Sweden and, [4] Karolinska Hospital, Stockholm, Sweden.

## INTRODUCTION

Among painters there is a high frequency of neck and shoulder complaints and a high risk for early retirements on account of these complaints. Some of the tasks which are particularly strenuous to the neck and shoulders are sanding and painting of ceilings with tools attached to handle extensions. The distribution of forces among the shoulder muscles when working with a handle extension is unclear. The anatomy and function of the shoulder joint is complicated and makes direct measurements of muscle activity difficult. Theoretical biomechanical models of the shoulder may allow an estimation of the muscle loads. The purpose of this study was to quantify the loads on the shoulder during sanding work and to determine the likely distribution of the muscle forces across the shoulder during this work by applying a computerized biomechanical model. The model predicts 46 different muscular and contact forces in the shoulder when external load, body position and anthropometric data are known (1).

## METHODS

Measurements were made by means of a handle extension supplied with strain gauges and a Kistler™ force plate, in an experimental set-up. Forty experienced male painters participated in the experiments and were instructed to perform sanding of a ceiling in their own free style while standing on the force plate. Signals from the handle extensions and from the force plate, as well as motion data recorded with an optoelectronic motion analysis system, MacReflex™, were analyzed with the biomechanical model for force distribution in the shoulder.

## RESULTS

The biomechanical analysis of the majority of the painters showed that a typical moment produced by the low-hand shoulder (the side where the hand is held low) consists of a combination of 20 Nm forward flexion and 15 Nm inward rotation. For the high-hand shoulder the moment was 30 Nm in forward flexion together with a 10 Nm abduction moment. However, although the work technique at visual inspection seemed to be similar between subjects the total shoulder load as well as the calculated force distribution sometimes varied largely. In Figure 1 is shown an example of the distribution of maximum forces larger than 20% of the estimated maximum force capacity for three individuals using the same work technique.

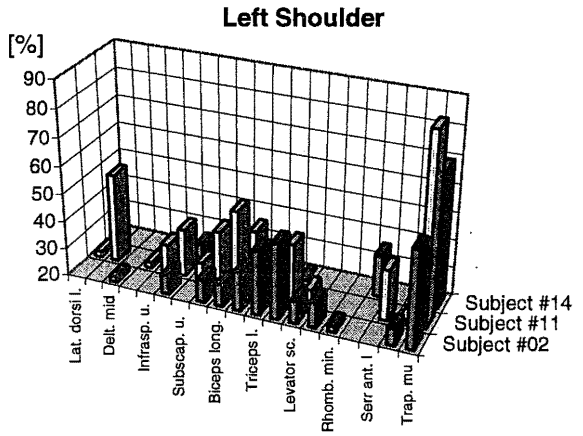



Figure 1. Distribution of the largest muscular peak forces during one working cycle for three subjects.

## DISCUSSION

Subtle, not visually observable individual variations in work technique, may explain why some workers develop musculoskeletal disorder while other do not. This study shows that differences in both the distribution of muscle forces and the total load in the shoulder can be detected between two persons, although they appear to use the same work technique.

## REFERENCES

1. Karlsson D, Peterson B. Towards a model for force predictions in the human shoulder. Journal of Biomechanics 1992;25(2):189-199.

## METHODOLOGY FOR THE "DRAG CRISIS" DETECTION IN SERVICES EXECUTED BY HIGH LEVEL VOLLEYBALL ATHLETES

P.Deprá[1]; R. Brenzikofer[2]; R. Barros[3]; E.C.Lima[3]
[1] - State University of Maringá/PR and State University of Campinas/SP - Brasil
[2] - Institute of Physics "Gleb Wataghin", State University of Campinas/SP - Brasil
[3] - Faculty of Physical Education - State University of Campinas/SP-Brasil

### INTRODUCTION

With this study we looked for the detection and characterization of the drag forces that opposes to the volley ball movement through air. We worked with services that have velocities between 10 and 20 m/s. For the volley ball these velocities, with Reynolds number between 1 and $2.5 \cdot 10^5$, correspond to the "drag crisis" described in the mechanics of fluids. The phenomenon of crisis is characterized by a sharp drop of the fluid resistance force magnitude on the body in movement near these Reynolds numbers[1]. The magnitude of the drag force acting on a smooth sphere of diameter D ($A=\pi D^2/4$) ,dislocating in a fluid of density $\rho$ with a velocity V is given by $F_D=\frac{1}{2}C_D\rho AV^2$. $C_D$ is called drag coefficient and is the variable whose sharp variation characterizes the drag crisis. The Reynolds number is given by the expression $R_e=\rho DV/\mu$, where $\mu$ is the fluid viscosity coefficient.

### METHODS

Five different types of services and others typical volley throws were realized by high level athletes. Two video cameras recorded the balls movementation in the previous measured scenery. Seven throws were selected, digitized and 3D reconstructed by the DLT method. We analyzed each trajectory in his own vertical plane. We obtained pairs of coordinates (x: horizontal; y: vertical) that corresponds to the ball's positions for each 1/30 second. x(t) and y(t) functions were adjusted by 5th degree polynomial fit for each trajectory. With this we obtained the velocities ($\dot{x}$, $\dot{y}$) and accelerations ($\ddot{x}$, $\ddot{y}$) functions.

$$C_D^2 = \frac{4m^2}{\rho^2 A^2} \frac{\ddot{x}^2 + (\dot{y}+g)^2}{(\dot{x}^2 + \dot{y}^2)^2}$$

Applying Newton's laws one obtains the time equations of the horizontal and vertical movement of the ball considering only his weight and the drag force. These two equations are summed in squared form and $C_D^2$ isolated.

### RESULTS AND DISCUSSION

We can observe that the $C_D$ results are compatible with the literature data. The "drag crisis" appears clearly in the 1.3 - 2.5 region of Reynolds Number (Re) where all of our successful services lies. Both extreme plotted points on the graphic lies out of the crisis region and were shown to be "not valid services".

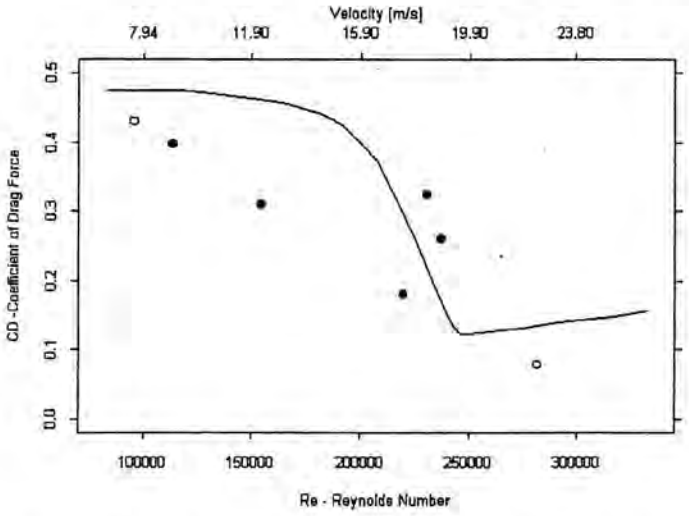



**Figure 1 - CD Coefficient versus Reynolds Number. Successful services (full circle); not valid services (open circle); fluid mechanics data from literature (line)**

### REFERENCE:

1.LANDAU, L.D. & LIFSHITZ, E.M. Fluid Mechanics. Pergamon Press, Oxford, 1993.

**Supported by: PICD/CAPES, CNPQ, FAEP, FAPESP**

# DRAG, LIFT, LIFT-TO-DRAG RATIO AND PITCHING MOMENT DURING EARLY FLIGHT IN SKI JUMPING MEASURED USING A 2 / 3 SCALE MODEL IN WIND TUNNEL

P.H. Luhtanen*, M. Pulli** and P.P. Kemppainen***

* Research Institute for Olympic Sports, Jyväskylä, Finland , ** Finnish Ski Association, Helsinki, Finland, *** Laboratory of Aerodynamics, Helsinki University of Technology, Espoo, Finland

## INTRODUCTION

Flight distance in ski jumping depends greatly on the actual velocity of the jumper at the release phase. The posture changes that take place during flight mean changes in the drag, lift, lift-to-drag ratio and pitching moment of the jumper - equipment system. A critical phase for a flight distance is the transition phase from the release phase to a proper flight posture. In this phase, there should be no decrease in velocity due to high drag. Vaverka et al. (1995) and Arndt et al. (1995) have described 3-D flying postures of ski jumpers but not their aerodynamics. The purpose of this paper was to measure the lift, drag, pitching moment coefficients and lift-to-drag ratio of a ski jumper's model in a wind tunnel during the transition phase of flight (0-1 sec, 1-25 m), as suggested by Vaverka et al. and Arndt et al.

## METHODS

The measurements were performed using a 2/3 scale model in a subsonic Göttingen-type closed circuit wind tunnel with cross section area of 3.68 $m^2$ (Luhtanen et al., 1995). The model was attached to an external six-component platform-balance. The force measurement accuracy was better than 1 % with a blockage ratio 5 % of most. The sampling rate for the force measurements was 0.6 Hz. The force values were averaged for six seconds. The drag (Fx), lift (Fz) and pitching moment (My) were transformed into dimensionless coefficients as follows: drag coefficient Cx = Fx / qS, lift coefficient Cz = Fz / qS and pitching moment coefficient Cmy = My / qSl, where q is the kinetic pressure, l the aerodynamic reference length and S the aerodynamic reference area. The lift to drag ratio was L / D = Cz / Cx. The model was officially equipped. Length of ski was 170 cm, and average width 77 mm. The tip of the ski was 57 %. In the model, the neck, shoulders, wrists, hip, ankles, toe-out and ski attachment had adjustable joints.

## RESULTS

Lift, drag, pitching moment and lift-to-drag ratio of the model with the different ski and selected joint angles with respect to the distance from the edge of the takeoff table are shown in Table 1.

Table 1. Ski and selected joint angles of the measurements at different distances from the edge of the takeoff table.

| Distance from the edge | 3 m | 4 m | 5 m | 6 m | 10 m | 15 m | 20 m | 24 m |
|---|---|---|---|---|---|---|---|---|
| Ski angle from horizontal (deg) | -3 | 2 | 6 | 10 | 18 | 22 | 24 | 25 |
| Ankle angle (deg) | 73 | 65 | 57 | 49 | 32 | 22 | 15 | 10 |
| Hip angle (deg) | 134 | 140 | 146 | 150 | 160 | 165 | 168 | 170 |
| V-angle of skis (deg) | 3 | 5 | 7 | 8 | 14 | 18 | 22 | 25 |
| Drag coefficient | .309 | .330 | .323 | .331 | .390 | .387 | .346 | .309 |
| Lift coefficient | .237 | .277 | .332 | .364 | .489 | .533 | .492 | .472 |
| Pitching moment coefficient | .025 | .039 | .055 | .045 | .040 | .020 | .008 | .001 |
| Lift-to-drag ratio | .765 | .839 | 1.029 | 1.099 | 1.254 | 1.378 | 1.420 | 1.529 |

## DISCUSSION

It was shown in this study that the drag coefficient reached its maximum between 10 - 15 metres from the edge of the takeoff table and the lift coefficient after 15 metres. The pitching moment coefficient was for all distances slightly positive, with the moment in the nose-up direction. This reached its maximum between 5 -6 metres. The maximum of the lift-to-drag ratio showed a linear increase up to 24 metres. A basis for long jumps in ski jumping is to fly the transition phase with low drag and to reach maximal lift-to-drag ratio at the end of the transition phase.

## REFERENCES

Arndt et al. (1995) Journal of Applied Biomechanics, 11, 224-237.
Luhtanen et al. (1995) Book of Abstracts of the XVth Congress ISB, pp. 578-579.
Vaverka et al. (1995) Proceedings of XIII ISBS Symposium, pp. 269-272.

## MULTIFACTORIAL METHOD FOR DRIVING POSTURE ANALYSIS

G.Andreoni[1,3], M.Rabuffetti[3], A.Tarzia[2] and A.Pedotti[1,3]
1 - Politecnico di Milano, Dept. of Bioengineering, Milan, Italy
2 - FIAT Auto, D.T. - I.S.D. Ergonomic Department, Turin, Italy
3 - Centro di Bioingegneria, Fnd. Pro Juventute "Don C.Gnocchi" - Politecnico di Milano, Milan, Italy

### INTRODUCTION

Automobile ergonomics is an emerging need strictly dependent from the extremely quick diffusion of cars as transport mean and environment to be lived. In particular the seat is the element which greatly influences driver and passengers comfort. Therefore it is much important to improve manufacturing criteria in order to assure both comfort and safety. New tools have to be developed to aid the designers to analyse and to evaluate the influence of materials, shape, and structural elements on the driver's comfort (Yamazaki 1992). Our aim is to design a new methodological approach that integrates driver's posture and anthropometry to the seat characteristics as described by shape and pressure map.

### METHODS

The proposed method is based on a multifactorial experimental protocol which measures the 3-D co-ordinates of back and seat surfaces, postural anatomical angles, and the pressure distribution between body and seat. 3-D co-ordinates of surfaces and anatomical points are measured by means of ELITE system, while pressures are measured through TEKSCAN system that consists of a sensors matrix (2016 cells) of piezoresitive ink with a resolution of 1cm$^2$ in area and 1mmHg in pressure. Back 3-D shape is reconstructed in the driving posture and it is matched with the seat surface calculating their interpenetration and correlating this datum with the local pressure.

### RESULTS

The protocol was applied on ten subjects of different sex and physical constitution which set the seat regulations each one in his own and more comfortable configuration. We used a commercial car seat mounted on a car cockpit of a 'Alfa Romeo 155' model.

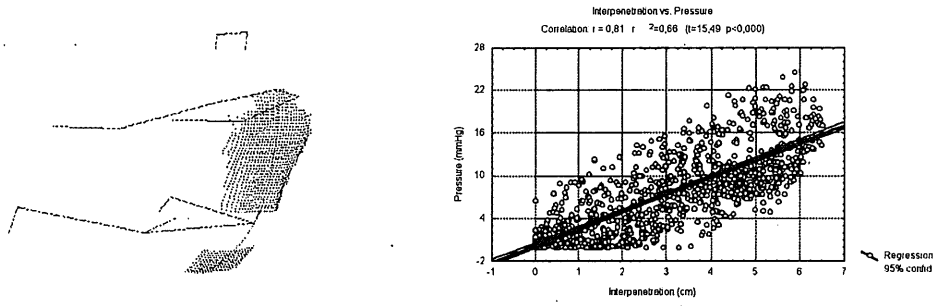



**Fig. 1: Subject and his back surface in driving posture (left) and the correlation ($r^2$=0.66) interpenetration-pressure (right).**

### DISCUSSION

Considering the pressure distribution on the back we have identified three sitting strategies: the dorsal sitting, in which there is uniform pressure at the interface body-seat; the upper scapular sitting, where the pressure peaks are placed in correspondence of the scapula; the lumbar sitting, that shows high pressure values in the lumbar region.

The role of the lateral zones both of the backrest both of the cushion was calculate as a percentage of the total load exerted on the seat region: this parameter could indicate the lateral support function which is an important characteristic for safety requirements of the vehicle.

Another result is the accurate 3-D description of the driving posture for different percentiles which allowed us for a validation of the comfort angular ranges. But if previous data in literature were taken by digitised photographs and referred only to the sagittal plane (2-D approach), in our work the anatomical angles were instead collected.

An important result is also the relationship between pressure and interpenetration that represents a predictive model for pressure distribution from the back surface. This is to be confirmed by a wider analysis and it could bring to new instruments for designing or virtual evaluation of seat prototypes.

### REFERENCES

Yamazaki, N. (1992), Analysis of sitting comfortability of driver's seat by contact shape. *Ergonomics* **35**, 677-692.

### ACKNOWLEDGEMENTS

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# FORCE PLATFORM STUDIES TO ASSES FOOT AND FOOTWEAR COMFORT

**Nachiappan Chockalingam and E Rajesh**
Gait Analysis Laboratory, Central Leather Research Institute, Adyar, Chennai 600 020

INTRODUCTION:

Different quantities that can be measured with the force platform can be used to quantity and asses the foot and footwear comfort. An attempt is made in the reported work to identify a few such parameters that could be measured using a bio-mechanical platform. The different reaction forces indicating the direction and magnitude of the resultant force vectors, the total foot contact time with the ground and the path of centre of pressure has been studied in detail. These quantities reflect the net effects of the continuously changing body configuration due to the combined effect of foot, footwear bottom components and ground interactions during normal level walking. The work reported correlate the effects of different footwear bottom components on the vertical ground reaction force within the first 20 to 40 ms of heel contact. A comparative study with the bare foot walking and values of the peak ground reaction forces at heel strike and pushoff are also reported.

METHODOLOGY:

The experiments were conducted in a Gait Laboratory equipped with a VICON movement analysis system running AMAS software. The laboratory also consisted of two AMTI force platforms embedded on a 10m walk way. Four pairs of casual footwear were fabricated and used for the trails. Of these four pairs, two had 7 mm micro cellular rubber (MCR) of 20 - Shore A as the midsole with leather and hard MCR of 65 - Shore A as outsoles. The remaining two pairs had no midsole and are outsoled with leather and hard MCR. Five test subjects with an average age of 23 years, weight of 60 kgs and a height of 173 cms performed seven trials with each of the four footwear. Data of their ground reaction forces during these trials and during barefoot walking was captured. Since the reaction force values varied with the velocity of walking, trials with a particular walking velocity from these seven trails were chosen. the corresponding analog data for this velocity was used in subsequent force value comparisons.

RESULTS:

The analysis of the barefoot ground reaction force (GRF) or the impact force within first 20ms of ground contact showed that a value of around 0.4 times the body weight (BW) on an average. Whereas the corresponding values for different combinations of outsoles are given in the table below.

| WALKING WITH | GRF VALUE (N) |
|---|---|
| 1) Leather outsoled Footwear | 0.050 BW |
| 2) Hard MCR outsoled Footwear | 0.048 BW |
| 3) Soft MCR Midsole + Leather outsoled Footwear | 0.030 BW |
| 4) Soft MCR Midsole +Hard MCR outsoled Footwear | 0.020 BW |

For barefoot walking the reaction force peak values that occurred late in the gait cycle, approximately 60 - 80 ms after heel strike is about 1.12 the body weight, whereas the corresponding values for different combination of outsole are given in the table below.

| WALKING WITH | GRF VALUE (N) |
|---|---|
| 1) Leather outsole Footwear | 1.16 BW |
| 2) Hard MCR outsoled Footwear | 1.17 BW |
| 3) Soft MCR Midsole + Leather outsoled Footwear | 1.30 BW |
| 4) Soft MCR Midsole + Hard MCR outsoled Footwear | 1.19 BW |

From the results listed, it can be noted that the vertical ground reaction force for the first 20 ms of heel strike has been reduced while walking with footwear. Footwear with soft midsole of Shore A 20 has considerably reduced the impact force by 0.37 times the body weight. While the force value for footwear without midsoles is about 0.35 times the body weight. However the vertical reaction force after 60 -80ms of heel strike has considerably increased while walking with footwear. Footwear with cushioned bottom increased the reaction force 0.10-0.18 times the body weight, while that for leather and hard MCR outsoled footwear it is about 0.04 - 0.05 times the body weight. This increase in reaction force for soft midsoled footwear may be attributed to the bottoming out effect. Bottoming out is due to the fast compression of the midsole material on application of the load during stance phase of walking. This results in the increased force peak. The other reason may be due to the different joint positions adopted by the subjects while walking.

References:

1) Bressler & Frankel, The forces and moments during level walking, Trans.Am.Soc.Mech.Engg,1950.
2) Fredfick, Biomechanics of sport shoes and playing surfaces, Human Kinetics Publishers,Illinois. 1989.

# A COMPARISON OF FORCE PLATE AND INSOLE PRESSURE SENSOR MEASURMENTS

Paul Fiolkowski, Jeff Bauer, Mark Tillman, Denis Brunt
Center for Exercise Sciencem, University of Florida, Gainesville, Florida, USA

## INTRODUCTION

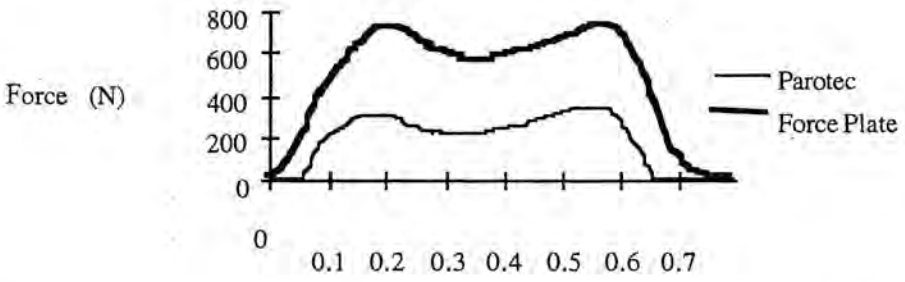
Force plates are often used in gait analysis to measure ground reaction forces (GRF). However, they are limited in their applicability to many situations due to their lack of portability. It would be beneficial to research in this area to develop a portable system to measure GRF (1). However the system must be shown to produce consistent and valid results. Therefore the purpose of this project was to compare the data produced by an insole pressure sensor system to the data measured by a force plate.

## METHODS

The equipment chosen for this comparison was the Parotec Insole pressure Sensor System ( Paromed Medizin Technik GmbH, Germany) and an AMTI force plate (American Medical Technologies Inc., Watertown Massachusetts, USA). Pressure sensitive insoles, having 24 hydrocell sensors across the plantar surface, were placed in the subjects' shoes. The insoles were connected to a module worn on a belt around the subject's waist. In order to best compare the systems, data was collected at identical frequencies of 200 Hz. The subjects, after having signed an informed consent, were instructed to walk across a force plate embedded in an 8m long runway. When they were able to consistently strike the force plate, the Parotec System was turned on and the foot strike on the force plate was noted by the investigator. The data was downloaded into a desktop computer for analysis.

## RESULTS

Floor contact time as measured by the two systems was 620 $\pm$ 41 ms for the Parotec as compared to 735 $\pm$ 57 ms for the force plate. However, these differences were consistent across all trials. The force measured by the Parotec were consistently 2 to 2.5 times lower than that recorded by the force plate.




## DISCUSSION

The Parotec Insole Pressure Sensor System provides data that is consistent with that from the force plate. That the Parotec's total force is less than that measured by the force plate results from the Parotec's sensors not contacting all areas of the foot's surface. The time shift is likely attributable to the characteristics of the insole system, which requires a threshold of 2.5N/cm$^2$ to record data. The force plate would be recording data as the foot reaches this pressure within the shoe and after the pressure in the shoe has fallen below this point. A 6$^{th}$ order polynomial regression has proven to closely fit the curves generated from the data. The consistency of the curves would seem to indicate that there is a transfer function that would allow the calculation of the GRF from the insole pressure data. Work is currently being done to determine the transfer function.

1. Cavanagh PR, Hewitt FG, Perry JE. In-shoe plantar pressure measurement: a review. *The Foot* 2:185-194, 1992

# AN ACCELEROMETER BASED POSTURE DETECTION SYSTEM

R. Toivonen, E.-P. Takala, E. Viikari-Juntura, T. Haapio
Finnish Institute of Occupational Health

## INTRODUCTION

Assessment of postural work load can be performed by questionnaires, observations, or direct measurements. The selected method should find out the main risk factors including posture, repetition, and duration. Direct measurements are often preferable because of their accuracy but their use might be impossible in the field; i.e. the apparatus may be complex or direct vision of the object is required by the optical system.

In this paper we describe a constructionally simple and cost effective posture and movement data collection system based on accelerometer recordings. Ambulatory system will be used together with surface EMG of selected muscles in the assessment of physical exposure in long term monitorings of work.

## METHODS

The system consists of four single axis capacitive accelerometers (±1.2g) connected via a signal conditioner to a data logger (ME3000P) using a sampling frequency of 10 Hz. The conditioner supplies voltage needed by the accelerometers and it linearly converts output voltage into a specific range defined by the data logger. Angles are calculated off-line by lowpass filtering acceleration data by a linear phase FIR filter and scaling results into the domain of arcsin function. Theoretical range of measurement for each channel is -90° to +90°

Sensors are mounted on the forehead, upper back (Th VII level), and both distal upper arms of the subject. On the back the accelerometer is attached directly on skin using self-adhesive tape. Other sensors are mounted using elastic straps.

It is possible to detect the elevation angle of the upper arm both in flexion/extension and abduction/adduction movements by properly positioning the accelerometers in relation to the humerus. Zero degree is detected when the arm is raised onto the shoulder level. For neck and back, angles of flexion/extension are recorded by the system.

Accuracy of the system was tested in a laboratory experiment. Three volunteers performed predefined sets of movements of the right arm, head, and back. Movements of the upper arm included both flexion/extension and abduction/adduction of different degrees using three paces (slow, normal, and fast). Movements of the head and back were made in flexion/extension direction using two distinct tempos (normal and fast). Data collected in these experiments were compared to posture data gathered simultaneously by a motion analysis system based on pulsed DC-magnetic field (Flock of Birds system) (Hirvonen et al., 1995). The precision of the measurements were judged by calculating RMS error between reference values and detected angles.

## RESULTS

For the upper arm median RMS error was 4.0° for movements both in sagittal and lateral plane. For the head median error was 5.4° and for the back the median error was 2.1°. In fast movements the detection accuracy of maximal posture angles dropped notable and range of motions got underestimated.

For slow movements and steady postures the RMS error in angle measurements was 3°. The most precise measurements were made near the 0° position of the transducers. Accuracy within the limits of 5.5° was obtained for all angles between -80° and +80°.

## DISCUSSION

Due to low sampling frequency and averaging sampling method, general inaccuracy in angle detection is expected especially in fast movements. Further, the nature of the accelerometer used by us sets certain limits to accuracy at both ends of the measurement range. To achieve accurate measurements, positioning of the transducers, especially for the arms, demands conscientiousness.

Advantages of our system are small size and low weight of the sensors, causing only minimal disturbance in the work during monitoring. Sensors are also mechanically robust and resistant to water thus being ideal for measurements in varying working environments. The chosen off-line data processing enables to keep the electrical construction of the system simple. Great advantage in work load analysis is achieved by combining the data collection with the simultaneously video recording, a feature supported by the data loggers used.

Our system, with its limitations, can be used in long term work monitorings alongside of other systems and methods to classify work postures with the reasonable accuracy.

## REFERENCES

Hirvonen M., et al. (1995) Measurement of human posture and movement with a system based on inductive sensors. In *Book of abstracts, XVth Congress of the International Society of Biomechanics.* University of Jyväskylä.

# OPTO-ELECTRONIC ANALYSIS OF IN-VITRO KNEE KINEMATIC. A PRELIMINARY STUDY.

M. VERNIZEAU*, Y. LEFEBVRE*, R. MINFELDE°, J. Y. JENNY+, F. LAVASTE*
*Laboratoire de biomécanique, ENSAM Paris, France
°AESCULAP S.A., Chaumont, France
+CTO Strasbourg, France

## INTRODUCTION

Knee joint kinematic is one of the most complex of the human body. Few authors were interested in the modifications of the normal knee kinematic induced by prosthetic implants (Clark, 1989; Nahass et al., 1991). The aim of this work was to develop and to use an experimental rig to study in-vitro tibio-femoral and patello-femoral relative displacements in normal knee or prosthetic knee during a standardized simulated movement.

## METHODS

The mobilization of bone segments is realised thanks to a rig that do not change the articular liberty. On the basis of the rig discribed by Harding (Harding et al.,1977), controlled actuators set femur and tibia in relation to the gravity direction. A 28 N vertical force P is constantly applied on the femur. The standardized simulated movement represents the transition from standing position (sagittal femoro-tibial angle = 180°, sagittal tibial-floor angle = 90°) to a sitting one (sagittal femoro-tibial angle = 90°, sagittal tibia-floor angle : 45°) then back to the initial position.

The VICON® opto-electronic system allows to record tibio-femoral and patello-femoral relative displacements by means of reflecting markers fixed on bone structures.

The same knee was first studied as a reference. A comparison was then made between this normal knee and the same knee after implantation of different prosthesis types (SEARCH® AESCULAP® implants : unicondylar prosthesis, total prosthesis with preservation of both or only posterior cruciate ligament, total prosthesis with both cruciate substitution).

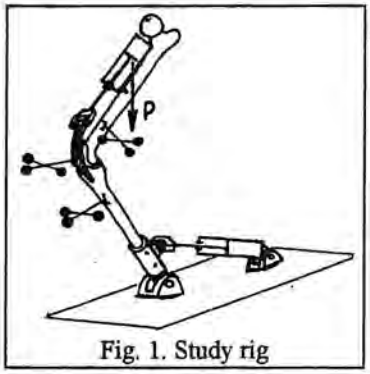


Fig. 1. Study rig

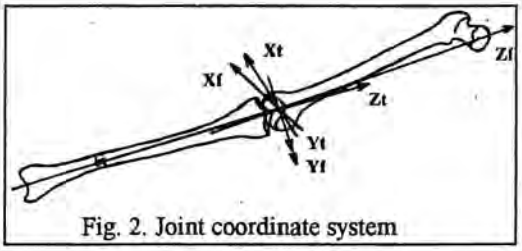


Fig. 2. Joint coordinate system

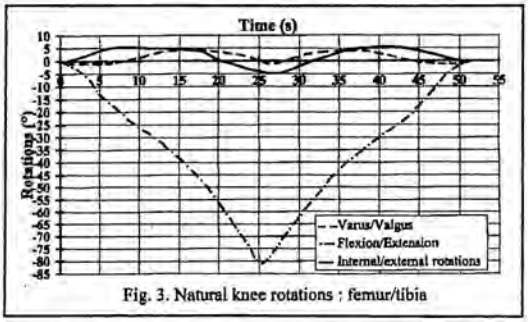


Fig. 3. Natural knee rotations : femur/tibia

## RESULTS

Typical results are described in the figure 3. The precision of measurements seems to be very accurate. A more detailed statistical study is to be performed.

## DISCUSSION

This experimental model has some inconvenients : in-vitro study, no peri-articular soft tissues, simulation of quadriceps action only. Nevertheless, in a given situation, it allows to compare different kinematics of the same knee without or with different protheses types. The modifications of the knee kinematic induced by prosthetic implants or surgical technic can then be analysed. Moreover, this rig will allow to study of the patello-femoral and tibio-femoral contact areas and pressures, or to analyse the quadriceps tendon forces.

## REFERENCES

Clark, K. J. (1989) A comparative kinematic evaluation of natural and prosthetic knees. M.S. thesis, Purdue University.

Harding, M. L., Harding, L., and Goodfellow, J. W. (1977) A preliminary report of a simple rig to aid study of the functional anatomy of the cadaver human knee joint. *J. Biomechanics* **10,** 517-523.

Nahass, B. El., Madson, M. M., and Walker, P. S. (1991) Motion of the knee after condylar resurfacing - an in vivo study. *J. Biomechanics* **24,** 1107-1117.

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# QUALITY ASSESSMENT OF THE PERFORMANCE OF A MOTION ANALYSIS LABORATORY

M Rabuffetti[1], DrEng, F Benvenuti[2], MD, R Mecacci[2], MD, S Nicolodi[3], DrEng, S Stanhope[4], PhD
**1**. Centro di Bioingegneria, Fnd Pro Juventute, Politecnico Milano, Milan, Italy. **2**. Geriatric Deparment, Ospedale INRCA "I Fraticini", Florence, Italy. **3**. B.T.S. S.r.l., Milan, Italy **4**. Biomechanic Laboratory, NIH, Bethesda, MD, USA.

**INTRODUCTION** Quality assessment of the performance of a motion analysis laboratory, composed by different instrumentation connected to give multifactorial analysis, is a fundamental issue in the daily laboratory activity since a degradation of the data acquisition system can endanger clinical conclusions. Several tests have been proposed to assure the correct operation of the equipment.

With regard to posture analysis, quality loss of acquired data depends upon: 1) items related to the motion analysis system as inaccurate calibration of TV camera stereo-pairs and not corrected optical distortion of the camera lenses; 2) the occurrence of drift on the measures of the dynamometric platform; 3) spatial coherence among the different instrumentation orthogonal reference frames, in particular an incorrect definition of the parameter of the transformation between reference frames of kinematic system and platform. Further, a clinically feasible quality assessment procedure should be fast and friendly. For this purpose, we designed a fast and reliable assessment protocol and data from a postural laboratory are reported.

**METHODS** Twenty-six quality assessment sessions were performed in 1996 in the INRCA posture analysis laboratory which uses a 4 TVCs motion analysis system (ELITE System, BTS Milano, Italy) and a piezoelectric force platform (Kistler, Winterthur, Switzerland). Kinematic and platform data were not filtered.

The assessment of kinematic system accuracy was based on the detection of two markers rigidly connected which were moved by an operator inside the calibrated working volume. The correlation of the inter marker distance and the actual position of the rod was evaluated by a linear regression analysis, in particular by the $r^2$ coefficient and the first order coefficient (null expected value for both indexes).

The assessment of platform drift required a long lasting measurement of the ground reaction of static load placed onto the platform. The expected ground reaction, constant in all its component, was compared with the observed one by means of a linear regression vs. time, computing the $r^2$ coefficient (expected null value) and the center of pressure (COP) drift (expected null value).

The spatial coherence, i.e. the correctness of the transformation between kinematic and platform reference frame, was assessed by using a rigid rod, pointed at each extremities, with eight rigidly attached markers to define its position. The rod was held vertically between a metal plate fixed onto the platform and a metal plate held by an operator and forcefully and slowly moved as describing a cone. A quasi static model of the rod has been implemented and the dynamic equilibrium considers null the sum of three items: reaction force, rod weight and manual operator exerted force. In this condition, it is expected that reaction force passes through the lower extremity and no moment of force is exerted at both virtual hinges. The reaction force direction was both measured in the platform reference frame and estimated in the kinematic reference frame via a mathematical model. Finally, two estimates of the moment of a standard reaction force (standard module of 1000 N and direction as estimated) applied at both hinges were performed in parallel. They were expected to coincide, the observed residual was assumed to be due to the transformation parameters inadequacy.

**RESULTS** *Kinematic assessment* - The observed standard deviation of the inter marker distance ranged from 1.06 to 2.04 mm; the correlation with the instantaneous position of the test object the $r^2$ coefficient ranges from -0.38 to 0.13, the incremental ratio ranges from -0.001 to 0.002. *Force platform drift assessment* - The COP drift ranged from 0.05 to 0.11 mm/minute. No linear trend depending upon time were present ($r^2$ coefficient range -0.05-0.16). *Spatial coherence assessment* - The residual between the two estimates of moment of standard force ranged from 1.22 to 4.44 Nm (these values can be alternatively assumed as moment arm differences expressed in mm).

**DISCUSSION** The proposed assessment protocol has proven to be feasible and fast. It checks for any distortion correlated to space or time, and for coherence of the relative transformation between measurement reference systems. The results show that, in the considered laboratory, no distortion significantly affects the acquisition setup, and that the system precision and spatial coherence may introduce errors lower than other possible experimental errors due, for example, to markers' placement or statistical joint models. The fortnightly test repetition ensured performance stability. The operative indication is to perform tests at the laboratory setting up phase until it reaches a desired quality level, and the to repeat them periodically to ensure the stability of the laboratory performance.

**REFERENCES**

A.Cappozzo et al., Measurement and data processing methodology ....., CAMARC II Report 1993
Brown et al., Preventing implementation and tracking errors ... , Gait and Posture 3:88, 1995.
Masiello et al., The first step towards standardisation... , Gait and Posture 3:54, 1995.
N.A.Borghese et al., An Algorithm for 3-D Automatic Movement .... , IEEE Trans. Biomed.Eng., 37:12, 1990.
S.Stanhope., A Procedure for evaluating gait clinical analysis ... , Gait and Posture 3:54, 1995.

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# GROUND REACTION ARTEFACT: IDENTIFICATION AND CORRECTION ALGORITHM

M Rabuffetti[1], DrEng, C.A.Frigo[12], DrEng
1. Centro di Bioingegneria, Fnd Pro Juventute, Politecnico di Milano, Milan, Italy.
2. Bioengineering Department, Politecnico di Milano, Milan, Italy.

INTRODUCTION Clinical gait analysis requires the application of well defined analysis protocols and, at the same time, does not allow the loss of any information about the currently observed patient, that may be severely affected by disability, unable to follow the operator instructions and to stand prolonged experimental sessions because of fatigue and pain. Therefore, incorrect acquisitions (incomplete foot contact, spurious contact of the other foot or assistive device) of the ground reaction (GR) are likely to occur: the corrupted data section is defined artefact. Repeating the trials unless the required number of correct recording has been collected is sometimes unpractical. A method for recovering the most of the information from partially corrupted data could provide greater impact to clinical application because of the improvement, quantitative and qualitative, of the experimental outcome. An algorithm is proposed for identification and correction of artefacts affecting the ground reaction measure.

METHODS The algorithm is based on the comparison of the reference GR to the artefacted GR, its preliminary requirements are that the artefact lasts less than half of the total GR duration and that a reference homogeneous (same subject and experimental conditions) GR is available.

The algorithm first step (matching) consists in identifying the transformation that, when applied to the reference GR, minimise a proper target function. The transformation has four variables: the rotation of the walking direction (1 dof), the displacement of the foot platform contact position (2 dofs) and a relative time shift (1 dof). The objective function is a global residual index consisting in the product of four indexes related to the GR components' differences: median residuals of the vertical (R_F) and horizontal (R_HF) force components, of the trajectory of the center of pressure (R_P), of the vertical torque (R_M). The R_P index is shown (other indexes are similarly defined):

$$R_P = \sqrt{median\left(\left(P_{Ax}(t) - P_{Bx}(t)\right)^2 + \left(P_{Ay}(t) - P_{By}(t)\right)^2\right)}$$

The algorithm second step (patching) consists in identifying the artefact section as the one characterised by large residuals with respect to the matched reference GR. For each GR components, the identificated section is purged and substituted by a patch. The patch is obtained by adding the not linear component of the corresponding reference GR section, to a linear component simply connecting the extremes of the purged section.

RESULTS A large GR data set (30 normal female subject, 5 trials each) has been virtually artefacted by two different approach: the first artefact simulation consisted in adding a square offset, lasting about 10% of the GR duration, in the middle of the current GR, the second artefact simulation consisted in the merging with the controlateral GR as measured by another platform (simulating the effect of a double support on platform).

The matching step has been performed, intraindividually, thrice: reference GR data set with itself, reference GR data set with the two artefacted GR data set. The total number of match is 1500. The 95th percentile values of the four indexes distribution are reported in the table.

The patching step was performed on the two simulated artefacted GR data set. The previous indexes have been evaluated with reference to the artefacted section only, the residuals among the true patterns and the estimated patches are reported in the table.

| 95th percentile (N=1500) | R_HF (N) | R_F (N) | R_P (mm) | R_M (Nm) |
|---|---|---|---|---|
| Reference | 18.6 | 31.2 | 8.3 | 8.8 |
| Internal artefact | 18.8 | 31.5 | 8.1 | 8.9 |
| Border artefact | 19.7 | 32.8 | 10.8 | 9.2 |
| Internal artefact (patch only) | 15.5 | 36.9 | 16.7 | 6.4 |
| Border artefact (patch only) | 28.2 | 105.6 | 16.8 | 12.1 |

DISCUSSION The specific features of the algorithm, no time normalisation and median based indexes' definition, are devoted to guarantee the feasibility of the matching when artefacts affect the data set. The rationale for not applying time normalisation is because one of the two extreme instant can be missing, for median based index definition is because it takes into account only the half section in which the difference is minimal, the section in which the artefact is absent. The presented results confirm the successful application of the algorithm.

REFERENCES

Brodland, G., W. et al. (1987) Gait reaction reconstruction and a heel strike algorithm. J. Biomech. 20, 767-772.
Davis, B., L. et al. (1993) Decomposition of superimposed ground reaction forces ..... J. Biomech. 26, 593-597.

# WAVELETS REPRESENTATION OF HUMAN MOVEMENT KINEMATICS SIGNALS

S. Fioretti, T. Leo and F. Verdini
Dipartimento di Elettronica ed Automatica - University of Ancona - Ancona - Italy

## INTRODUCTION

It is well known that human movement kinematics, even in the case of steady state motor tasks, is not really time invariant, neither along repeated cycles nor within a single cycle.Moreover, small details characterize important features of the movement. The time or frequency analysis of time-variant signals involves a tradeoff between accurate localization of transients and precise identification of long-term behavior. Here the time-frequency analysis technique named wavelet trasform (WT) is used for accurate and reliable analysis of human movement kinematics. WT is able to locate the local transients without loosing the long term trend of the signal. It is worth noting that the same goal is simultaneously attained also in the frequency domain. By WT one looks at the signal at different resolution or scales.
From a DSP point of view, a W. is a bandpass filter: the fine temporal analysis is done with contracted (high frequency) versions of the W., while fine frequency analysis uses dilated (low frequency) versions [1]. In this study WT is applied to the test signal [4], that represents accurately the displacement along the progression axis of the iliac crest tubercle of an amputee measured during level walking.The features of the WT are measured according to the criteria proposed in [5].

## METHODS

According to Mallat [2], the fast WT algorithm consists of decomposition and reconstruction phases. In decomposition phase, the original signal is high-pass filtered by analysis filter $h_1$ and down-sampled by two to produce the high-frequency content signal (detail signal). Similarly, the original signal is low- pass filtered by $h_0$ (e.g. $h_1(n)=(-1)^n h_0(1-n)$) and down-sampled by two to produce the low-frequency content signal (smooth signal)[3]. The low-pass version is splitted again using the same filter bank, and so on. Finally the signal is reconstructed using Symmlet wavelet at compact support with 8 vanish moments. As usual in simulation "raw" data are obtained corrupting the test signal with an additive, pseudo-white noise sequence. It is normally distribuited with zero mean and variance of 0.25 as suggested in [4] and usually done in the relevant literature.

## RESULTS

Fig A,B show the signal reconstruction using increasing number of W. The best fit has been obtained with 8 coefficients both for the noisy and for the "true" signal (Fig.C). The goodness of fit has been tested using the "$p_0$ %" parameter defined in[5]. A much better value has been obtained : 0.56 w.r.t 1.85. Moreover tests to evaluate estimates' goodness w.r.t. data present in literature[5] have been performed.

## CONCLUSIONS

The wavelet analysis allows to characterize the human movement kinematics both in time and in the frequency domains. The method resulted: (1) effective, since the reconstructed signal fits well the original signal samples and the reconstruction has good filtering capability; (2) efficient, since the signal can be represented using a small number of significant WT coefficients; (3) robust ,since the number of coefficients does not depend on noise; (4) having very low computational cost. Due to the orthogonality of the WT, the proposed representation seems candidate to allow a sound identification of the characterizing details of human movement.

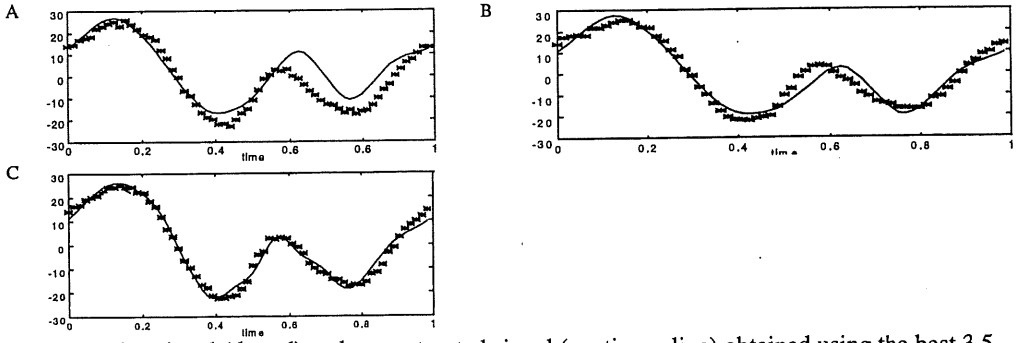



Figure: Noisy signal (dotted) and reconstructed signal (continous line) obtained using the best 3,5 coefficients (A,B respectively) and 8 coefficients (C).

## REFERENCES

1) M. Vetterli and Kovacevic, Wavelets and Subband Coding, Prentice-Hall, 1990
2) S. G. Mallat *IEEE Trans.Patt. Recog. and Mach. Intell.*, vol. 11,pp 674-693, 1989
3) M.Vetterli and C. Herley *IEEE Trans. Signal Processing*, vol. 40, No.9,1992;
4)B. J. Andrews, A. Cappozzo, *Computing in medicine*;1982
5)M. L. Corradini et al., *Med.& Biol. Eng. & Comput.*, 31,pp 187-197, 1993

# EXTRACTION OF REPEATING PATTERNS VIA OPTIMALLY WEIGHTED ENSEMBLE AVERAGING

V. P. Stokes, H. Lanshammar*, and A. Thorstensson
BMC, Neuroscience, Karolinska Institute, Stockholm, Sweden
*Systems and Control Group, Uppsala University, Uppsala, Sweden

## INTRODUCTION

The main purpose of this study was to develop a methodology for the extraction of repeating patterns from cyclic biomechanical data. This new method, singular value decomposition pattern analysis (**SVDPA**), is based on the recent work of Kanjilal and Palit [1] and can be applied to repeated trials which are not strictly periodic. It is closely related to the Kahunen-Loève transform and can extract the dominant repeating pattern even when the trajectories are nonstationary. The average energy component (**AEC**) is obtained directly from the **SVD** of a rectangular data matrix and is equivalent to the optimally weighted ensemble averaged pattern described by Unser and Eden [2] which maximizes the signal-to-noise ratio (SNR).

## METHODS

A Selspot II system with two cameras was used to obtain the trajectory of a LED attached to the trunk (L4) for a subject walking at $1.5m{\cdot}s^{-1}$ on a treadmill. Fig. 1A shows the 3D trajectory of this marker over 39 stride-cycles (SCs). Suppose that we have collected $m$-SCs with approximately the same number of samples in each SC. We replace all missing data, remove low-frequency trend (drift) and time-normalize the SCs to have the same number of samples, $p$. We stack these SCs, each of period $p$, to form the rectangular data matrix $X(m \times p)$,

$$X = \begin{bmatrix} x[1] & x[2] & \dots & x[p] \\ x[p+1] & x[p+2] & \dots & x[2p] \\ \dots & \dots & \dots & \dots \\ x[(m-1)p+1] & x[(m-1)p+2] & \dots & x[mp] \end{bmatrix}$$

where, each element is defined by $x_{ij} = x[(i-1)p+j]$. Optimally weighted ensemble averaging is obtained by finding the weights $w_i$ and offsets $b_i$ in the linearly rescaled elements of $X$; i.e.,

$$y_{ij} = w_i(x_{ij} - b_i) \quad i = 1, 2, \dots, m \quad j = 1, 2, \dots, p.$$

If the criteria for optimality is maximum SNR then it can be shown that $b_i$ is the row-$i$ average of $X$ and $w_i = u_{i1}$, the $i$th component of the left singular vector of $X$, which is related to the Kahunen-Loève transform. **SVDPA** is a general solution to multiple repeating pattern analysis which includes the optimally weighted ensemble average pattern. We call the dominant repeating pattern the **AEC** which is defined by

$$\mathbf{AEC} = \frac{\sigma_1 v_1^T}{\sqrt{m}}$$

where $\sigma_1, v_1^T$ are the dominant singular value and its right singular vector, respectively. Further, since the degree of periodicity depends on the proportion of the total energy contained in the dominant singular value, we define the coefficient of periodicity by $CP = \sigma_1^2 / \sum_{k=1}^{p} \sigma_k^2$ , $CP = 1$ for a perfectly periodic pattern.

## RESULTS AND DISCUSSION

The **AEC** was computed for each component of the 3D trajectory and the estimated dominant repeating pattern for the 3D trajectory is shown in Fig. 1B, including estimates of the uncertainty in the trajectory at each point. The coefficient of periodicity for the 3D trajectory (average CP over each component) is 0.976, a high degree of periodicity.

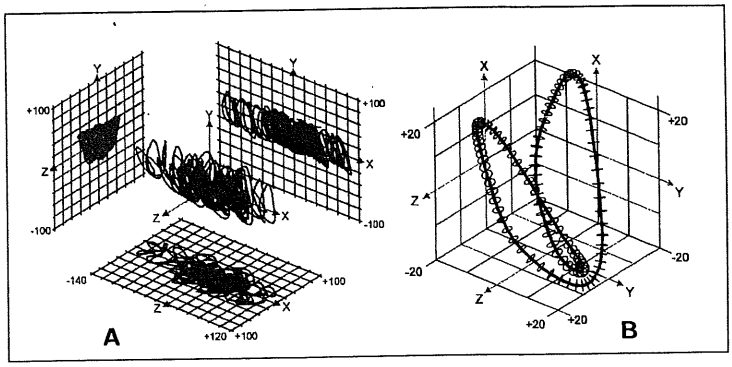



**Fig. 1. A**—Unprocessed 3D trajectory, 4300 samples at 100 samples per second. **B**—**AEC** trajectory with 1-SD error ellipses.

Conventional data analysis methods based on trigonometric basis functions may be inappropriate for cyclic biomechanical data which are often noisy, quasiperiodic and nonstationary. **SVDPA** does not use preselected basis functions and is applicable to nonstationay quasiperiodic data which makes it a useful alternative to these methods (e.g. harmonic analysis).

## REFERENCES

[1] Kanjilal, P. P. and Palit, S. (1995) On multiple pattern extraction using singular value decomposition, *IEEE Transactions on Signal Processing* **43**, 1536–1540.

[2] Unser, M. and Eden, M. (1990) Weighted averaging of a set of noisy images for maximum signal-to-noise ratio, *IEEE Transactions on ASSP* **38**, 890–895.

# EXHAUSTING STRETCH-SHORTENING CYCLE (SSC) EXERCISE CAUSES GREATER IMPAIRMENT IN SSC THAN IN CONCENTRIC PERFORMANCE

**T. Horita, P. V. Komi, I. Hämäläinen and J. Avela**
Department of Biology of Physical Activity, University of Jyväskylä, Finland

## INTRODUCTION

Exhausting stretch - shortening cycle (SSC) exercise induces delayed muscle damage which is associated with impaired muscle function as revealed by decreased stretch reflex sensitivity and/or stiffness regulation during drop jump (*Nicol et al.* 1996, *Horita et al.* 1996). Thus exhaustive SSC exercise may modify muscle function primarily during the eccentric phase. However, less information is available on the effect of SSC exercise on concentric muscle function. The purpose of the present study was to investigate the effect of SSC exercise on the concentric dynamic performance.

## METHODS

A total of 17 male subjects participated in this study. Ten of them performed SSC exercise (EX) and seven served as a controls. EX group performed repeated submaximal rebound jumps on the sledge apparatus until exhaustion similarly to our previous study (*Horita et al.* 1996). Concentric squat jumping (CSJ) measurement was performed before and immediately after the SSC exercise. CSJ measurement was repeated 10 min, 20 min, 2 days and 4 days after the SSC exercise. Maximum rebound jumping (Max SSC) measurement was also performed before, 20 min after, 2 days after and 4 days after the SSC exercise. In the present study, it was assumed that the CSJ measurement evaluates pure concentric muscle performance.

Jumping motion was recorded for the joint moment analysis. Surface electromyograms (EMG) were recorded from the VL, VM, SOL and GAS muscles for analysing integrated EMG (iEMG) activity. Blood samples were taken for the analysis of lactate (LA) concentration and creatine kinase (CK) activity.

## RESULTS AND DISCUSSION

EX group repeated 92±30 jumps which lasted 160±49 seconds. This jumping exercise induced relatively high blood LA concentration (5 min after, 7.21 $mmol \cdot l^{-1}$) and delayed increase of serum CK activity (2 days after, 486±301 $U \cdot l^{-1}$). This exercise was therefore characterized by metabolic fatigue and muscle damage. In EX group, CSJ performance (jumping height) decreased immediately after the SSC exercise ($P < 0.0001$). However, CSJ performance recovered 10 min after the exercise (Fig. 1). Corresponding iEMG activities of the knee extensors (VL, VM) during CSJ did not show any significant changes until 4 days after the SSC exercise. In contrast, Max SSC performance decreased 2 days after the SSC exercise ($P < 0.01$, Fig. 2). Both pre - activity and iEMG activity during eccentric phase decreased 2days and 4 days after the SSC exercise in the knee extensor muscles ($P < 0.01$). These results indicate that the pure concentric muscle function could recover faster after the SSC fatigue. However, SSC muscle function including pre - activity and eccentric muscle action shows long lasting impairment. It is suggested that the repeated SSC exercise could preferentially impair SSC muscle action, and that this impairment is in close connection with soreness/damage induced α-motoneuron inhibition.

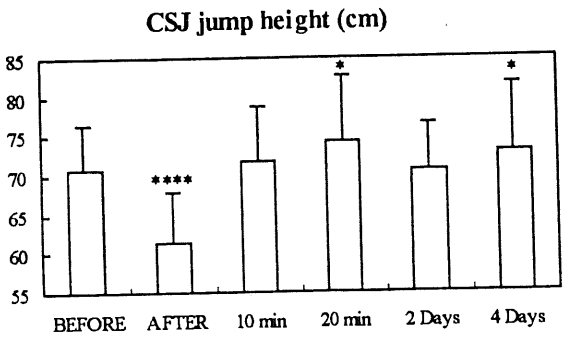



Fig. 1

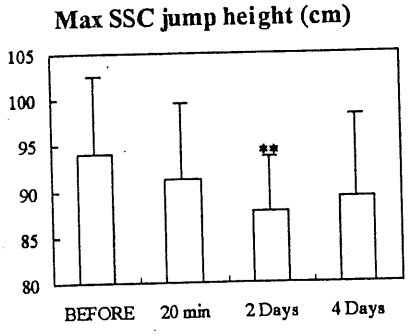



Fig. 2

## REFERENCES

T. Horita , P.V. Komi, C. Nicol, H. Kyröläinen: Stretchshortening cycle fatigue: interaction among joint stiffness, reflex, and muscle mechanical performance in the drop jump. Eur. J. Appl. Physiol. 73: 393 - 403, 1996.

C. Nicol, P.V. Komi, T. Horita, H. Kyröläinen: Reduced stretch-reflex sensitivity after exhausting tretch-shortening cycle exercise. Eur. J. Appl. Physiol. 72: 401 - 409, 1996.

# EFFECTS OF ECCENTRIC AND CONCENTRIC TRAINING ON MUSCLE STRENGTH, MORPHOLOGY AND EMG

Jan Y. Seger and Alf Thorstensson
Departments of Neuroscience, Karolinska Institute and Human Biology, University College of Physical Education and Sports, Stockholm, Sweden

## INTRODUCTION

Maximal voluntary torque output is greater in eccentric than in concentric muscle actions. Given that high muscle tension *per se* is an important training stimulus for increasing muscle strength, pure eccentric training would be the most efficient type of basic strength training. Hybrid training, i.e. combining eccentric and concentric muscle actions, has been shown to cause greater strength gains than concentric training alone (Colliander and Tesch, 1990). The purpose of this study was to investigate possible specific effects of pure eccentric and pure concentric strength training on muscle strength, morphology and myoelectric activity.

## METHODS

Two groups of five male students, equivalent in age, body measures and initial strength, performed 10 weeks of either eccentric or concentric maximal voluntary isokinetic knee extensor training (90 deg · sec $^{-1}$) for one leg, followed by another 10 weeks of the same type of training for the other leg. Training and testing of muscle strength were done on a custom-built dynamometer (Seger *et al.*, 1988). Muscle strength measurements at angular velocities of 0, 30, 90 and 270 deg · sec $^{-1}$, EMG recordings (bipolar surface electrodes) from the quadriceps muscle group, muscle biopsies (needle biopsy technique) from the vastus lateralis muscle and MRI scanning of the thigh were performed before, in the middle of, and after the entire 20-weeks period.

## RESULTS

Peak torques increased for the eccentric training group (ten trained legs) with 17.5% for eccentric actions and 2% for concentric actions (average over speeds). The corresponding values for the concentric group were 14 % and 9.5%. The largest increases were seen in the eccentric strength at 90 deg · sec $^{-1}$ in the eccentric group (35 · %), while for the concentric group the increases were evenly distributed over velocities and modes. Both training groups increased their thigh cross-sectional area by about 3%. No consistent changes were seen in muscle fibre type distribution or area. Comparisons of eccentric to concentric ratios for average torque per average EMG (electromechanical efficiency) showed that the eccentric group increased their ratios by 8, 22 and 10 % at the three velocities measured, whereas no changes were seen in the concentric group.

## DISCUSSION

Both training groups produced similar, modest, overall gains in strength. It is noteworthy that four of the five subjects in the eccentric group experienced pain in the musculature around the knee causing involuntary training interruptions or submaximal efforts during testing, whereas none in the concentric group showed similar symptoms. There was a clear mode and velocity specificity in the effects of eccentric as opposed to concentric training. The eccentric/concentric torque/EMG results indicate that the eccentric muscle actions became relatively more efficient than the concentric actions in the eccentric training group, whereas the efficiency was unaffected by concentric training. Increases in thigh cross-sectional area were rather small and unspecific, which is in accordance with other training studies of similar duration and intensity (Colliander and Tesch, 1990). Hence, the results of the present study supported the hypothesis on speed and mode specificity only for eccentric training. However, there was no indication of a general superiority of eccentric actions over concentric ones regarding strength gain or muscle growth.

## REFERENCES

Colliander, E. B. and Tesch, P.A. (1990) Effects of eccentric and concentric muscle actions in resistance training. *Acta Physiol Scand* **140,** 31-39.

Seger, J. Y., Westing, S.H., Hanson, M., Karlson, E. and Ekblom, B.(1988) A new dynamometer measuring concentric and eccentric muscle strength in accelerated, decelerated or isokinetic movements. *Eur J Appl Physiol* **57**, 526-530.

# FATIGUE EFFECTS OF DIFFERENTLY INTENSIVE EXHAUSTIVE STRETCH-SHORTENING CYCLE EXERCISES

V. Strojnik[1] and P.V. Komi[2]

1 University of Ljubljana, Faculty of Sport, Ljubljana, Slovenia
2 University of Jyväskylä, Department of Biology of Physical Activity, Jyväskylä, Finland

## INTRODUCTION

There are numerous sites between motor cortex and muscle contractile mechanism where fatigue may occur. According to the site, fatigue may be divided into central and peripheral one, the later further subdivided into high- and low-frequency fatigue. A fatigue development during intermittent maximal intensive stretch-shortening cycle (SSC) exercise undergoes different phases, starting with a high-frequency fatigue and ends with a low-frequency fatigue (Strojnik and Komi, 1996). The aim of the present study was to characterize the appearance of fatigue after differently intensive exhaustive stretch-shortening cycle exercises.

## METHODS

Ten students of physical education (age: 28.1 $\pm$ 5.8 yrs, height: 179.8 $\pm$ 5.3 cm, mass: 78.9 $\pm$ 12.3 kg) volunteered for the study. They performed jumps on a special sledge apparatus (Komi et al., 1987) at different intensities (controlled as a percent of a maximal jumping height) until exhaustion. At the maximum intensity (MAX), the jumps were performed with the maximum possible effort. When intensity dropped under 90%, the workout was interrupted. The second intensity was set to 80%. The subjects jumped at that prescribed height (feed-back information was provided) until exhaustion which was defined as inability to maintain the target height. The third intensity was set to 60% and was carried out in the same manner as 80%. Before and one minute after the workout the following tests were performed: an explosive maximal voluntary knee extension (MVC), a supramaximal double twitch (DTW) of relaxed quadriceps femoris muscle (QF), and a knee torque during 20Hz (F20) and 100Hz (F100) electrical stimulation of the vastus lateralis muscle. A ratio (LHF) was calculated according to formula F100/F20*100-100. All tests were performed in isometric conditions.

## RESULTS

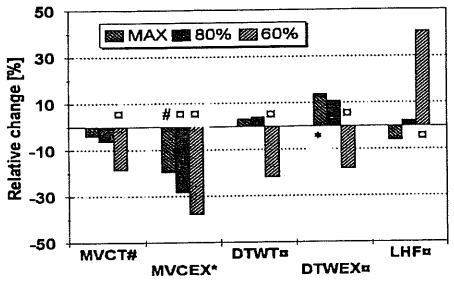



Figure 1. Relative changes in observed parameters after the workouts of different intensities according to formula after/before*100-100. MVCT - maximal voluntary knee extension torque, MVCEX - maximal rate of torque development during MVC, DTWT - maximal torque during DTW, DTWEX - maximal rate of torque development during DTW. *, #, ¤ denote statistical significance (P<.05, P<.01, P<.001, correspondingly).

## DISCUSSION

The most intresting results were found in tests with electrical stimulation, implying important qualitative differences in fatigue mechanisms among fatigue protocols. Events after MAX workout were dominantly characterised by a high frequency fatigue. The contractile part of QF was potentiated as denoted by a stronger and shorter double twitch. At the same time, the increase in LHF implies impaired action potential propagation over the sarcolema building up a high frequency block. Since during explosive contractions the action potentials are delivered with much higher frequency than during attaining maximal force level, this might be an explanation for the fatigue events after maximal intensity. After 80% workout, reduction in the maximal isometric voluntary contraction was more pronounced as at MAX. However, the other parameters did not follow the same trend. After 60%, MVCF and MVCEX were both together significantly impaired for the first time. The reason may be mainly contributed to a low-frequency fatigue. It was possible to conclude that fatigue after differently intensive stretch-shortening workouts differed not only quantitatively, as noted from maximal isometric voluntary contraction test, but also qualitatively, revealing different fatigue mechanisms as observed in intermittent maximal intensive SSC workout.

## REFERENCES

Komi P.V., Kaneko M., Aura O. (1987) EMG activity of the leg extensor muscles with special reference to mechanical efficiency in concentric and eccentric exercise. Int. J. Sports Med., 8: suppl 22-29.

Strojnik V., Komi P.V. (1996) High-low frequency fatigue during prolonged stretch-shortening cycle exercise. In: Abstract book, International Pre-Olympic Congress, Dallas, pp 119.

# APPLICATION OF NEUROMUSCULAR ELECTRICAL STIMULATION IN DEVELOPING ROWERS' MUSCLES

Dongsheng Guo, Chunying Sun*, Yue Li, Haiying Guo, Kaosheng Jin
Zhejiang Institute of Sports Science
*Zhejiang College of Traditional Chinese Medicine

## INTRODUCTION

Rowing techniques can be devided into three categories by the characteristics of F-t curves. Category A Emphasises the force of legs, Category B pays more attention to the cooperation of upper body and arms. Category C emphasises the beginning of the rowing and neglect the middle and the ending of the cycle. Different kinds of rowing style is the result of different habit and training, but they also reflect the condition of specific muscle groups of each individual. Neuromuscular electrical stimulation(NMES) is developed from clinical electrical stimulation. Many researchers are interested in its applications in sports training. Pan Huiju et al. found out that NMES has many advantages compared with direct muscle electrical stimulation. So we applied NMES to the rowers as a supplemental training method.

## METHODS

20 elite male rowers in Zhejiang Rowing Team volunteered to take part in this study. The subjects were devided into two groups, group A received NMES from NMES-168 Stimulators every night plus normal training every day for two weeks, and the muscle groups that received NMES were decided by the features of their F-t curves measured with a strain gauge set in a Concept- II rowing ergometer. Group B was trained normally. Heart rates were monitored during the stimulations. Before the stimulations , two tests were done. Test 1 is to measure the blood lactate acid concentration (BLAC) 2 minutes after the 20 mins rowing with intensity of 1:50/500m and rowing frequency of 18 on a Concept - II rowing ergometer. 10 mins after Test 1. Test 2 measured the time the rowers take to finish 2500m on a Concept-two rowing ergometer. After two weeks of stimulation, the F-t curves, Test1 and Test2 were done again. t tests were done on all the measurements before and after the experimental training period.

## RESULTS AND DISCUSSION

The F-t curves of both Group A and Group B were generally in better form after two weeks of intensive training. In Test1, the average BLAC of Group A decreased from 5.38mmol/l to 4.27 mmol/l, and the average BLAC of Group B decreased from 5.32 to 4.89 after the experiment. In Test2, the average time for rowing 2500m of Group A decreased from 8:20 to 8:03, and the average time for rowing 2500m of Group B decreased from 8:21 to 8:11. T tests showed that that all the rowers had significant progress after two weeks of conscious training, and that Group A had greater improvement. After measuring the Force -time curves, the rowers learned the weaknesses of their technique, and then trained with a definite aim to reach the best model. The improved technique of the rowers caused the decreament of BLAC after the same amount of training. Rowers of Group A not only know the features of their technique through F-t measurement, but also adopted NMES to the weak part of the human muscles besides normal training. That is why they got better results after two weeks of experimental training. During the stimulations, the heart rates of the subjects increased by 10-15%, which were much smaller than those in general strength training.

We concluded that by measuring F-t curve we can train the rowers according to the characteristics of each rower, and that NMES-168 played a positive role in the development of muscles of elite rowers.

## REFERENCES

Ulrich Hartmann et al., Intl. J. Sports Med., 1993 1:42-45
Ma Zheming et al., J. of China Sports Med., 1990,1:32-35
Selkowitz, D. Z., Am. J. Sports Med., 1989, 1:103-111
Stefanovska, A. et al., J. Rehabil Med., 1985, 3:141-144
Pan Huiju et al., J. Zhejiang Sports Sci. 1991, 3:50-54

# SHORT AND LONG TERM EFFECTS OF RHYTHMICAL NEUROMUSCULAR STIMULATION ON THE FLEXIBILITY

J. Künnemeyer and D. Schmidtbleicher
Institute of Sport Sciences, Johann Wolfgang Goethe-University Frankfurt, Germany

## INTRODUCTION

The rhythmical neuromuscular stimulation (RNS) is a rather new stretching technique which was developed by NASAROV and offered 1990 in Austria.
Additionally to a stretching exercise, an RNS-stimulator applicates small mechanical vibrations to the subject so that the muscle is frequently longitudinal stretched (23Hz). This method should cause many positive effects on the neuromuscular system. In the manual of an RNS-stimulator NASAROV emphasizes the developement of flexibility and the warming-up effect of the RNS. It is asserted that in rehabilitation improvements on the range of motion (ROM) of the shoulder of 30-40° are observed. The aim of this study was to check the short and long term effects of the RNS on flexibility in comparison to a popular static stretching method (SS).

## METHODS

Data were obtained by two studies. The short term effect of RNS on flexibility of calf muscles were checked by electrogoniometric measurement of active ankle dorsiflexion and H-reflex for neuromuscular interpretation. 12 healthy, male students of Physical Education (mean age: 26.8 years) stretched the calf muscles two minutes with alteration in stretching intensity every ten seconds. This procedure had to be done three times each side.
The tests of flexibility were done before, directly after the stretching procedure and after a rest of 30 minutes. The H-reflex amplitude of m.gastrocnemius medialis was recorded before RNS application and was observed every two minutes after stretching for a period of 30 minutes.
The second investigation treated the effects of an 8-week stretching program with RNS and SS on the flexibility of the hamstrings. 26 healthy leisure sportsmen (n=17) and -woman (n=9) (mean age: 30,88 years) were stretching their hamstrings at least two times per week by RNS (n=12) respectively by SS (n=14) for 3x15s each leg. The flexibility was measured before and after stretching by ROM of hip-flexion at sit&reach-test.
The statistical analysis was done by t-test and ANOVA.

## RESULTS

The influence on the range of motion at dorsiflexion is not significant. The obtained averaged changes are lower than 2°.
The H-amplitude shows a reduction in motoneuron excitability of 37% which doesn't recover completely after 30 minutes.
The flexibility of hamstrings increases significantly after 8 weeks ($p<0.001$). The mean increase in ROM was 5,2cm (RNS) and 4,8cm (SS). There is no statistical difference in improvement differing between RNS and SS. The direct stretching effect is significant at 5% level with mean increase in ROM of 4,0cm and is assigned to both strecthing techniques.

## DISCUSSION

The obtained results in improvement of hamstring flexibility correspond with further studies by other investigators [e.g.2]. Concerning the measurement of flexibility of calf muscles it probably can be blamed on passive insufficiency of m.tibialis because of the joint capsule in ankle.
While stretching the calf muscles with RNS the EMG shows rhythmical activation of m.gastrocnemius wich can be proved as stretch reflexes. A depression of spindle Ia-endings can be expected. Moreover, the depression of H-reflex shows the influence of vibration to the $\alpha$-motoneurone pool. The smaller H-response reflects a lower excitability. Additionally it is known that vibrations leads to a desensibilization of the $\gamma$-system [1]. This relaxation in neuromuscular system probably benefit stretching effects especially in rehabilitation.
The role of connective tissue and its reaction to rhythmical mechanic stimulation is discussed in another paper.

## REFERENCES

[1] Fromm,C./Noth,J.: Reflex responses of gamma motoneurones to vibrations of the muscle they innervate; Journal of Physiology, 256, 1976, 117-136
[2] Hartley-O'Brien,S.J.: Six mobilisation exercises for active range of hip flexion; Research Quaterly for Exercise and Sport 51, 4, 1980, 625-635

# NEUROMUSCULAR ADAPTATIONS DURING STRENGTH TRAINING IN MIDDLE-AGED AND ELDERLY MEN AND WOMEN

K.Häkkinen[1], M.Kallinen[2], M.Alen[2], M.Izquierdo[3],K.Jokelainen[4],H.Lassila[4],E.Mälkiä[4],W.Kraemer[5] and R.Newton[6]
[1]Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, [2]Peurunka-Medical Rehabilitation and Physical Exercise Centre, Laukaa, Finland, [3]Institute of Physical Education of León, León, Spain, [4]Department of Health Sciences, University of Jyväskylä, [5]Center for Sports Medicine, the Pennsylvania State University ,USA,[6]Centre for Exercise Science and Sport Management, Southern Cross University, Australia.

## INTRODUCTION

Human muscle strength decreases with increasing age, especially at the onset of the sixth decade in both sexes. This is related to a great extent to the decrease in muscle mass due to a reduction in the size and/or a loss of individual fibres, especially of fast twitch types, since aging is associated with alterations in hormone balance and with a decrease in the intensity of daily physical activities. However, strength training in middle-aged and elderly people can lead to substantial increases in their strength mediated during the initials weeks of strength training by considerable neural adaptations (Moritani & DeVries 1980, Häkkinen 1994). Further information is needed to what extent the increase in strength in the elderly may be accounted for by changes in the quantity and/or quality of activation so that voluntary activation of the agonists is increased and/or coactivation of the antagonists is reduced (Carolan&Cafarelli 1992). Skeletal muscle of elderly people seems to retain the capacity to undergo training-induced hypertrophy, when the intensity and the duration of the training are sufficient (Frontera et al.1988). This study was designed to examine neuromuscular adaptations in middle-aged and elderly men and women during a strength training period of six months with special interest in the function of both leg extensor and flexor muscles.

## METHODS

A group of forty-two healthy men (M) and women (W) divided into two age groups, i.e. M40 yrs (range 40-44; n=10), W40 yrs (range 37-42; n=11), M70 (range 69-75; n=11) and W70 (range 64-70; n=10) volunteered as subjects by signing a written consent form prior to participation.The study was conducted according to the declaration of Helsinki and was approved by the Ethics Commitee of the University of Jyväskyla. Strength training for all major muscle groups took place two times a week for six months using progressively increasing loads of 40-80% of 1RM. The movement velocities in the exercises for the legs varied from medium to maximal (heavy resistance and explosive strength training). Muscle cross-sectional area (CSA) (Aloka Fansonic,SSD-190), maximal isometric force and force-time curve (rate of force development; RFD) and concentric 1 RM of the leg extensors and maximal isometric knee flexion force (David) as well as electromyographic activity of the VL,VM and BF muscles of both legs (by surface electrodes) were recorded.The measurements were taken one month before (-1), immediately before training (0) and at two month intervals (2,4,6) during the 6-month training period.

## RESULTS

Maximal forces remained unaltered during the control period (from -1 to 0). Great increases took place during the training in isometric extension force in M40 from 2296±92 to 3102±133 N ($p<0.001$), M70 from 1799±155 to 2468±239 N ($p<0.001$), W40 from 1335±113 to 2186±221 N ($p<0.001$) and in W70 from 1104±139 to 1682±204 N ($p<0.001$). The RFD values increased in all groups ($p<0.05$-0.01). Each group showed great increases ($p<0.05$-0.001) in the IEMG of the VL and VM muscles of the isometric actions. The 1RM values improved ($p<0.001$) by 21-34 % with increased ($p<0.05$-0.001) IEMG of the VL and VM muscles in all groups. Knee flexion forces increased ($p<0.05$-0.001) by 14-22 % with increased ($p<0.05$-0.01) IEMGs of the BF muscles in all groups. The BF activity during the isometric extension remained unaltered in M40 and W40 but decreased in both M70 (from 24±6 to 21±6 %; $p<0.05$ for the left leg) and W70 (from 31±9 to 24±4 %; $p<0.05$ for the right leg). The CSA of the leg extensors increased in M40 from 53.3±1.5 to 56.2±2.5 $cm^2$ ($p<0.05$), M70 from 43.7±1.9 to 44.6 ±1.9 $cm^2$(ns.), W40 from 39.7±1.8 to 43.4±1.8 $cm^2$ ($p<0.01$) and in W70 from 33.8±1.0 to 35.7±1.2 $cm^2$ ($p<0.05$).

## DISCUSSION

The findings show that progressive prolonged strength training with only two sessions a week can lead to great increases in both maximal isometric and dynamic strength accompanied by improvements also in explosive force characteristics of the trained muscles not only in middle-aged but also in elderly men and women. The increases in muscle strength characteristics during the present training were accompanied by considerable neural adaptations so that the maximal voluntary activation of the agonist muscles increased to a great extent in all subject groups. The training led also to some reduction in the coactivation of the antagonist muscles in the present elderly men and women to about the same level recorded for the middle-aged subjects. The enlargements observed in the CSA of the trained muscles indicate that skeletal muscle of elderly people retains the capacity to undergo training-induced hypertrophy. However, the degree of muscular hypertrophy caused by the present heavy resistance/explosive training regimen seemed to be in both age groups of both genders much smaller than the role of neural adaptation.

## REFERENCES

Carolan, B., Cafarelli, E. J. Appl. Physiol. 73, 3, 911-917, 1992.
Frontera, W. et al. J. Appl. Physiol. 64, 1038-1044, 1988.
Häkkinen, K. Crit. Rev. Phys. Rehab. Med. 6, 2, 161-198, 1994.
Moritani,T., DeVries, H. J. Geront. 35, 672-682, 1980.

## NEUROMUSCULAR FUNCTION AFTER MARATHON

**T. Pullinen, M. Leynaert and P.V. Komi**
Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

### INTRODUCTION

Marathon running has been shown to induce dramatic acute reductions in maximal isometric voluntary contraction (MVC) and in maximal integrated EMG of the muscles, while neural activation at a given submaximal force level may even increase (Nicol et al. 1991). The purpose of this study was to investigate the recovery rate of the neuromuscular function after a marathon run.

### METHODS

Seven endurance trained athletes (6 men and 1 woman) performed explosive MVCs with knee extensors 7 days before, just before and shortly after an individually run marathon race, as well as 2 days, 4 days and 6 days after the run. The MVCs were always followed by a step-wise isometric endurance test until exhaustion ($ISOM_{STEP}$ ; each contraction 10 s, first step 10% MVC, increases 10% MVC with 5 s recovery between the steps). EMG was measured in each test with bipolar surface electrodes bilaterally from the vastus lateralis, vastus medialis and rectus femoris muscles. Average EMG (aEMG) and median frequency (MF) were analyzed for each force level and for each muscle. Blood lactate (La) and serum creatine kinase activity (CK) measurements were done on each test occasion. Muscle soreness was measured with a questionnaire (scale; 0 = no soreness, 5 = intolerable soreness).

### RESULTS

The subjects ran the race in 3h19min ± 21min. The results showed significant reductions in MVC and in the mean maximal aEMG of all the studied muscles (aEMG) after the marathon. Both MVC and aEMG recovered slowly and in parallel during the following six days (Fig. 1).

In $ISOM_{STEP}$ the mean terminal force level stayed fairly constant (78.8 - 83.1 % of MVC). The aEMG/force ratios tended to increase after marathon in $ISOM_{STEP}$ at a given force level but they returned to the pre-marathon level already two days after the race. No significant changes in MF in MVC or in $ISOM_{STEP}$ could be observed during the follow-up. CK-activity increased by 548 ± 284 % on the first two recovery days. Muscle soreness was highest shortly after the race (4 ± 1) and still high two days after the race.

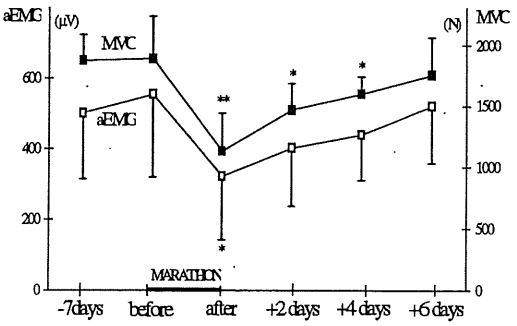



*Fig. 1. aEMG and MVC (n = 7, * p <0.05, ** p < 0.01, statistically significant difference from - 7 days).*

### DISCUSSION

The marathon race reduced dramatically both MVC and aEMG. These reductions recovered slowly and in parallel, so that no significant changes in aEMG/force -ratio could be observed. These data would suggest that insufficient neural input to the muscles was mainly responsible for the decreased MVC in these subjects. There are several possible explanations for the impaired neural drive: motoneuronal adaptation to a constant excitatory input, reflex inhibition via group III and IV muscle nociceptors and reflex disfacilitation by a decline in the muscle spindle excitatory input (Gandevia 1992).

The increase in aEMG/force -ratio in $ISOM_{STEP}$ at a given force level after the marathon may be seen as a neuromuscular compensation to a rather acute failure in the excitation-contraction coupling, since the aEMG/force -ratio had already returned to the pre-marathon level two days after the race, although high increase in CK and muscle soreness still indicated deterioration in muscle contractile apparatus.

### REFERENCES

Gandevia, S.C. (1992) Some central and peripheral factors affecting human motoneuronal output in neuromuscular fatigue. Sports Medicine 13, 2, 93-98.

Nicol, C, Komi, P.V. and Marconnet P. (1991) Fatigue effects of marathon running on neuromuscular performance. II. Changes in force, integrated electromyographic activity and endurance capacity. Scand J Med Sci Sports, 1, 18-24.

# MOTOR UNIT RECRUITMENT AND RATE CODING DURING VOLUNTARY LOW FORCE SHOULDER ABDUCTION

B.R. Jensen, M. Pilegaard, and G. Sjøgaard.
Department of Physiology, National Institute of Occupational Health, Denmark.

## INTRODUCTION

It has been shown that muscles compensate for the development of fatigue during a task of 30° arm abduction for 30 min by recruitment of new and larger motor units (MU) and by rate coding. Recruitment of MU's was indicated by increased MU action potential (AP) amplitudes and rate coding was primary seen as an increased firing rate variability (1). MU recruitment and rate coding are also involved in force regulation of non-fatigued muscles. However, neither MU recruitment as well as rate coding relative to specific isometric force production nor the chronologic occurrence of the 2 phenomena relative to each other during dynamic contractions and ramp contractions have been studied in details.
The aim was to study MU recruitment and rate coding at two different isometric submaximal contraction levels in a non-fatigued supraspinatus muscle and to relate the findings to MU recruitment in the fatigued muscle (1). Further, to study the significance of MU recruitment and rate coding during isometric ramp contractions and dynamic contractions.

## METHODS

Seven healthy female subjects, 30(24-36) years, performed in the sitting position 4 different bilateral tasks: isometric arm abductions at 11 %MVC (30° abducted arms) and 30 %MVC (vertical upper arm), dynamic contractions (abduction 0°-30°/adduction 30°-0°, cycle time 4 s), and isometric ramp abductions with vertical upper arms (15 %MVC/s from 0 to 30 %MVC with resting periods of 1 s between ramps). Motor unit activation was studied in m. supraspinatus using quadripolar needle electrode recordings with a duration of 8 s each and a decomposition technique for the analysis. Increase in average MUAP amplitude indicates recruitment of larger MU.

## RESULTS

During the isometric contractions MUAP amplitude averaged 376(SE 30)μV at 11% MVC (63 MU, n=6) and 593(SE 67) μV at 30 %MVC (29 MU, n=6) in the non-fatigued muscle, which is in line with the Henneman size principle. Assuming a linear relationship between average MUAP amplitude and force from 11 to 30 %MVC it was calculated that MUAP amplitude at 26 %MVC corresponded to the MUAP amplitude previously measured at 11 %MVC in the fatigued muscle (547(SE 39) μV) (1). Mean firing rate averaged 11.2(SE 0.2) Hz at 11 %MVC and 17.3(SE 0.9) Hz at 30 %MVC in the non-fatigued muscle. Thus, isometric muscle force can be adjusted efficiently by few Hz changes in firing rate.
During the dynamic contractions 19 MU (n=4) with an average MUAP of 512(SE 89) μV were identified. 14 MU fired continuously throughout the 8 s recording although shoulder torque varied 4 times between 0 and 11 %MVC. A significant rate coding was seen in all identified MU's and varied from 5.8(SE 0.8) Hz in a position with vertical arm to 15.3(SE 0.4) Hz in the position with a 30° abducted arm. Thus, the average mean firing rate was higher in the 30° abducted position during the dynamic contractions than during the isometric contraction.
As during dynamic contractions a significant rate coding was seen during ramp contractions (45 MU, n=5) and 15 of the identified MU's fired continuously during the recording period.

## CONCLUSION

The present study indicates that rate coding account for the largest part of the force regulation at low force contractions of the non-fatigued muscle whereas in the fatigued muscle recruitment of larger MU's account for the largest part of the compensation to maintain force.

## REFERENCE

1 Jensen B.R., Pilegaard M., Sjøgaard G. Electromyographic changes at motor unit level during fatiguing contraction and recovery. XVth Congress of the International Society of Biomechanics 1995; 432-433.

# DYNAMIC POSTURAL STABILITY OF THE SHOULDER

**Leonard A. Rozendaal**

Dutch Shoulder Group, Faculty of Human Movement Science,
Vrije Universiteit, Amsterdam, The Netherlands.

**INTRODUCTION** For a satisfactory performance, the musculoskeletal system must not only show equilibrium of forces and moments, but this equilibrium must also be stable. This means that after the application of transient disturbances, the system must return to its equilibrium position, or to its nominal trajectory in case of a movement. A quasi-static analysis is useful to get a preliminary insight into the effect that muscle and other forces have upon the stability of the musculoskeletal system. This analysis shows that often the contraction of the muscles of the shoulder will have a destabilizing effect on the shoulder joint. The reflexive control system may be used to counteract these destabilizing influences and to achieve stabilisation of the system. Also the effective properties of the musculoskeletal system, i.e. its stiffness and viscosity, can be modified by reflexive muscle control. As the reflexive control mode is essentially dynamic, with significant time lags occurring between a change of the joint angle and the resulting reflexively induced force change, a dynamic analysis is required to assess its influence.

**METHODS** A model of simplified geometry is used, the shoulder being represented by a single hinge joint. The model includes the requisite dynamic elements (passive arm properties, muscle activation dynamics, sensory time delays) and a control structure that resembles the most simple level of reflexive control (proportional feedback of the muscle length, velocity and active force). By optimizing the feedback gains for the three sensory modalities, the behaviour of the musculoskeletal system can be adjusted within the limits that are imposed by the inherent system properties.

**RESULTS WITHOUT MUSCLE VISCO-ELASTIC PROPERTIES** In the model, a marked interaction is present between the intrinsic visco-elastic properites of the muscles and the reflexive control. In first instance the visco-elastic properties are neglected and the effectiveness of the reflexive control alone is investigated. To increase the joint stiffness, the length feedback gain must be increased, but this is only useful as long as the reflexive control system can generate sufficient damping at the same time. Otherwise the result is a stiff but underdamped or even an unstable system. Damping can be achieved by sensory feedback of the muscle velocity. However, the dynamics of the sensory feedback path (time delays and muscle dynamics) filter the velocity feedback signal and limit its effectiveness. The feedback of active muscle force can be used to effectively speed up the dynamics of the muscle. As a result, the length-and-velocity feedback becomes more effective and the velocity feedback can generate more viscosity, hence the length feedback can be increased and the maximum obtainable joint stiffness increases. To obtain the optimum performance of the model, the force feedback gain must have a maximal value (while avoiding excessive resonance in the force feedback loop). For the combination of feedback gains which results in the optimum system behaviour, the ratio between velocity and length feedback gains approximately amounts to 0.10, which is consistent with experimental data concerning the dynamic behaviour of the muscle spindle Ia afferent. The maximum stiffness at an acceptable level of damping equals 28 Nm/rad with force feedback, but only 12 Nm/rad without force feedback. The former value is in the range of experimentally determined values, but lower than the highest reported shoulder stiffness magnitude.

**RESULTS INCLUDING MUSCLE VISCO-ELASTIC PROPERTIES** Inclusion of the intrinsic muscle properties into the model increases the maximum model stiffness to realistic values, and at the same time gives a fundamental insight concerning the interaction between the intrinsic muscle properties and the reflexive control of the muscles. When the intrinsic stiffness increases while the intrinsic viscosity remains constant evidently the total joint stiffness increases as well, but less than expected at first sight. The presence of the intrinsic stiffness reduces the effectiveness of the reflexive control, and the intrinsic stiffness partly substitutes the stiffness of reflexive origin. An increase of the intrinsic viscosity at a constant level of the intrinsic stiffness also results in an increase of the maximum joint stiffness, but by a completely different mechanism: due to the added viscosity the reflexive length feedback gain can be increased, resulting in a higher maximum joint stiffness. Both effects are of the same order of magnitude. It appears that the destabilizing effect of muscle contraction that is found in a quasi-static analysis of the system, in the dynamic model can be compensated for by the reflexive control to a large extent (more than 60 per cent). This, in combination with the effect of the muscle intrinsic properties, is quite sufficient to ensure a stable system with useful properties.

**DISCUSSION** However strongly simplified, the model gives a useful insight into the effectiveness of the reflexive control system in controlling the behaviour of the musculoskeletal system, and its interaction with the muscle visco-elastic properties. Such insight can be very useful in interpreting experimental results and the results of large-scale musculoskeletal models, where the present in-depth analysis is no longer possible.

**REFERENCES** Rozendaal LA (1997) Stability of the shoulder, intrinsic muscle properties and reflexive control. PhD thesis, Delft University of Technology, The Netherlands.

# QUASI-STATIC POSTURAL STABILITY OF THE SHOULDER

**Leonard A. Rozendaal**

Dutch Shoulder Group, Faculty of Human Movement Science,
Vrije Universiteit, Amsterdam, The Netherlands.

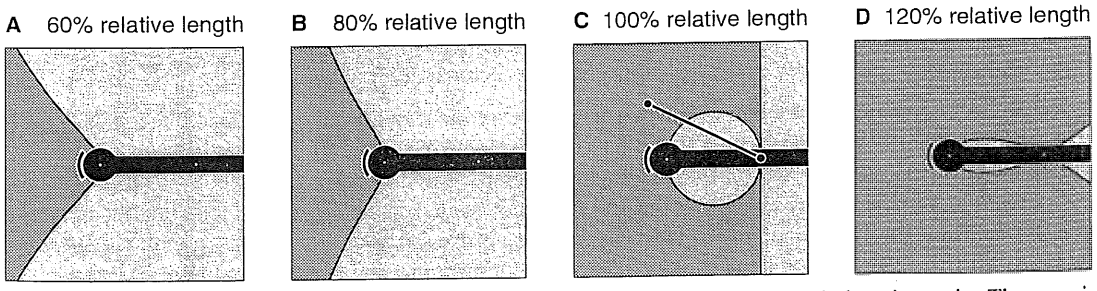
**INTRODUCTION** In large-scale musculoskeletal models usually the muscle forces are computed under two conditions: 1) equilibrium of forces and moments; 2) an assumption concerning the use of the muscles, e.g. energy-efficiency or minimum movement duration. The requirement that the model must be stable as well, i.e. that it must return to its desired position or nominal trajectory after small transient disturbances, generally is not taken into account. However, this requirement is essential for an adequate behaviour of the system. In this paper the effect of forces on postural stability is investigated in a simple model, for later use in large-scale musculoskeletal models.

**QUASI-STATIC POSTURAL STABILITY** The model contains a single hinge joint. In the quasi-static context, the change of the joint moment with a change of the joint angle determines the stability of the system. If the moment increases with the joint angle ($dM/d\phi > 0$), the additional moment reinforces the angle deviation and the system is unstable. If the moment decreases with the joint angle ($dM/d\phi < 0$), the moment change counteracts the change of angle and the system is stable (quasi-statically speaking). The moment $M$ that a force exerts is equal to the moment arm $r$ times the force magnitude $F$ : $M = r(\phi)\, F(\phi)$. Both the moment arm and the force magnitude may depend on the joint angle $\phi$. Hence the stability-determining term $dM/d\phi$ can be expressed in two terms as well:

$$\frac{dM}{d\phi} = \frac{dr}{d\phi} F + r \frac{dF}{d\phi} .$$

The first term is called the *force contribution to stability*, as it depends on the magnitude $F$ of the force. The second term is called the *stiffness contribution to stability*, as it depends on the change of the force magnitude with the change of the joint angle, i.e. on the angular stiffness of the force-generating structure. Both these stability contributions depend on the geometry of the force-generating structure as well, as comprised in either $dr/d\phi$ or $r$.

**PARTICULAR CASES** For particular cases, the force and/or stiffness contribution takes a special form. For example, the magnitude of gravity does not depend on the joint angle ($dF/d\phi = 0$). The remaining force contribution to stability acts stabilizingly for elevation angles below 90°, and destabilizingly for elevation angles larger than 90°. For muscles with a constant moment arm $dr/d\phi$ is equal to zero, and only the stiffness contribution remains, either from the intrinsic muscle properties or from the reflexive control of the muscles. For the common straigth-line geometry of point-to-point muscles, the force contribution to stability can be determined analytically. The result is summarized in in Fig. C. When the fixed-world attachment site of the point-to-point muscle lies within the light area, the force contribution is stabilizing, and when it is in the dark area (as for the indicated muscle) the force contribution is destabilizing. For shoulder muscles in general the force contribution will be destabilizing. The stiffness contribution deriving from the active length-force relationship can be added to this force contribution. This stiffness contribution depends on the length of the muscle relative to the optimum length; coarsely speaking, as the muscle is shorter the stiffness contribution is more stabilizing. In Fig. A to D, the effect of adding the stiffness contribution to the force contribution is shown for muscles at four different lengths. (In Fig. C the muscle is at its optimum length and hence the stiffness contribution from the length-force relation equals zero, the force contribution being the only remaining factor). Even in the most advantageous case, the stable area still excludes a large part of the thorax muscle origins. Hence a quasi-static analysis does not explain the stabilizing effect of co-contraction.




**CONCLUDING REMARKS** In the real world, systems are not quasi-static but dynamic. The quasi-static force and stiffness contributions remain relevant for dynamic systems, but lose their exclusive stability-determining quality. In particular, a quasi-statically stable system need not be stable dynamically. This depends on the time lags between the joint angle change and the resulting moment change, together with other properties of the system. Only a dynamic analysis can assess the system stability, which can also clarify the effectiveness of muscle co-contraction.

**REFERENCES** Rozendaal LA (1997) Stability of the shoulder, intrinsic muscle properties and reflexive control. PhD thesis, Delft University of Technology, The Netherlands.

# A NEW KINEMATIC MODEL OF THE SHOULDER COMPLEX

D. Cote* and M. Hubbard*†

*Biomedical Engineering Graduate Group, †Department of Mechanical and Aeronautical Engineering, University of California - Davis, Davis, CA, U.S.A.

## INTRODUCTION

The magnitude of the forces during throwing sports is a frequent cause of injuries to ligaments and other supporting tissues of the shoulder. In particular, baseball pitching requires forces much higher than those developed solely by the shoulder muscles and greatly affects the muscular and static stabilizing structures of the glenohumeral joint. The powerful shoulder internal rotation and high angular accelerations associated with pitching make it nearly impossible at present to measure directly the muscle stresses and forces on the bones and ligaments. A better understanding of the mechanisms for injury to the pitching shoulder can be achieved through dynamic musculoskeletal models of the pitching motion. At the foundation of such models is the kinematics of the shoulder girdle. Two important components of the pitching mechanics are the motion of the scapula and the relative motion between the humeral head and scapula. These essential degrees of freedom should be included in a model so that glenohumeral laxity and scapular rhythm can be accurately represented. Shoulder models to date have not completely addressed the intricacies of these motions (Högfors, et al., 1987; van der Helm, 1994), while investigations of the pitching mechanics represent the shoulder as a simple ball-in-socket between the upper arm and torso (Feltner and Dapena, 1986; Sakurai, et al., 1993).

## METHODS

The shoulder complex is composed of two units: (1) the clavicle, scapula, and thorax (CST) and (2) the humerus. Here it is assumed that the primary function of the clavicle is to define a sphere on which the acromioclavicular joint moves. The function of the scapula during pitching is twofold: (1) dictation of the lines of action for many different muscles, and (2) transmission of forces developed in the back muscles attached to the scapula (serratus anterior, trapezius, and rhomboids) to decelerate the throwing arm during follow-through. Due to the glenohumeral rhythm, the scapula changes position throughout the pitching motion. There is also some evidence of protraction of the scapula (anterior sliding motion on the surface of the thorax) during follow-through. Thus, the scapula is modeled as a rigid body with its motion confined to the surface of the thorax. The humerus is attached to the scapula at the glenoid and is free to rotate and translate relative to the scapula in all directions, subject only to the restrictions placed by the glenohumeral ligaments and the rotator cuff muscles.

## RESULTS AND DISCUSSION

The resulting kinematical model for the CST unit is described by ten generalized coordinates; 3 each for the clavicle and scapula, and 4 for the thorax. The 3 clavicle coordinates are the orientation angles associated with a body three: 2-3-1 rotation of the clavicle reference frame relative to the thorax frame. The scapula generalized coordinates are similar orientation angles using the same rotation sequence between the scapula and clavicle reference frames. The thorax is represented geometrically as an elliptical cylinder anchored in inertial space. The concave, anterior surface of the scapula contacts the surface of the thorax at two points fixed on the scapula's medial and lateral borders, respectively. These contacts confine the motion of the scapula to the surface of the thorax (see figure 1). The final four coordinates describe the location of the scapular contact points in the thorax reference frame in terms of an angle and an elevation on the surface of the thorax. Six holonomic constraint equations were derived to restrict the motion of the scapula to the surface of the thorax resulting in a CST unit with four degrees of freedom (d.o.f.). Additionally, the humerus has six d.o.f. - three rotations and three translations - relative to the scapula. Thus, the entire shoulder complex is described by ten d.o.f.

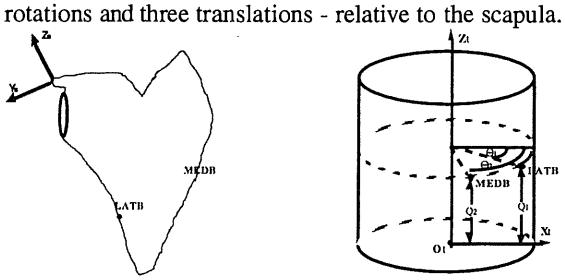



**Figure 1**: Depiction of the scapular contact points, LATB and MEDB, on the anterior surface of the scapula in the scapular reference frame, S, on the left, and the thorax reference frame, T, on the right. In the thorax frame, the contact points are described by two displacements, $Q_1$ and $Q_2$ on the Zt axis and two angles, $\theta_1$ and $\theta_2$. The angles are measured from the Xt axis in the XY plane containing the point.

This kinematic model forms the basis for a dynamic model of the shoulder complex derived using AUTOLEV - a symbolic multi-body simulation package. The inputs to the dynamic model are 21 shoulder muscles previously described by Högfors, et al. (1987), and the outputs are the orientations of the bones of the shoulder complex. The model is used to simulate abduction in the scapular plane, and the results include a prediction for the scapulohumeral rhythm. The model is also being utilized in an investigaton of shoulder muscle forces during baseball pitching. It is hoped that this model will help better elucidate some of the mechanisms for injury to the glenohumeral joint in throwing sports.

Feltner, M., & Dapena, J. (1986) Dynamics of the shoulder and elbow joints of the throwing arm during a baseball pitch. *Int. J. Sport Biomech.*, 2: 235-259.

Högfors, C., Sigholm, G., & Herberts, P. (1987) Biomechanical model of the human shoulder - 1. Elements. *J. Biomechanics*, 20: 157-166.

Sakurai, S., Ikegami, Y., Okamoto, A., Yabe, K., & Toyoshima, S. (1993) A three-dimensional cinematographic analysis of upper limb movement during fastball and curveball baseball pitches. *J. App. Biomech.*, 9: 47-65.

Van der Helm, F. C. T. (1994) Analysis of the kinematic and dynamic behavior of the shoulder mechanism. *J. Biomechanics*, 27(5): 527-550.

# CONTRIBUTION OF CUTANEOUS MECHANORECEPTORS TO THE STABILITY OF HUMAN UPRIGHT POSTURE - AN EXPERIMENTAL INVESTIGATION

G. Wu

Department of Physical Therapy, University of Vermont, Burlington, VT, USA

## INTRODUCTION AND BACKGROUND

Clinical studies have demonstrated that cutaneous mechanoreceptors in the foot play an important role in control of postural balance [Rothwell, 1994]. However, previous studies using biomechanical approaches have shown conflicting results. Diener et al. [1984] suggested that the pressure input from the soles becomes less important for the compensation of fast postural perturbation, whereas kekoni et al. [1989] demonstrated that the threshold of vibration sense in the human sole is remarkably lower with higher vibration frequency, suggesting that the mechanoreceptors are more sensitive to fast changing stimuli than to slow changing stimuli.

The goal of this study was to gain an understanding of the contribution of cutaneous mechanoreceptors in the sole in the stability of human upright posture. The body movement of human subjects was measured when their balanced upright posture was disturbed by a transient, forward/backward movement of a supporting platform. The loss of or change to the cutaneous mechanoreceptors in their feet was achieved by testing neuropathic patients with loss of vibratory sensation in the foot. It was hypothesized that the use of cutaneous mechanoreceptive information could change the characteristics of the human postural control system.

## METHODOLOGY

Patients with peripheral neuropathy were used as subjects since they have diminished sensation in the lower extremities (with minimal vibratory perception threshold of the plantar surface of the foot above 42). In addition, age and gender matched healthy subjects were selected to participate in this experiment for comparison. Their maximal vibratory perception threshold on the plantar surface of the foot was less than 23.

Each subject was tested for a total of 4 conditions, 4 times each. They included: vision/no vision, moving forward/backward. Postural perturbation was applied to the subject through an unexpected platform movement in the anterior-posterior direction. In response to the platform movement, the subject was asked to try to maintain the upright balance by moving any part of the body except for the arms.

The angular displacement and velocity of the ankle joint in the sagittal plane were directly measured by Integrated Kinematic Sensors. The maximum peak-peak value and maximum frequency range were then calculated for both angular displacement and velocity of the ankle joint for all the trials.

## RESULTS

The results showed that the neuropathic patients have significantly higher peak-peak magnitudes of ankle joint displacement (with mean and standard deviation of $9 \pm 2°$), velocity ($117 \pm 4°/s$) and frequency band ($18 \pm 3Hz$) than the healthy subjects ($4 \pm 1°$, $42 \pm 3°/s$, and $5 \pm 2Hz$, respectively).

## DISCUSSION

It is observed in this study that when the mechanoreceptive information is diminished (such as in the patients with peripheral neuropathy), the ankle movement becomes larger, and changes at a higher frequency. This tends to support the notion that the feedback from the mechanoreceptors possessed a damping characteristic which helped to further stabilize the dynamic response of the system [Wu and Zhao, 1996].

To date, the exact role of cutaneous mechanoreceptive information in the human foot in postural control is still not clear. In fact, only a limited number of researches has been devoted to this subject partly because of the difficulty of directly accessing these receptors. This study uses the neuropathic patients as a model in simulating the loss in the mechanoreceptive information. The results not only allow an investigation in the role of mechanoreceptive information in postural control, but also provide guidance towards clinical diagnosis and fall prevention strategy planing.

## REFERENCES

Rothwell J. Control of Human Voluntary Movement. New York: Chapman and Hall, 1994
Diener HC, Dichgans J, Guschlbauer B, Mau H. Brain Res. 1984, 296:103-109
Kekoni J, Hamalainen H, Rautio J, Tukeva T. Exp. Brain Res. 1989, 78:419-424
Wu G, Zhao W. Motor Control 1996, 1:3-19

## ACKNOWLEDGMENT

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# THE BODY ORIENTATION AS A CONTROLLED VARIABLE IN HUMAN UPRIGHT STANCE

G. Wu

Department of Physical Therapy, University of Vermont, Burlington, Vermont, USA

## INTRODUCTION AND BACKGROUND

The maintenance of upright balance represents a major goal of human postural control. It has been proposed that the task of maintaining a balanced upright posture is mainly controlled by the muscles at the ankle and hip joints, referred to as ankle and hip strategy [Nashner and McCollum, 1985]. When standing on a soft supporting surface and the balance is suddenly perturbed by a sudden movement of the supporting base, not only are the medium latency and long latency responses from the leg muscles significantly delayed comparing to standing on a hard surface, but also the ankle joint movement are significantly increased [Chiang and Wu, 1996]. The questions remain to be explored are: how would these changes affect the stabilization of the body? Does the increased ankle movement affect the orientation of both foot and shank segments?

In this study, the maximum peak to peak magnitude of foot and shank orientations were measured when subjects' postural balance was unexpectedly disturbed. The variation of these two variables with respect to the supporting surfaces was examined. It is hypothesized that body orientation is a controlled variable that is independent of the foot movement.

## METHODOLOGY

A total of 9 healthy young subjects (mean age 24.3 years) participated in the study. Each subject was tested for a total of 12 conditions, 4 times each. They included: vision/no vision, moving forward/backward, and standing on hard/soft/reduced surfaces. Before each trial, the subject stood quietly on the movable platform, with feet comfortably separated in the lateral direction, arms folded across the chest, and eyes looking at a target 25*m* in front of the subject. Postural perturbation was applied to the subject through an unexpected platform movement in the anterior-posterior direction of the subject, with a maximum acceleration of $8m/s^2$, speed 40cm/s and displacement 8cm. In response to the platform movement, the subject was asked to try to maintain the upright balance by moving any part of the body except for the arms.

The spatial orientation of the shank and foot in the sagittal plane was measured by two Integrated Kinematic Sensors [Wu and Ladin, 1993]. Data was collected at 120Hz. The maximum peak to peak magnitude (PP) of the foot and shank movement was calculated. In this paper, only three surface conditions with eyes open and platform moving forward are considered. They were analyzed using Analysis of Variance to identify differences among the three surfaces. The significant level was set at $p < 0.05$.

## RESULTS

For the foot, there was a significant increase from standing on a normal surface (1.8+/-1.1 deg) to a soft (6.8+/-2.7 deg), and to a reduced surface(14.1+/-4.0 deg), respectively. For the shank, however, no significant difference was found between any of the surfaces (9.5+/-3.6, 8.8+/-2.7 and 7.8+/-3.4 deg for normal, soft and reduced surfaces, respectively).

## DISCUSSION AND CONCLUSIONS

It is observed in this study that, although the maximum foot movement changes remarkably when standing on three different supporting surfaces, the maximum shank movement is essentially a constant. This supports the hypothesis that the shank orientation is a controlled variable that is independent of the foot movement.

It is well documented that the leg muscle activation is controlled by the central nervous system which integrates information from the peripheral sensory systems [Nashner and McCollum,1985]. These muscular activities are then believed to regulate the ankle joint movement. However, the findings in this study seems to suggest that the goal of the leg muscle activation is mainly for maintaining a stabilized body (or shank) orientation. When the foot is allowed to move such as standing on a soft or a reduced supporting surface, the changes found in muscle activities are to compensate for the changes due to the foot movement such as allowing more flexibility and involving the use of foot muscles.

## REFERENCES

Nashner LM, McCollum G. The Beh. & Brain Sci. 1985, 8:135-172

Chiang JH, Wu G. Gait & Posture, 1996, in press

Wu G, Ladin Z. ASME Trans. J. Biomech. Eng. 1993, 115(1):53-62

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# THE ROLE OF INTRINSIC MUSCLE PROPERTIES AND REFLEXIVE FEEDBACK DURING PERTURBED POSTURE TASKS OF THE HUMAN ARM

G.G. Brouwn
Delft University of Technology, Department of Mechanical Engineering, Laboratory of Measurement and Control, Delft, The Netherlands

INTRODUCTION: Impedancecontrol of a mechanical system is control of the dynamic relation between forces exertedon the system and the position of the system. Stability of the system is definedin terms of position deviations from a resting position, causedby external force perturbations. Optimization of system impedance yields a compromise between minimal position variance and minimization of control energy. Stability of the human arm during perturbedposture tasks is investigated. The goal is to estimate contributions of neuromusculoskeletal factors to mechanical impedanceof the system. Of particular interest are the contributions of intrinsic impedanceby co-activation of muscles and feedbackby spinal reflexes. The results will be used for validation purposes of a detailed musculoskeletal model of the human shoulder. Another application is the clinical assessment of the stability of patients with reducedproprioception.

METHODS: Stochastic force perturbations are exertedon the arm of four human subjects during a posture task. The perturbations have a rectangularfrequencyspectrum with variable bandwidthandequal weighing of all frequencies.The assumptions made for spectral estimation are: 1. Subjects behavelinear; this is a valid assumption in most cases. 2. Control signals are constant; this is a valid assumption for experiments not lasting longer than 40 seconds. 3. One experiment is representativeof the process; this is also a valid assumption in most cases. Analysis is performedusing a linearizeddynamical model, describing the control structure based on known physiological factors. The model structure contains four variables: co-activation, and feedbackgains of position, velocity and force. Linearization is justified for posture tasks under stationary stochastic force perturbations. The model parameters describethe desired contributions to mechanical impedance. The model is optimized using a criterion function that minimizes position variance, weighedby an additionalenergy criterion. By optimizing the model parametersunderdifferentconditions, the role of differentreflexive feedbackpaths will be hypothesized. Additionally the model parameters are estimatedfrom the experimental data, using frequency-domainidentification techniques. The estimation results will be usedto assess the validity of the linear model and the criterion function used.

RESULTS: The optimization results point out that proprioceptive position feedbackallows substantial increase in stiffness, but only for low-frequentexternal force perturbations. Velocity feedbackaddssignificant damping, which is beneficial in case of negligible co-activation. Force feedbackappears to have no functional role. The identification results show that the use of a linear model is justified, becausethe estimatedcoherencesare reasonably high. Parameter identification from the experimental data confirms
the hypothesized role of proprioceptive position feedback.Low-frequentforce perturbations lead to a large position feedbackgain, whereas the gain decreasesto zero when higher frequenciesare included. The feedback gain typically drops quite steeply to zero between 2 and 3 Hz. The explanation for this is that frequenciesaround the oscillation frequencyof the system would increase the position deviations significantly, as a consequenceof the time-delayin the feedbackloop.

DISCUSSION: The estimated models describethe observedbehaviour quite well and agree with the predicted behaviour by numerical optimization. Co-activation increases intrinsic stiffness and damping at the expense of considerablecontrol energy. Proprioceptive position and velocity feedbackincrease stiffness resp. damping of the system effectively, but only for low frequenciesbecause of time-delays. The implication of these results is that the model andcriterion function form a good basis for estimation of lumped contributions to mechanical impedanceunder differentconditions. The observedfrequency-dependentnonlinearity is also foundby Stein & Kearney, who studiedthe ankle joint andfoundinhibition of reflex activity during high-frequencyperturbations. Another nonlinearity, predicted by the model is increase in reflex activity with muscle activation level. Kirsch et al. have demonstratedthis effect clearly in human triceps muscle. Other experiments with isolated cat hindlimb muscle demonstrate that reflexive muscles have differentstiffness characteristics from areflexive muscles, e.g. Hoffer & Andreassen. This is also a property of the proposed model. The identification method could be a basis for clinical assessment of stability, by quantifying the amount of loss of proprioception of patients.

REFERENCES:

Hoffer JA, AndreassenS (1981) Regulation of soleus muscle stiffness in premammillary cats: intrinsic and reflex components. J Neurophysiol 45:267-285.

Kirsch RF, Kearney RE, MacNeil J (1993) Identificationof time-varying dynamics of the human triceps surae stretch reflex. Exp Brain Res 97:115-127.

Stein RB, Kearney RE (1995) Nonlinear behavior of muscle reflexes at the human ankle joint. J Neurophysiol 73:65-72.

# SACRUM LOAD-DISPLACEMENT BEHAVIOR AND SACROILIAC SELF-LOCKING IN A WEIGTHLIFTER

Guy Gosselin, , Martin C. Normand, Danyk Lafond
Laboratoire de Biomécanique

Université du Québec à Trois-Rivières

Trois-Rivières, Canada

## INTRODUCTION

The purpose of this study was to measure the load-displacement behavior of the sacrum in a 22 year old weigthlifter. There has been extensive studies on the articular physiology of the sacroiliac (SI) joint, although the loads necessary to produce movement at this articulation, *in vivo*, are not well understood. We therefore developed an original apparatus that can provide precise measurements of load-application and displacement of the segment tested in our subjects(1,2). One of the particularity of this study is that the load-displacement behavior of the sacrum was measured at 0° and 30° hip flexion, thus stretching the posterior layer of the thoracolumbar fascia. Vleeming proposes that this fascia participates in the sacroiliac locking mechanism through load transfer (3). Measurements were also taken with the presence and absence of a commercially pelvic belt, and with and without maximal hip extensors contraction.

## METHODS

The apparatus has been well described in previous communications (1,2). It is a simple lever arm that is used to apply a load on a sacral segment. The displacement is measured by a linear potentiometer activated by a vertical rod connected to the lever arm. Previously molded plaster casts were placed under the anterior superior iliac spines, under the pubis, and overlying the sacrum. The casts was molded in such a way to permit movement of the iliac bones and to apply loads perpendicular to the $S_1$ and $S_5$ segments. Loads were added from 0 to 200 Newtons (N) in 50 N increments. Measurement of displacement was taken 30 seconds after the application of the loads to allow for creep and damping. The subject was placed prone on the testing table and then prone on a triangular platform providing a 30° hip angle. Maximal isometric hip extensor contraction was obtained by immobilizing ankles to a fixed strain gauge. Contraction during load application was compared to previously measured muscular peak force.

## RESULTS

Our results show that the situation were there is the least movement (> 1 mm) is during contraction of the hip extensors with the presence of the pelvic belt, on $S_1$ or S. One exception to this result was during load application at 0° hip flexion were the displacement was 4 mm. The situation of the most displacement ( 9,2 mm) is on $S_5$ in a relaxed 0° prone position without the pelvic belt. The flexed position did not permit more than 6 mm displacement.

## DISCUSSION

The results indicate that the voluntary locking mechanism is efficient, and limits the movement at the SI during all but one position. Yet, there is also decreased movement in the flexion/relaxed/no belt/$S_1$ and $S_5$ positions. This may be explained by the increased tension provided by the load transfer from the thoracolumbar fascia, although the increased stress at the SIJ through the friction of the facet irregularities produced by the flexion of the hip joint may also contribute the self-locking mechanism. Concomitant EMG studies would help in evaluating the role of the thoracolumbac fascia in SIJ locking. Furthermore, the voluntary contraction of the hip extensors seems more important in the limitation of SI movement then the presence or absence of a pelvic belt as a stabilization device. We may conclude that posture education and hip extensors strength should be considered in the presence of an individual suffering from SI dysfunction. Further studies in the relationship of increasing strength and subsequent SI movement will provide additional knowledge and will help us in the rehabilitation process of individual suffering from low back pain.

(1) Gosselin, G., Normand, MC., (1993) « Development of a sacrum nutator: Tool for the load-displacement behavior of the sacroiliac joint in vivo », **International Society of Biomechanics XIV Congress**, Paris.
(2) Gosselin, G., Normand, MC, (1995) « Load-displacement behavior of the sacrum in vivo », **Second Interdisciplinary Wrold Congress on Low Back Pain and its Relationship to the Sacroiliac Joint**, San Diego.
(3) Vleeming, A., et al, (1995) « The Posterior Layer of the Thoracolumbar Fascia », **Spine**, 20(7), pp 753-758.

# INTERVERTEBRAL MOVEMENTS PRODUCED BY POSTEROANTERIOR MOBILISATION

RYW Lee* and JH Evans#
*The University of Sydney, Sydney, Australia
#The Hong Kong Polytechnic University, Kowloon, Hong Kong

## INTRODUCTION

Posteroanterior mobilisation is frequently employed by physiotherapists and other clinicians in the examination and treatment of patients with low back pain. Earlier research[1,2,3] has led to a better understanding of how the spine deforms during mobilisation. Video motion analysis system[1] and displacement transducers[2,3] were used to measure the displacement of skin over the spinous processes when the spine was subjected to mobilisation load[1,2,3]. However, these surface techniques did not allow the researchers to determine the movements at the intervertebral level.

The purpose of this study was to measure the intervertebral movements of the lumbosacral spine and the displacements of the spinous processes when the L4 vertebra was subjected to mobilisation.

## METHODS

Twelve healthy male volunteers with a mean age of 26.9±7.4 years were recruited for this study. They were positioned in prone-lying and a specially designed loading mechanism was employed to exert a vertical mobilisation force of 150N onto their L4 spinous processes. Lateral radiographs were taken of the subjects before and during the mobilisation load was applied. The images of the vertebrae on the two radiographs were traced and digitised. The sagittal rotation and translations of the motion segments, and the displacements of the spinous processes were computed. The process of tracing and digitisation was repeated three times. The computed results were averaged to provide the best evaluation of the movement.

## RESULTS

It was shown that the mean error in determining sagittal rotation and translation, and displacement of the spinous process was found to be 1.0°, 0.6mm and 0.8mm respectively. It was considered that the experimental technique was sufficiently accurate for the purpose of this study.

Under the application of the mobilisation load, the lumbar motion segments (L1/2-L4/5) were shown to extend. The mean extension movements ranged from 1.2° to 2.4°. The L5/S1 segment was less consistent in regard to the direction of rotation. The magnitude of intervertebral translations observed was small (less than 2mm in all cases). In general, the upper lumbar segments (L1/2-L3/4) translated posteriorly and the L5/S1 segment anteriorly. The L4/5 segment appeared to be a transitional one with subjects showing translations in both directions.

All the five lumbar spinous processes moved in the anterior direction, indicating a sagging of the spine during mobilisation. The mean anterior displacement ranged from 8.8mm to 11.5mm.

## DISCUSSION

It is suggested that the spine is a flexible beam resting on the pelvis and thoracic cage. Posteroanterior mobilisation produces reaction forces at these supporting points, leading to three point bending of the beam. Under this loading condition, the spine is subjected to a combination of extension moment and shear. The direction of shear will be posterior above the point of load application and anterior below such point. The movement pattern observed in this study are in remarkable agreement with the loading conditions described. This strongly supports the hypothesis that posteroanterior mobilisation is effectively a three point bending of the lumbar spine.

Posteroanterior mobilisation elicits a complex movement pattern. This refutes the physiotherapists' belief that it simply produces anterior gliding of one vertebra upon another[4]. In addition, the intervertebral movements produced by mobilisation are only in the order of a few degrees or mms. It is extremely unlikely that therapists could perceive such a small magnitude of movement. Manual assessment of intervertebral movements is therefore unreliable. It is probable that what the therapist perceives is the sagging of the spine or the anterior displacement of the spinous process rather than the corresponding intervertebral movements. Clearly, mobilisation is not a test of a single motion segment as suggested by previous author[4] but the bending stiffness of the entire spine.

## REFERENCES

[1] Watson MJ, Burnett M and Dickens W (1989) Experiment in recording passive spinal movement. *Physiother* 75: 747-749.
[2] Thompson R (1983) Movements of relative intervertebral displacements in the lumbar spine during application of a PAIVM, Postgraduate Diploma dissertation, Lincoln Institute of Health Sciences, Melbourne.
[3] Lee RYW and Evans JH (1992) Load-displacement-time characteristics of the spine under posteroanterior mobilisation *Aust J Physiother* 38: 115-123.
[4]Maitland GD (1986) *Vertebral Manipulation.* 5th edition, Baltimore, London.

# STABILITY OF THE WHOLE LUMBAR SPINE AFTER GRADED FENESTRATIONS AND DISCECTOMIES

WW Lu[1], DK Ruan[2], KDK Luk[1], JCY Leong[1]

1. Department of Orthopaedic Surgery, The Duchess of Kent Children's Hospital, The University of Hong Kong
2. Department of Orthopaedic Surgery, Tangdu Hospital, Fourth Military Medical University, Xi'an, P.R. Chine

**INTRODUCTION:** Degeneration of the lumbar spine is the most frequent cause for lumbar stenosis. The pathological changes consist of bony spur of vertebral body, posterior protrusions of intervertebral disc, and osteophyte formation on the facets. Clinically, multilevel fenestration with undermining enlargement of spinal canal is the effective method for treatment of lumbar spinal stenosis. However, the stability and the biomechanical effects of graded fenestrations and discectomies on the motion behaviour of the whole lumbar spine after the treatment remain unclear. The purpose of the current study was to investigate the effects of graded fenestrations and discectomies on the whole lumbar spine, including segment angle changes, horizontal (sagittal) and vertical shifting.

**METHODS**: Five fresh human specimens from L1 to S1 were used in this study. The graded fenestrations and discectomies consisted of L4-5 bilateral fenestrations, L5-S1 bilateral fenestration, L3-4 bilateral fenestration, L4-5 discectomy and L5-S1 discectomy. Three kinds of loads: flexion/extension, lateral bending and axial rotation (torsion), were applied using a material testing system (INSTRON® 1341). The ranges of motion (ROM) were determined three-dimensionally by the Elite 3-D motion analysis system (BTS, Italy) with two infrared-cameras. The data from the injured tests were normalised to the intact results. The normalised data were thereafter pooled for statistical analysis.

**RESULTS and DISCUSSION**: After the second injury, the ranges of motion in sagittal translation (horizontal) increased from 0.7 to 1.6 mm at L4-5 and L5-S1 segment during the flexion/extension loads. Such sagittal translation was increased with almost 100% ($p< 0.01$) after the fourth injury was performed. Meanwhile, the axial rotations were increased from 3.3% to 12% after the second graded injury. After the graded injuries, however, no significant motions were found in each of the lumbar segment during the lateral bending. The ROM in axial rotation and vertical shifting did not show statistical significant differences in all loading modes after all the operative injuries.

The traditional management of lumber stenosis was total laminectomy. Based on long-term follow up, this method has high rates of lumbar spinal instability and secondary stenosis. Along with better understanding of the pathophysiology of lumbar stenosis, several alternative surgical treatments for lumbar stenosis have been proposed recently. This method using multilevel fenestrations was able to reduce laminectomy, thus effective decompression of the spinal canal.

**CONCLUSION:** The present results demonstrate that multilevel fenestrations and discectomies affect the lumbar spinal stability in flexion/extension, but do not make the lumbar spine unstable in lateral bending and axial rotation.

# SPINAL AND RIB CAGE 3-D KINEMATICS FOR NORMAL AND IDIOPATHIC SCOLIOSIS SUBJECT

WW Lu, JCY Leong, KDK Luk, MK Lee

Department of Orthopaedic Surgery, The University of Hong Kong

Pulmonary impairment is the most serious consequence of idiopathic scoliosis, and absence of treatment may lead to cardiorepiratory failure. No single feature of deformity accounts for pulmonary function reduction to a clinically useful extent. Rib cage movement is closely related to lung function, whether the reduced lung capacity was related to rib cage movement change was unknown. In addition, whether the spinal motion has any relationships to the deep breathing was not reported before. Thus the purpose were to determine the rib cage as well as thoracic spine motion patterns during deep breathing for normal and pre-operative scoliosis subjects. This study was part of a comprehensive evaluation of the pulmonary deterioration before and after surgery.

**SUBJECTS AND METHODS** The study sample consisted 27 pre-operative adolescent idiopathic scoliosis patients (thoracic) and 15 normal subjects (10 to 35 years). The most adolescent idiopathic scoliosis had a cobb angle of less than 60° . Four infrared-camera (Elite, Italy) were used to record the 3-D motion of Rib cage and spine during deep breathing. The subjects were seated and kept upright posture with their thigh constraint to the chair by a belt. The rib cage and thoracic spine relative to T12 motion were calculated. In order to evaluate the stiffness of the spine, the lateral bending of the spine was also measured. Student's t tests (unpaired and two tailed distribution) were used to determine whether statistically significant differences existed between the normal and patient regarding rib cage and spine maximum movements. A level of $p< 0.05$ was chosen as a level of statistic significance.

**RESULTS AND DISCUSSIONS** We found the upper and middle level rib cage have the larger movement in vertical and a-p directions with the range from 2 to 4 cm. The low level rib cage has smaller range of motion. (about 1 cm). Compare to normal subjects, the patients with thoracic scoliosis showed smaller range of movements three dimensionally with about 2 cm movement vertically and 1 cm in a-p directions. The differences in motion of the upper and middle level of the rib cage were statistically significant.

We had further detected the thoracic spine motion patterns during deep breathing. Within normal subjects, the spine at C7-T1 level moved posteriorly for about 3.6 cm with the range of 3 to 4.5 cm. Meanwhile, the spine also moved vertically by about 2 cm. In other words, the spine was straightened from its nature kyphosis during deep breathing. Although the similar spine motion patterns were found in scoliosis patients, the average range of motion was much smaller. (0.6 cm in vertical and 2.1 cm in a-p direction). All these differences were statistically significant. As well, the inter-subject variation for the scoliosis subjects were larger than those of normal. The results also shown that the scoliosis patients had more stiff spine than those normal subjects. The mean lateral bending angle for normal was 58 ° (right bending) and 54° (left bending), and the bending angle for the scoliotics was 36 ° and 45 ° respectively.

Spinal movement during deep breathing was not well defined in literature. In most respiratory text-books, breathing were only related to the rib cage motion, the diaphragm motion, and the abdominal wall motion. From our tests, we found during deep breathing, the spine (thoracic part) was also involved. We assume after spine fusion for the scoliosis patients, the motion would be further reduced. Whether such smaller motion was related to disproportional changes in pulmonary functions following spinal fusion have to be further investigated.

**Prolonged Biomechanical Load On Spinal Tissues Is A Risk Factor For Occupational Low Back Pain**

Norman, R., Wells, R., Neumann, P., Kerr, M.†, Frank, J.†, Shannon, H.†, and the OUBPS Group
Occupational Biomechanics Laboratories, Department of Kinesiology, University of Waterloo, Waterloo, Institute for Work and Health†, Toronto, Ontario, Canada

## INTRODUCTION:

Although prolonged physical loading of spinal tissues is often assumed to be a risk factor for occupational low back pain, epidemiological evidence is meagre. Injury absence costs in the work place are huge; well documented risk factors and injury mechanisms are essential for reliable identification of high risk jobs and assessment of the effectiveness of job modifications intended to reduce risk. A case/control epidemiological study was conducted in a large auto assembly operation to identify risk factors. This paper addresses the role of prolonged loading as a predictor of reported low back pain in the auto assembly industry.

## METHODS:

Peak and prolonged low back exposure variables were collected using specially developed and tested biomechanical methods adapted for field use. The frequency and duration weighted integral of all of the worker's spinal loading peaks or time averages of loading variables across tasks were taken to be the measure of prolonged spinal loading exposure. Exposure variables were developed from outputs of a two dimensional, quasi-dynamic biomechanical model that could deal with symmetrical or asymmetrical pushes, pulls or lifts and output L4/L5 moment, compression and shear forces. Kinematic measures of trunk motion were obtained from a digital video posture analysis system. Measures were made over a two year period on 283 auto workers for durations of 1 to 8 hours. Of these, 84 reported back pain to a nursing station and, together with 20 "proxies" (same job as those in pain), were categorized as "cases"; 179 uninjured workers were randomly selected as "controls" from a base of approximately 10,000 hourly paid workers.

Table 1: Means of peak and prolonged loading variables

| | Cases | | | Random | | |
|---|---|---|---|---|---|---|
| **Variable** | **n** | **Mean** | **SD** | **n** | **Mean** | **SD** |
| Peak Compression (N) | 104 | 3423 | 1421 | 130 | 2733 | 1073 |
| Peak Moment (N·m) | 104 | 182 | 84.3 | 130 | 140 | 62.7 |
| Peak Shear (N) | 104 | 465 | 176 | 130 | 353 | 159 |
| Peak Hand Force (Kg) | 104 | 22.6 | 20.5 | 129 | 13.7 | 12.5 |
| Peak Flexion Level (deg) | 104 | 51.2 | 22.4 | 130 | 39.3 | 23.3 |
| Peak Flexor velocity (deg/s) | 104 | 41.5 | 15.14 | 130 | 34.1 | 17.2 |
| Integrated Compression (N·s) | 104 | 1.48E7 | 1.01E7 | 130 | 1.23E7 | 6.68E6 |
| Integrated Moment (N·m·s) | 104 | 5.03E5 | 4.85E5 | 130 | 3.65E5 | 2.64E5 |
| Integrated Shear (N·s) | 104 | 1.36E6 | 1.30E6 | 130 | 1.01E6 | 8.14E5 |
| Usual Hand Force (Kg) | 104 | 8.8 | 6.8 | 129 | 5.7 | 5.3 |
| # flexion moves/min | 104 | 2.9 | 2.1 | 130 | 2.3 | 2.3 |
| Ave. degrees moved / min | 104 | 307.5 | 137.0 | 130 | 252.3 | 133.3 |

## RESULTS, DISCUSSION AND CONCLUSIONS:

All peak ($p<0.01$) and prolonged loading variables ($p<0.05$) showed significant differences (t-tests) between cases and controls (Table 1). Univariable odds ratios (ORs) and 95% confidence intervals were calculated for each variable from the 25th to 75th percentile of exposure of the random controls (Fig. 1). Large estimates of risk resulted. Calculation of ORs from the full range of observed exposure level showed extremely large risk estimates for workers at maximum observed values. Stepwise logistic regression resulted in inclusion of both "peak" and "prolonged" loading variables in a multivariable model (asterisks, Fig. 1). Spinal shear force was a strong predictor, indicating a low back injury mechanism that needs further study. To our knowledge, for the first time both prolonged and peak loading of spinal tissues, and trunk motion biomechanical variables, have been epidemiologically verified as independent risk factors for occupational low back pain.

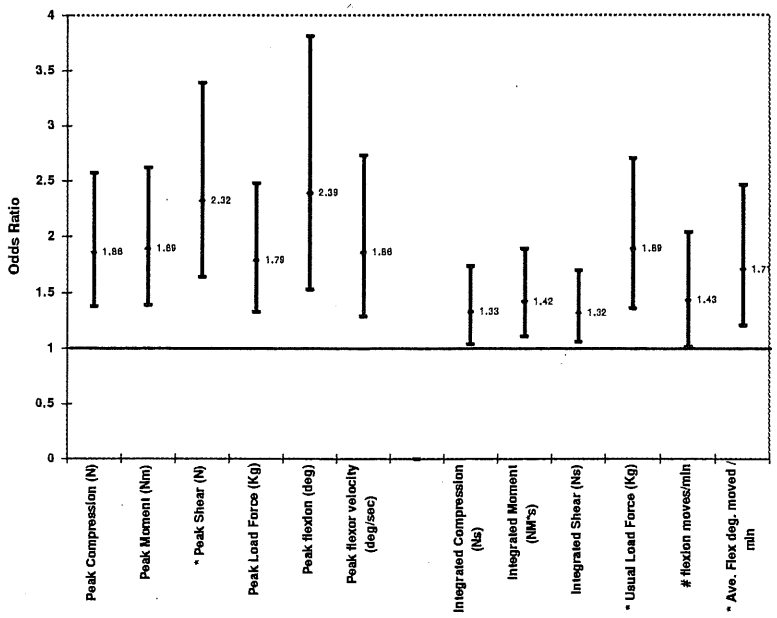



**Figure 1**: Univariable ORs using the inter-quartile spread as the exposure difference.

# THE RELATIONSHIP BETWEEN ARM POSTURE AND TRAPEZIUS EMG ACTIVITY IN VOCATIONAL WORK TASKS

R.H.Westgaard and O. Vasseljen jr
Div. of Organization and Work Science, Norwegian University of Science and Technology, Trondheim, Norway

## INTRODUCTION

In the attempt to measure biomechanical exposure as a risk factor for work-related shoulder and neck complaints, both arm posture and trapezius EMG are used as variables to indicate loading of the shoulder (Aarås et al. 1988). Examples of guidelines for acceptable work load are muscle load below 8 % of maximal muscle force and arm elevation below 15°. The two measurements of shoulder load are correlated in laboratory experiments when static, or near static postures are observed (Sigholm et al. 1984), but we do not know whether the two variables represent equivalent measurements of vocational work load. The aim of the study was to examine whether arm elevation and trapezius EMG can be used as interchangeable measures of vocational biomechanical exposure in the shoulder and neck region for low to moderate work loads.

## METHODS

Twentyeight manual and 48 office workers participated in the study, one half had experienced shoulder and neck complaints, the other half was symptom free. Thirty minute recordings of surface EMG were used to quantify the activity of both upper trapezius muscles during work. An electrode assembly with a preamplifier and integrated bipolar silver electrodes, 6 mm in diameter and 20 mm intercenter distance, was used for the recordings. The center of the electrode assembly was placed 2/3 the distance from the spine of C7 to the edge of the acromion. The signals were full-wave rectified, averaged at 0.2 s resolution and quantified in percent of the EMG activity at maximal voluntary contraction (%$EMG_{max}$). Postural angles of the upper arms were recorded continuously, in parallell with the EMG recordings, by electrolytic liquid level sensors (Physiometer PHY-400, Premed A/S, Norway). Two sensors were placed at right angles, allowing measurements in two planes (flexion, abduction). The signals were sampled and time-averaged at a resolution of 0.2 s. These values were later transformed into polar coordinates, quantifying elevation and horizontal orientation. The association between arm elevation and trapezius EMG was quantified by comparing median elevation angle with median trapezius activity across all subjects in each of the two groups. Cross-correlation functions were generated for the right trapezius muscle of 8 office and 8 manual workers, to compare EMG activity and arm elevation at high time resolution.

## RESULTS

Median arm elevation varied from 8 to 28° and median trapezius activity from 1 to 10 %$EMG_{max}$ for both work groups. There was no correlation between median arm elevation and median upper trapezius activity in either group. Similar results were seen when other EMG and posture variables, presumably quantifying related exposure dimensions, were analysed, e.g. time with muscle activity below 0.5 %$EMG_{max}$ vs. time with the arm below 10° arm elevation. The cross-correlation function showed that the arm elevation was delayed relative to the trapezius muscle activity, mean time delay was 0.44 (manual workers) and 0.34 s (office workers). After adjusting for the time delay, the correlation between arm elevation and trapezius EMG was 0.43 (95% CI: 0.37-0.49) for the manual workers, 0.32 (95% CI: 0.26-0.37) for the office workers. When sorting the EMG activity level according to arm elevation for different arm elevation intervals (0-15°, 16-30°, 31-45°, >45°), a considerable inter- and intraindividual variation in muscle activity was observed within each interval for both work groups. For the three intervals below 45°, there was a slight increase in median EMG activity and no increase in static level with increasing arm elevation for both work groups. The EMG variables showed a relatively larger increase when the arm elevation increased beyond 45°.

## DISCUSSION

Arm elevation and trapezius muscle activity are not equivalent indicators of shoulder load for muscle activity levels below 10 %$EMG_{max}$ and arm elevations below 45°. There were considerable differences between individuals in how they activate their trapezius muscles for the same arm elevation, when performing similar work tasks. Furthermore, arm elevation did not distinguish periods with relatively high and low trapezius activity for the same subject, within the range studied here. This was true both for the near-static work situations of the office workers and for the much more dynamic work patterns when performing postal sorting. External hand loads were either nonexistent (office workers) or similar for most subjects (postal workers). Muscle activation caused by shoulder elevation and co-contraction of muscles to stabilize the scapula are possible factors to explain this differentiation of the postural and electromyographic variables. The result has important implications for biomechanical models that describe the function of the shoulder, by pointing to a considerable deviation of force levels measured by EMG recordings from force levels estimated by theoretical optimization criteria.

## REFERENCES

Aarås A, Westgaard RH, Stranden, E. *Ergonomics* 1988; 31:915-933
Sigholm G, Herberts P, Almström C, Kadefors R. *J Orthop Research* 1984; 1:379-386

# REGAINING OF POSTURAL STABILITY AFTER TRAUMATIC THORACIC SPINAL CORD INJURY: A LONGITUDINAL CLINICAL STUDY.

**H.A.M. Seelen and Y.J.M. Potten**
Institute for Rehabilitation Research, Hoensbroek, The Netherlands

## INTRODUCTION

In thoracic spinal cord injured (SCI) subjects postural control, essential for performing activities of daily living, is impaired due to sensorimotor function loss in body parts caudal to the lesion. During rehabilitation these patients undergo physical therapy aimed at (partially) compensating this loss using residual sensorimotor functions. It has been shown that thoracic SCI subjects who have completed their rehabilitation have learned to compensate the loss of erector spinae (ES) muscle activity by increased use of the latissimus dorsi (LD) and the trapezius pars ascendens (TPA), which in turn necessitates the use of scapular protractors to stabilize the shoulder girdle [1]. In fact new postural muscle synergies have developed in these subjects. The aim of the present research was to investigate (a) the time course this development of alternative postural muscle use followed in thoracic SCI patients during their clinical rehabilitation and (b) the effects of this motor development on sitting balance during bimanual task performance. Results from a longitudinal, clinical, experimental study will be presented.

## METHODS

Twelve patients, divided over two groups of subjects with either a complete low thoracic SCI at level T9-T12 (L-group) or a complete high thoracic SCI at level T2-T8 (H-group), participated in this study. At several well described points during their clinical rehabilitation patients were tested on their ability to maintain sitting posture during balance perturbing reaching movements. These time points or sessions were chosen as (A) 1.5 weeks after admission to the rehabilitation centre, (B) immediately after spinal braces were removed, (C) four weeks after B, D halfway between session C and the discharge date and (E) at discharge. During each session patients were asked to perform a standardized task involving a number of bimanual reaching movements towards buttons placed at distances of 15, 30, 75 and 90% of their individual maximum reach in the mediosagittal plane, thus introducing graded postural perturbations. Subjects sat in a chair mounted on an AMTI platform. During task execution surface EMG activity of the bilateral erector spinae (ES) at level L3, T9 and T3 as well as the TPA, LD, pectoralis major (PM), serratus anterior (SA) and oblique abdominal (OA) muscles was recorded. From the force plate signals centre of pressure (CP) displacement was calculated. Maximal CP displacement (CPmax) was determined. Mean rectified EMG values were computed over a period in which maintaining sitting posture was most difficult, i.e. during the initial phase of balance restoration after reaching towards the peripheral buttons. This coincided with the phase in which CPmax was found. Statistical analyses included Page tests for ordered alternatives and Mann-Whitney U-tests.

## RESULTS

Results from the CP calculation show a high within-subject reproducibility. CPmax, as an indication of the range in which postural perturbation elicited by the task could be compensated actively by the patients, increased during rehabilitation in the L-group ($p<0.01$). CPmax declined after removal of the spinal brace, i.e. between session A and B, in the H-group subjects. However, in this group it gradually increased from session B onwards until clinical discharge. In the H-group increased balance restoration activity across the rehabilitation was found in the ES-T9 ($p<0.01$), LD ($p<0.001$) and TPA ($p<0.05$). In the L-group the bilateral ES-L3, ES-T9 and TPA showed an activity increase during the rehabilitation phase ($p<0.01$). The H-group showed more bilateral LD and TPA activity and left ES-T3 activity towards the end of the rehabilitation period in contrast to the L-group ($p<0.01$). The SA, PM and OA activity during task execution did not change significantly across the rehabilitation period in either group.

## DISCUSSION

Results confirm that the methods used provide valuable extra (quantitative) information as to the development of different strategies of (alternative) postural control in SCI patients during the clinical rehabilitation period. Data from this study indicate that in both high and low thoracic SCI patients partial restoration of postural function of the ES at levels near the lesion site occurs during rehabilitation. Also in high thoracic SCI patients a clear increase of LD and TPA activity as a function of rehabilitation was observed, whereas in the low thoracic SCI patients this increase was less. These data are in agreement with the concept of alternative postural muscle use as mentioned earlier [1]. Results give reason for future research, including evaluation of effects of specific therapy interventions on the development of alternative postural muscle use.

## REFERENCES

1. Seelen HAM, Potten YJM, Huson A, Spaans F, Reulen JPH: Impaired balance control in paraplegic subjects. J Electromyogr Kinesiol (in press).

# RELATIONSHIP BETWEEN MUSCLE STRENGTH AND POSTURAL SWAY IN SUBJECTS FOLLOWING INVERSION SPRAIN OF ANKLE

Rong-Ju Cherng[1,2], Chun-Pin Lin[2], Fong-Chin Su[1], Jin-San Her[1]
[1]Institute of Biomedical Engineering, [2]Department of Physical Therapy
National Cheng Kung University, Tainan, Taiwan

## INTRODUCTION

Deficits of postural control associated with sprained ankles have been substantiated by many researchers(1,2). Muscle weakness of the affected ankle was noted among subjects with unilateral ankle sprain(3). However, whether muscle weakness contributes to the exaggerated postural sway in ankle sprain has not yet been investigated. The purpose of our study was to examine the relationship between muscle strength and postural sway in subjects with ankle sprain.

## METHODS

Twelve subjects (male:8, female:4; age20-35 years, mean:23, SD:4; body weight 52-70kg, mean:62.3, SD:6.6) with unilateral ankle sprain participated in this study. All of them completed a postural sway test, in which subjects were tested barefooted in one-leg-stand for 30 seconds under four conditions: 1) injured leg, eyes open; 2) noninjured leg, eyes open; 3) injured leg, eyes close; and 4) noninjured leg, eyes closed. Postural sway was measured by the variability of the force signal in the mediolateral (Fx) and anterioposterior (Fy) directions on a Kistler-9284 force platform with a sampling rate of 100 Hz. Ten of the subjects also finished a muscle strength test. Muscle strength of peak torque of ankle evertors were tested isokinetically on Cybex 350 under three angular velocities: 30/sec, 60/sec and 180/sec. Repeated measures ANOVA was used to analyze the effects of eye and injury, and the interaction of both factors on the measure of postural sway. The effect of injury on the peak torque of ankle evertors was similarly analyzed. Pearson Correlation Coefficient was computed between postural sway and muscle strength.

## RESULTS

Postural sway was significantly different between injured and noninjured legs ($F(1,11)=6.02$, $p< .05$ for Fx, $F(1,11)=9.87$, $p< .05$ for Fy) and between the eyes open and eyes closed conditions ($F(1,11)=24.19$, $p< .05$ for Fx, $F(1,11)=22.78$, $p< .05$ for Fy). In addition, the effect of injury was significantly greater when subjects' eyes were closed than when their eyes were open ($F(1,11)=5.83$, $p< .05$ for Fx, $F(1,11)=8.08$, $p< .05$ for Fy). However, there was no significant effect of injury on the peak torque of ankle evertors ($t(9)=1.79$, $p> .05$, for 30/sec; $t(9)=1.94$, $p> .05$, for 60/sec, $t(9)=1.71$, $p> .05$ for 180/sec). The correlation between muscle strength and postural sway was not statistically significant ($r= - .23$, $p>. 05$ for Fx, $r= - .16$, $p>.05$ for Fy ).

## DISCUSSION

Our results show that ankle sprain affects postural control, and more so when subjects' eyes are closed. Ankle sprain, however, does not affect muscle strength of ankle evertors and postural control is not correlated with muscle strength of ankle evertors. We interpret our results as follows: Postural control depends on integration of sensory information, neural process, and musculoskeletal mechanism. Accordingly either sensory abnormality and/or muscle weakness can lead to postural control problems. Vestibular, somatosensory and visual information are organized to provide the brain with accurate information of body's orientation in space. When eyes are closed, somatosensation is relied on for postural control. With noninjured legs, the somatosensation is not impaired and it can provide enough information for postural control. That's why there was no difference of postural sway with noninjured legs when the eyes are open or closed. However, with sprained ankle, the proprioceptors may be damaged at the time of injury, resulting in insufficient or inappropriate somatosenory information to compensate for the occlusion of vision. This explains the greater postural sway with injured legs when eyes are closed than when eyes are open. The insignificant correlation between muscle strength and postural sway may indicate that somatosensation, rather than muscle strength of ankle everators play a major role in the deficits of postural control of subjects with ankle sprain. Research is needed to clarify this point further.

## REFERENCES

1. Patricia A. Goldie, et al. Arch Phys Med Rehabil 1994; 75: 969-975
2. Friden T. Am J Sports Med 1989; 17:118-122
3. Tropp H. Med Sci Sports Med 1986; 7: 291-295

DIVERGING ACTIVITY PATTERNS IN DEEP TRUNK MUSCLES DURING TRUNK ROTATIONS

E.A. Andersson, H. Grundström[a] and A. Thorstensson.
Departments of Neuroscience, Karolinska Institute and Human Biology, University College of Physical Education and Sports, and [a]Department of Radiology, Danderyds Hospital, Stockholm, Sweden.

INTRODUCTION

The activity levels in deep and superficial trunk and hip muscles have been investigated in various postures and training exercises (Andersson 1997), but the information concerning rotations of the trunk is scarce.

METHODS

Electromyography (EMG) was recorded with fine-wire electrodes, inserted with needles under ultra-sound guidance, at the level of the third to fourth lumbar vertebrae from the midportion of psoas (PS) and quadratus lumborum (QL), the deep lateral and the superficial medial portion of erector spinae (ES-d and ES-s), obliquus externus (OE) and internus (OI) as well as rectus abdominis (RA). The iliacus muscle (IL) was studied deep in the groin. The average EMG-amplitude was expressed in percent of the highest activity observed in all the static rotational efforts, for each muscle and subject.

RESULTS

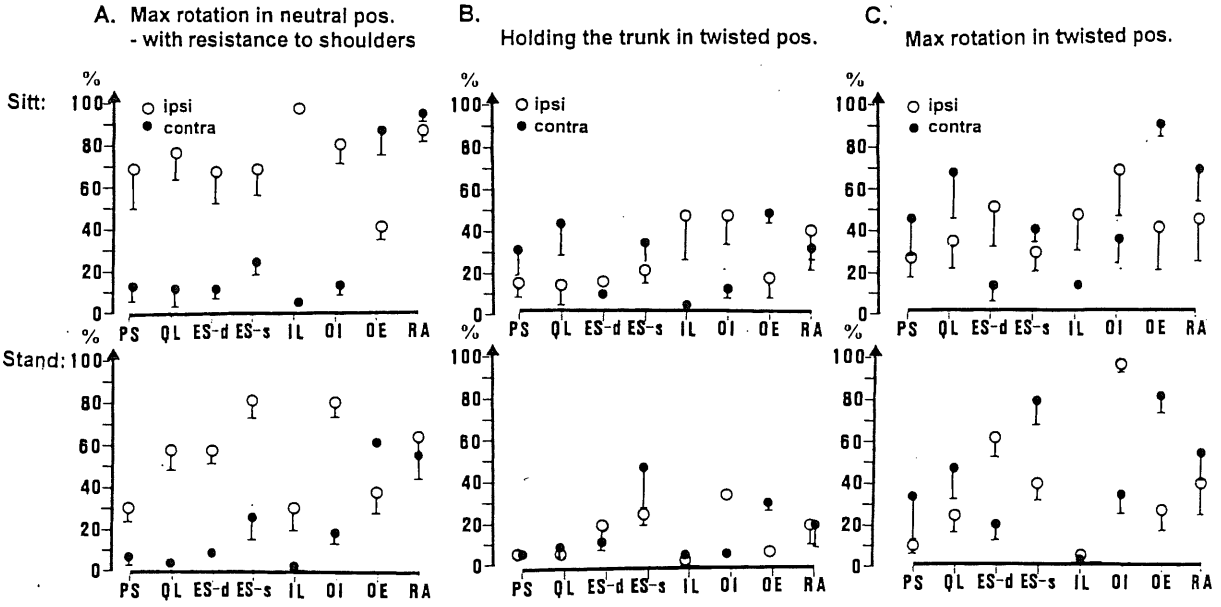



Figure 1. Mean EMG levels (±SE, n=4) in three static rotations to either side, performed in sitting and standing.

**A:** A higher activity was generally seen for all muscles in a maximal static rotation to the ipsi- than to the contralateral side, with external resistance to the shoulders and the trunk held in a neutral position, both in sitting and standing (Fig. 1A). The only exceptions were present for OE, with an opposite pattern, and RA, presenting similar levels between the two sides. The most striking difference between sitting and standing was observed for PS and IL, showing higher values in the former position.

**B and C:** Holding the trunk, without any external resistance to the shoulders, in an extremely twisted position, either relaxed (B) or with maximal effort (C), generally created similar patterns for all muscles regarding the direction dependent ratios between ipsi- and contralateral activity (Fig. 1B,C). As compared to task A, a drastically different pattern was seen in B and C for PS, QL and ES-s, all three presenting relatively higher involvement in turns to the opposite side in the latter tasks.

DISCUSSION

The present data show that the activation pattern of deep trunk and hip muscles can be specific even within the same type of task, here attempted trunk rotations. Factors like initial posture, neutral and twisted, sitting or standing, and presence of resistance to the shoulders or not, may change, or even reverse the ratio af activation between corresponding muscles on the ipsi- and contralateral side. Additional studies are needed to clarify the functional role of each muscle in different situations.

REFERENCES

1. Andersson E. A. (1997) EMG and strength in trunk and hip muscles, especially the iliopsoas. Thesis. Karolinska Institute, Stockholm, Sweden.

# TRUNK FLEXION FORCE DURING PREGNANCY

W. L Gilleard
Faculty of Health Sciences, University of Sydney, Lidcombe, Australia.

## INTRODUCTION

For most women, pregnancy is a normal healthy state throughout which trunk flexion tasks utilising the abdominal muscles must be performed. Previous studies of abdominal muscle function using manual muscle tests such as curl-ups and pelvic stabilisation during straight leg lowering have reported a reduction in abdominal muscle functional ability in late pregnancy[1,2]. In the post-birth period, both reduced functional ability[2] and no difference to control groups[2-4] have been reported. Where the same subjects were assessed by the two different manual muscles tests, conflicting results were found[2].

These previous studies, although giving an indication of muscle function, have not been able to provide measures of trunk flexion force. Quantitative knowledge of abdominal muscle functional abilities during pregnancy and the immediate post-birth period is important for the development of post and prenatal care programs. Therefore the objective of this study was to quantitatively investigate abdominal muscle function in a range of postures.

## METHODS

Volunteer primigravida subjects choose to participate in either a) 25-30 weeks gestation group or b) 34-38 weeks gestation group or c) eight weeks post birth. Nullipara subjects were used as a control group. Each subject was positioned standing on a height adjustable platform with a non-skid surface. The height and angle of the crossbar was adjusted such that the padded bar and load cell unit were against the skin at the level of the manubriosternal junction. Padded bars were located at the distal anterior thigh and posterior to the gluteal area so that the lower limbs were stabilised. The subject was first asked to rest quietly and a baseline level was recorded. The subject then performed five trials of a maximum isometric trunk flexion. The base line recording and five trials of a maximum isometric trunk flexion were then repeated at 25 degrees and 50 degrees of hip flexion. The order of testing was randomly allocated. Each trial was displayed and the first 500ms of the visually determined sustained maximum contraction and 500ms of the baseline trial was selected for further analysis. For each of these samples, the maximum force less the baseline value was calculated. The largest resultant force for each subject was then included in the statistical analysis. Independent t-tests with assumed unequal variances were used to determine differences between groups for each test position. A two-tailed test was conducted with alpha set at 0.05.

## RESULTS

There was significantly less force produced during isometric trunk flexion at all three positions in the post-birth group when compared to the control group. There was no significant difference between the test groups during pregnancy, and between pregnancy groups and the control group.

## DISCUSSION

The control group mean flexion force while standing is similar to that previously reported by McNeill et al. (1980)[5]. Trunk flexion force measured during pregnancy was not significantly different from controls. Therefore abdominal muscle adaptations to pregnancy do not appear to affect the ability of the abdominal muscles to isometrically flex the trunk in the standing or flexed posture during pregnancy. Post-birth eight weeks however, the ability of the abdominal muscles to flex the trunk was reduced in all positions. This may be due to incomplete muscular re-adapting to a nongravid state.

## REFERENCES

1 Fast, A., Weiss, L., Ducommun, E., Medina, E. and Butler, J. (1990) Low-back Pain in Pregnancy, Abdominal muscles, Sit-up performance and Back pain. *Spine* 15, 28-30

2 Gilleard, W. L. and Brown, J. M. M: (1996) The Structure and Function of the Abdominal Muscles in Primigravida during Pregnancy and the Immediate Post-birth Period. *Phys Ther.* 76, 750-762

3 Jackson, J. and Kleinig, A. (1990) A Comparison of Abdominal Muscle Strength in Postnoatal and Nulliparous Women of Two Age Groups. Journal of the National Women's Health Group, Australian Physiotherapy Therapy Association 10: 1, 31-36

4 Spence, M. (1978) Postnatal Survey. Australian Journal of Physiotherapy 24:3, 151-160.

5 McNeill, T., Warwick, D., Andersson, G. and Schultz, A. (1980) Trunk strengths in attempted flexion, extension, and lateral bending in healthy subjects and patients with low-back disorders. *Spine* 5, 529-538.

This study was supported by Physiotherapy Dept., The Royal Hospital for Women, Paddington, NSW and a University of Sydney Research Grant.

# FUNCTION OF INTRA-ABDOMINAL PRESSURE IN BALLISTIC AND NON-BALLISTIC ISOMETRIC FORCE EXERTION OF THE TRUNK

Kazuyoshi Gamada *1, Masami Harimoto *1, Toru Fukubayashi *1, Hiroyuki Nakajima *2

*1 : Department of Life Sciences, University of Tokyo, 3-8-1 Komaba, Meguro-ku, Tokyo 153, Japan
*2 : Research Institute of Physical Education, Nippon Sport Science University, 7-1-1 Fukazawa, Setagaya-ku, Tokyo 158 Japan

## INTRODUCTION

Intra-abdominal pressure (IAP) is regarded as one of the protective mechanisms of the spine against mechanical stresses. Marras et al. (1985)[1] reported the relationship between trunk extension strength and IAP during standing lifting motion using an isokinetic dynamometer. He also mentioned IAP-torque onset delay (ITOD), which is the fact that there is a delay in the onset of exerted torque relative to that of IAP. The aim of this research is to investigate the function of intra-abdominal pressure (IAP) during isometric force exertion of the trunk, with a special focus on the difference between ballistic and non-ballistic actions.

## METHOD

The volunteered subjects are composed of 14 healthy male adults without trunk pain or dysfunction. Trunk strength was tested in the chair sitting position with the pelvis and lower extremities being fixed, and IAP was measured intrarectaly with a catheter-type pressure sensor. The tasks were maximal and submaximal isometric trunk flexion, extension, and right and left rotations with and without ballistic action

## RESULTS.

In a comparison of the trunk strength, the extension force was significantly higher ($p<0.005$) than flexion and rotations, whereas flexion was significantly lower ($p<0.05$) than rotations. Meanwhile, no difference was found between left and right rotations. In a comparison of IAP, higher IAP was observed during extension than flexion ($p<0.005$) and during left rotation than right ($p<0.01$), whereas no difference was found between rotations and extension or flexion. In the non-ballistic condition, strong correlation was found between IAP and extension strength ($r=0.767$), right rotation ($r=0.774$) and left rotation ($r=0.799$), whereas a weak correlation was found between IAP and flexion ($r=0.270$) (Fig. 1). Similarly, in the ballistic condition, strong correlation was found between IAP and extension ($r=0.796$), right rotation ($r=0.763$) and left rotation ($r=0.738$), whereas a moderate correlation was found between IAP and flexion ($r=0.515$) (Fig. 2). In IAP-torque onset delay, IAP curve was advanced to the force curve in all of the trunk motions, and time delay in rotation was longer ($p<0.05$) than flexion and extension.

## DISCUSSION

Consequently, it is hypothesized as follows. (1) IAP contributes to the stabilization of the spine during force exertion of the trunk. (2) Maximal trunk force might be limited by the maximal IAP. (3) IAP is an inevitable factor for stabilization and protection of the spine during force exertion of the trunk.

**Fig. 1 : Relationship between non-ballistic isometric trunk flexion / extension**

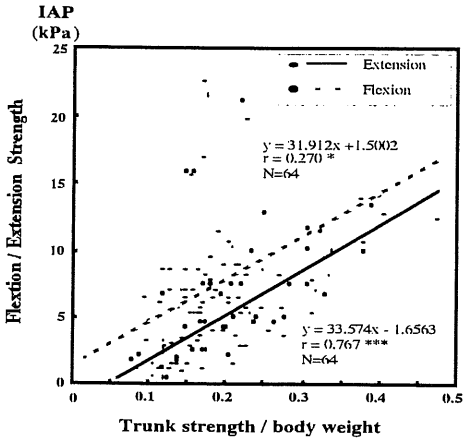
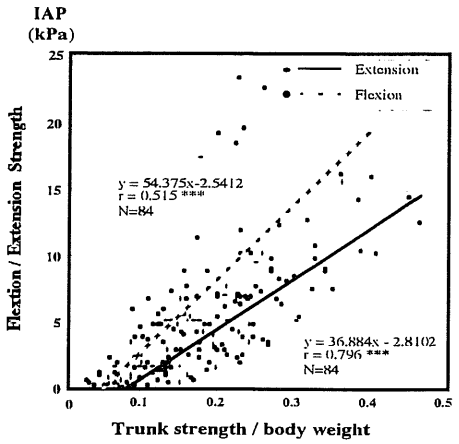



**Fig. 2 : Relationship between ballistic isometric trunk flexion / extension**




## REFERENCE

1) Marras, W. S., Joynt, R. L., and King, A. I., 1985, The force-velocity relation and intra-abdominal pressure during lifting activities. Ergonomics, 28, 603-13.

# BIOMECHANICS OF ABDOMINAL BELTS

Evaluation of the effect of abdominal belts on a cross sectional image of the trunk using the CT scan

K. Miyamoto[1], S. Ueki[2], N. Iinuma[3], A. Kaneko[4], Y. Miyata[1]

[1]Department of Orthopaedic Surgery, Takayama Red Cross Hospital, Takayama, Japan

[2]Department of Orthopaedic Surgery, Kizawa Memorial Hospital, Minokamo, Japan

[3]Department of Orthopaedic Surgery, Gifu Red Cross Hospital, Gifu, Japan

[4]Department of Radiology, Takayama Red Cross Hospital, Takayama, Japan

## INTRODUCTION

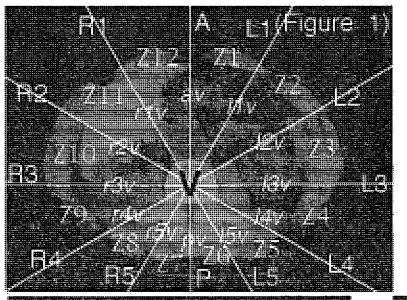
The biomechanical effects of abdominal belts in lifting are controversial. Intra-abdominal pressure, activities of trunk muscles and the like had been measured in recent studies. However, the effects of the belts on the motions of abdominal wall are not well understood. This study was designed to evaluate the effects of belts during Valsalva maneuver on cross sections of the trunk.

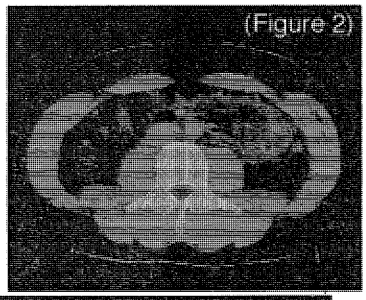
## METHODS

Twelve asymptomatic young male subjects participated in this study. CT scanned images (Toshiba, TSX-011A/40) of the trunk at the level of the 3rd lumbar spine were obtained in 8 conditions (with or without deep-inhalation, with or without full-Valsalva, with or without a belt) (Figure 1,2,3,4).
A weight-belt made of leather for the use of weightlifters, 10 cm of its width, was used.
Data were analyzed on a Macintosh model computer, using the public domain "NIH Image" program.
Twelve lines were drawn from the center of the 3rd lumbar vertebral body (V), each at 30 degrees.
The points at which the lines reach the skin were plotted (A, R1, R2, R3, R4, R5, P, L1, L2, L3, L4, L5).
The distance between the points and "V" , and areas of 12 separated zones were measured (Figure 1).

## RESULTS AND DISCUSSION

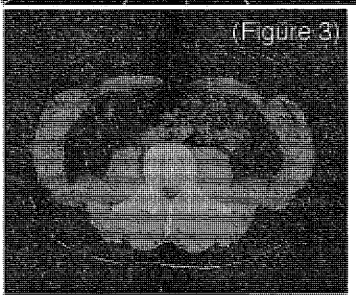
Geometrical measurement of the CT scanned images demonstrated that, during Valsalva maneuvers, wearing the abdominal belt makes the cross section of the trunk nearly round (Figure 4, Table).
This is the first report of an analysis of a cross section of the trunk during Valsalva maneuver to assess the effects of abdominal belts. The postero-lateral region of the trunk was supposed to be compressed by the tension of the belt during Valsalva maneuver (Table) with the increased activation of the rectus abdominis muscle[1]. It may also be suggested that intra-muscular pressure in the erector spinae is increased by the belt, which was measured in our previous study[1]. This finding may provide a clue to understanding the biomechanical effect of abdominal belts on stability of the trunk.

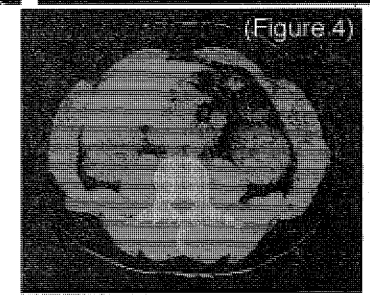



(Figure 1) Measurements using NIH-Image.
CT image at L3 level with deep inhalation, with full-Valsalva, without belt.

(Figure 2) without inhalation, without-Valsalva, without belt.




(Figure 3) without inhalation, without-Valsalva, with belt.

(Figure 4) with deep-inhalation, with full-Valsalva, with belt.
CT image shows the belt makes the cross section of the trunk nearly round.

| | without-belt (cm) | with-belt (cm) | with/without ratio |
|---|---|---|---|
| AV | 13.4±1.4 | 12.7±1.0 | 0.95±0.04 |
| L1V | 13.5±1.7 | 12.7±1.1 | 0.94±0.08 |
| L2V | 14.0±1.3 | 13.1±1.0 | 0.94±0.05 |
| L3V | 14.4±1.0 | 12.8±0.9 | 0.89±0.03 |
| L4V | 12.5±1.5 | 10.8±0.9 | 0.87±0.05 |
| L5V | 9.3±0.7 | 9.0±0.6 | 0.96±0.04 |
| PV | 7.7±0.6 | 7.9±0.5 | 1.02±0.03 |
| Average | 12.1±2.6 | 11.3±2.1 | 0.94±0.05 |

(Table) Mean and standard deviation of each distance.
Both conditions are with deep-inhalation and with full-Valsalva.

## REFERENCE

1) Miyamoto, K., Iinuma, N. et al. Biomechanics of abdominal belts (Evaluation of the effectiveness of abdominal belts on intra-abdominal pressure and intra-muscular pressure in the erector spinae muscles). In the proceedings of 20th Annual meeting of the American Society of Biomechanics, 1996.

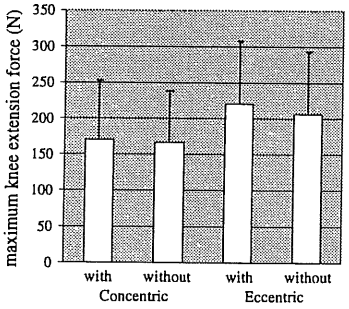
# DIFFERENCES IN BRACED AND NONBRACED MAXIMUM CONCENTRIC AND ECCENTRIC KNEE EXTENSION STRENGTH IN SUBJECTS WITH PATELLOFEMORAL JOINT PAIN

Tammy M. Owings and Mark D. Grabiner
Department of Biomedical Engineering
The Cleveland Clinic Foundation
Cleveland, Ohio, USA.

**Introduction**: The use of orthotic appliances unarguably plays an important role in the rehabilitation of certain families of injuries. The literature related to the use of orthotics, while abundant relative to the immediate effect of application on simple and complex motor performance, yields no evidence of the intermediate and long term effects. The intent of this study was to determine if the effects of a rehabilitation program using an orthotic device has immediate carry-over to the period following diminished or discontinued use of the device. Specificity of functional improvement to conditions in which the orthotic is used may present conditions in which the likelihood of re-injury is increased.

**Methods**: Twenty male and females with patellofemoral pain were recruited into the study and divided into two experimental groups. Each performed a four week training protocol with a different commercially available patellar stabilization orthotic (Bauerfiend P3 and DonJoy). The training protocol, which emphasized accurate matching of submaximum isokinetic concentric and eccentric (15 $deg*s^{-1}$) knee extension force to a target value through a 50° range of motion, has been previously reported to result in significantly improved maximum concentric and eccentric knee extension strength in a group of healthy older adults (Owings and Grabiner, 1995). Subjects trained twice a week for four weeks. Real-time visual feedback was provided at target force values of 15,30,45, and 60% of the pre-training maximum isokinetic concentric and eccentric knee extension force (MVC). The instructions to the subjects were to match as closely as possible the generated force to the target level through the entire range of motion. In addition to the accuracy with which these forces could be controlled before and after the four week protocol, concentric and eccentric isokinetic knee extension MVC was measured as the peak force generated through the range of motion. The concentric-eccentric MVC data were initially subjected to a 2 by 2 by2 (group by test by contraction) repeated measures ANOVA.

**Results**: The ANOVA revealed that the differences between the subjects using each brace were not significant ($p=0.84$), thus the data was collapsed across subject groups. The subsequent ANOVA revealed significant interactions between pretest-posttest and contraction type ($p=0.036$) and between contraction type and brace condition ($p=0.029$). Bonferoni-adjusted multiple comparisons revealed that the former interaction was associated with a significant increase in eccentric knee extension MVC but no change in concentric MVC. Eccentric MVC was significantly larger than the concentric MVC for both the pretest and posttest. More important were the results of the *a posteriori* multiple comparisons of the latter interaction (see Figure) that revealed that the concentric MVC was not significantly affected by the brace condition ($p=0.393$) but eccentric MVC was significantly smaller during the non-braced condition ($p=0.029$).




**Discussion**: It is notable, that the short-term training protocol was effective in increasing eccentric, but not concentric, knee extension MVC in subjects with patellofemoral pain. This is consistent with previous findings that demonstrated an acute sensitivity of eccentric MVC to *very* short-term training protocols (Owings and Grabiner, 1996). Relative to the primary purpose of the study, the observation that eccentric (but not concentric) MVC was significantly diminished in the absence of the orthotic suggests an initial functional deficit that endures in spite of significant protocol-related improvement. This raises concern as to when rehabilitation can actually be considered complete.

## References

Owings, T.M., Grabiner, M.D. Submaximum stimulus intensity increases maximum strength in elderly. Proceedings of the American Society of Biomehanics, Stanford University, 24-26 August, 1995, pp.261-261.

Owings, T.M.,Grabiner, MD: Disparate ipsi- and contralateral effects of concentrically vs. eccentrically-induced fatigue. *Medicine and Science in Sports and Exercise*, 28:S141, 1996.

## Acknowledgments

The authors gratefully acknowledge Bauerfiend, Incorporated for their support of this work

# EFFECTS OF STRENGTH TRAINING AND DETRAINING ON MUSCLE STRENGTH AND FORCE-TIME VARIABLES IN PATIENTS WITH RECENT ONSET INFLAMMATORY ARTHRITIS

A. Häkkinen[1] , P. Hannonen[2], I. Jäppinen[1], L. Laitinen[1], and K. Häkkinen[3]
[1]Department of Physiatry and [2]Medicine, Central Hospital, Jyväskylä
[3]Department of Biology of Physical Activity, University of Jyväskylä, Jyväskylä, Finland

## INTRODUCTION

Decreased muscular strength in patients suffering from rheumatoid arthitis (RA) (Nordensjö et al.1983, Danneskiold-Samsoe et al. 1986) is thought to be due to inflammation-related processes in the joints, muscles and in periarticular connective tissues, which lead to disuse and thereby functional and structural changes in the neuromuscular system. The purpose of this study was to: 1) compare maximal strength of various muscle groups and the force-time curve of the knee extensors in patients with recent onset RA to those of healthy people; 2) examine effects of strength training and 3) detraining on isometric and dynamic force production chracteristics of various muscle groups in a group of RA patients.

## METHODS

Thirty-nine patients with recent onset RA were divided into experimental (PE; 41.4; 26-55 yrs; 11 men and 10 women) and control groups (PC; 45.6; 24-57 yrs; 8 M and 10 W). Eighteen healthy persons (H; 41.8; 24-57 yrs; 9 M and 9 W) volunteered to act as controls. The study was approved by the Ethics Committee of the hospital. PE carried out strength training (ST) for 6 months (0-6),while PC maintained habitual physical achtivities. ST in PE took place 2-3 times a week for major muscle groups with loads of 40-80 % of the repetition maximum. Thereafter, PE maintained habitual physical activities for 3 years (detraining; DT; 6-42). PE and PC received antirheumatic medication. Maximal unilateral 1RM and maximal bilateral isometric force-time variables of the knee extensors were measured by the David 200 (Häkkinen et al. 1987), maximal isometric force of the trunk flexors and extensors by the Digitest,and maximal grip strength of both hands by the Jamar Standard dynamometers. Ritchie (Ritchie et al. 1968) articular index (RI) and erythrocyte sedimentation rate (ESR) were used for the estimation of disease activity.

## RESULTS

At 0 knee extension 1RM per body mass was lower in PE and PC ($p<0.01$) than in H. An increase of 37% ($p<0.001$) occurred in PE during ST followed by a decrease of 11% ($p<0.05$) during DT and at 42 PE showed greater ($p<0.05$) values than PC. Isometric knee extension forces did not differ until at 42 when PE and PC showed 14% and 24% ($p<0.05$) lower values than H. The force-time curve showed at all tests lower ($p<0.05$) values for PE and PC than for H. At 0 grip strengths were in PE and PC lower ($p<0.001$) than in H and remained lower, although increases of 16% ($p<0.001$) and 11% ($p<0.05$) occurred in PE and PC during ST. Trunk extension and flexion forces were at 0 in PE and PC slightly (ns.) lower than in H. ST in PE led to respective increases of 10% ($p<0.01$) and 13% ($p<0.01$) followed by the decreases of 17% ($p<0.01$) and 14% ($p<0.01$) during DT. PC showed decreases of 10% ($p<0.05$) and 10% ($p<0.05$) between 0 and 42. At 42 trunk forces were in PE 15% and 10% and in PC 16% and 19% lower ($p<0.05$) than in H. RI improved in PE during ST from 11.2±1.3 to 4.9±1.1 ($p<0.001$) and in PC from 13.4±2.0 to 7.0±1.7 ($p<0.05$) remaining in both groups unaltered at 42. ESR decreased in PE from 25.3±5.4 to 14.0±4.1 ($p<0.05$) and increased slightly (ns.) to 18.2±3.4 at 42, while the values in PC were 22.1±2.4, 16.7±2.7 (ns.) and 21.1±4.5, respectively.

## DISCUSSION

Lowered dynamic strength and shape of the isometric force-time curve of the knee extensors in the middle-aged patients with recent onset RA indicate that the loss of functional capacity seems to occur already in the early phase of disease. This is probably due to the decrease in the intensity of daily physical activities with reduced neural activation, muscle fibre characteristics and/or joint tenderness leading to avoidance of various strength tasks of muscles. Grip strength in PE and PC was lower than in H, since the disease usually begins from the small joints of the hands and feet. ST for six months in both M and W of PE led to great increases in strength of all muscles examined even back to that of healthy people except for grip strength without observable negative effects on disease activity. The strength gains in PE were lost only to a selective degree during a prolonged period of "detraining", whereas in PC strength of the trunk extensor and flexor muscles decreased continuously throughout the experiment. The force-time curve of the knee extensors in PE remained unaltered during ST and both patient groups showed lower explosive force values than H over the entire experiment. The present results suggest that patients with RA seem to need continuous physical exercise including also exercises for maintaining and/or improving both maximal and explosive strength characteristics of their muscles.

## REFERENCES

Danneskiold-Samsoe, B. et al. Clin. Rheumat. 5: 459-467, 1986.
Häkkinen, K. et al. Med. Sport Sci., Kargel, 224-237, 1987.
Nordensjö, L. et al. Scand. J.Rheumatol. 12: 152-156, 1983.
Ritchie, D. et al. Q.J.Med. 37: 393-406, 1968.

# ANKLE SPASTICITY AND STRENGTH IN CHILDREN WITH SPASTIC DIPLEGIA CEREBRAL PALSY

S.A. Ross*, J.R. Engsberg*^, K.S. Olree*, T.S. Park^
*Human Performance Laboratory, Barnes-Jewish and St. Louis Children's Hospitals, ^Washington University School of Medicine, St. Louis, MO, USA.

## INTRODUCTION

Spasticity and lack of strength are two problems associated with children with cerebral palsy (CP). Recently; objective measures to quantify spasticity (i.e., velocity dependent resistance to stretch) and strength (i.e., peak torque and maximum work) in the hamstrings of children with CP have been reported.[1,2] However, knowledge regarding ankle plantarflexor spasticity and strength is also critical when evaluating children with CP. It has been reported that distal areas of the lower limb are often more affected than proximal areas.[3] The purpose of this investigation was to quantify the spasticity and strength in the ankle plantarflexors in children with spastic diplegia.

## METHODS

Both legs of 14 children (mean 8 years, range 4-17) with spastic diplegia were tested on a KinCom dynamometer. Each child sat on the KinCom and had ankle joint range of motion limits established. For the spasticity measure, the machine rotated the passive ankle from maximum plantarflexion to maximum dorsiflexion at speeds of 10, 30, 60, 90, and 120°/s while recording the resistive plantarflexor torque. Torque-angle data were processed to partial out effects of gravity and minimize acceleration and machine dynamic responses. Areas within the torque-angle curves were calculated for each speed yielding 4 work values. Linear regression was used to derive the slope of the line of best fit for the work-velocity data.[1] For the strength measure, the machine rotated the ankle from maximum dorsiflexion to maximum plantarflexion at a speed of 10°/s while the child performed a maximum concentric contraction. Peak plantarflexion torque and work by the plantarflexors over the entire range of motion were calculated.[2] Values for the right and left legs were averaged. Unpaired t-tests were used to test for significant differences ($p<0.05$) between the children with cerebral palsy and an able-bodied control group (n=7).

## RESULTS

Spasticity in the plantarflexors for the children with spastic diplegia was significantly greater than the able-bodied norms. Peak plantarflexor torque and maximum plantarflexor work were significantly less than the able-bodied norms.

Table 1. Means and standard deviations for strength and spasticity. *Significantly different from able bodied.

| Group | Spasticity [J/(°/s)] | Peak Torque [Nm/kg] | Maximum Work [J/kg] |
|---|---|---|---|
| Spastic Diplegia | 0.050* | 0.59* | 0.28* |
| (SD) | (0.111) | 0.40 | 0.27 |
| Able Bodied | 0.015 | 2.12 | 1.50 |
| (SD) | (0.013) | 0.69 | 0.45 |

## DISCUSSION

A previous study has indicated that the average spasticity at the hamstrings [0.033 J/(°/s), SD 0.019] for a similar group of children with spastic diplegia (n=17) was less than the spasticity reported here for the ankles.[1] The strength measures for peak knee flexion torque and maximum knee flexion work for a similar group of children were 63% and 43% of able-bodied norms, respectively.[2] At the ankle, similar values were 28% and 19%, respectively. The greater spasticity and weakness at the ankle support statements indicating a greater level of involvement distally.[3]

Surgical and therapeutic procedures are performed on these children to reduce the effect of ankle spasticity (e.g., selective dorsal rhizotomy, heel cord lengthenings, serial casting). It is generally assumed that a weakening of the plantarflexors is a necessary, but undesirable limitation of the procedures. However, objective data investigating these assumptions do not exist. The methods described in this investigation could provide evidence to support or refute the assumptions and assist medical professionals in defending or modifying their techniques based upon objective outcomes. Quantifying these impairment measures pre- and post-surgery, developing similar techniques for hip adductors, and relating to functional outcomes are important future considerations.

## REFERENCES

1. Engsberg JR et al. 1996. Quantitative clinical measure of spasticity in children with cerebral palsy. *Arch Phys Med Rehabil* 77:594-599.
2. Engsberg JR et al. 1997. Maximum active resultant knee joint torques in children with spastic diplegia. *J Applied Biom* In Press.
3. Gage JR. 1991. *Gait Analysis in Cerebral Palsy.* MacKeith Press, London.

# THE FEASIBILITY OF USING ACCELEROMETRY IN THE ASSESSMENT OF MUSCLE STRENGTH

William W.N. Tsang, Kevin S.C. Kwong
Department of Rehabilitation Sciences, Hong Kong Polytechnic University, Hong Kong.

## INTRODUCTION

Assessment of muscle strength is one of the interest of Physical therapists. Attentions have been focused on the isometric and isokinetic muscle strength. However, there is lacking instrumentation for the measurement of muscle strength during natural physiological motion (Seger et al 1988). Also, the means of measuring isometric and isokinetic strength is either subjective, not accurate or is too expensive. The accelerometer is inexpensive and its use is focused on the measurement of human movement (Lafortune & Henning 1991).

## AIM

A method which applies accelerometry was tested of its feasibility in measuring the muscle strength of quadriceps during natural physiological motion.

## METHOD

The study began with an investigation of the performance of a mechanical model comprising an accelerometer mounted on a swinging rod which was then attached to a stable frame. The measurement was then carried out on the dominant legs of 18 subjects. The linear acceleration of a point at the distal end of the tibia, and with known distance from the knee joint was measured by a piezo-electric accelerometer attached at the described site during a natural knee extension at the subject's full effort. The reliability of the set-up was tested. A mathematical model, incorporating published data on body segment parameters (Seireg & Arvikar 1989) was established for the estimation of maximal knee extension torque. The estimated peak torque was then compared with a slow speed mode (60 degree/s) and a fast speed mode (180 degree/s) of Merac, an isokinetic dynamometer.

## RESULT

The coefficient of variation was 4.3 %. The intraclass correlation coefficient of the accelerometer was found to be 0.90. The correlation coefficient with isokinetic measurement for slow speed mode and fast speed mode were 0.74 and 0.79 respectively ($p < 0.001$).

## DISCUSSION

The implication of the result is that there is high feasibility of using accelerometry in the development of a new device which can be used for measuring muscle strength in a natural physiological motion in the clinical situation. The muscle strength obtained from the accelerometer correlates more with the fast speed mode of the isokinetic measurement because the natural physiological movement is of the fast movement type. The curve measuring the acceleration and deceleration of limb was also obtained during the investigation which can be used to study the pattern of limb movement and calculate the eccentric muscle strength by the antagonistic muscles. Therefore, further study on the natural physiological movement is highly recommended.

## REFERENCE

LAFORTUNE,M.A., HENNIG,E.M. (1991) Contribution of angular motion and gravity to tibial acceleration. Medicine and Science in Sports and Exercise;23;pp360-363.
SEIREG,A., ARVIKAR,R. (1989) Biomechanical analysis of musculo-skeletal structure for medicine and sports. pp47-97:Hemisphere:Pub. Corp.:New York..
SEGER,J.Y. WESTING, S.H., HANSON,M., KARLSON,E., EKBLOM,B. (1988) A new dynamometer measuring concentric and eccentric muscle strength in accelerated, or isokinetic movements. Eur. J. Appl. Physiol;57;pp526-530.

# THE EXTERNAL POWER OUTPUT IN SIMULATED MOVEMENTS FOR RUNNING AND THE MOVEMENTS OF LOWER EXTREMITIES IN SPRINT RUNNING

Matsuo, A., Y. Kaku, H. Kintaka, and H. Hirata
Laboratory of Sports Biomechanics, National Institute of Fitness and Sports in Kanoya
1 Shiromizu, Kanoya, Kagosima, JAPAN

## INTRODUCTION

The ability of a sprinter to accelerate the body quickly is considered to be caused by high power output of muscle contraction of the knee and hip extension, and the velocity of sprint run. The running performance is too complicated to be able to explain briefly the power output produced by the hip and the knee extension movement. The purpose of this study was to investigate the effect of mechanical power output in simulated movements of sprinting on the movements of sprint running.

## METHODS

The subjects of this study were 11 male collegiate athletic sprinters whose best performance in the 100m dash ranged from 10.7 sec to 11.9 sec. The subjects did a 60m dash with maximum effort, and the five different movements as described below.

In order to measure the mechanical power displayed in various movements, the specially designed apparatus called a power processor was used (Fukunaga, et al, 1992). After warm up exercises, the subjects were requested to perform the following 5 movements with maximum effort.: 1. MOS:stand up, elevate the knee joint to the point at which the hip joint is flexed; 2.KIS:stand up, kick from back to front; 3.KF:flex the knee joint; 4.KE:extend the knee joint; 5.HIC:grip the handle with both hands, extend the ankle, knee and hip joints.

A video camcorder (SONY DXC-537) was used for videotaping, with the frame rate set at 60 Hz and the shutter speed at 1/500 sec, and it was secured on a tripod and positioned so that the focal axis of the lens was perpendicular to the plane of motion. In order to identify factors of the running performance, the following kinematics parameters of lower extremity during sprint running were chosen on the basis of report of men's sprint running studies(Ito, 1994): $\theta_{MOS}$, defined as the maximum value of the angle of the thigh in the recovery phase; $\theta_{FD}$, defined as the maximum value of the angle between the line connecting the hip joint and ankle joint and the vertical line; $\theta_{FB}$, defined as the minimum value of the angle between the line connecting the hip joint and ankle joint and the vertical line.

## RESULTS AND DISCUSSION

The running speed indicated the highest value of 10.36 m/s in the 60 m dash. The external power output of MOS in the right lower extremity related significantly to the running speed in 60m dash ( $r = 0.89$, $p<0.001$). On the otherhand, the value of MOS in the left lower extremity bore no significant relation to that. Other power output values of other movements did not relate significantly to performance time in the 100m dash. There were significant differences in power output between the right and left side lower extremity, in MOS, KIS, KE and KF. The values of mechanical power output of the right side were larger than the left side except for KE movement.

The kinematics parameters ( $\theta_{MOS}$, $\theta_{FD}$, and $\theta_{FB}$ ), related significantly to performance time in the 60m dash. The movements of the lower extremities of faster sprinters showed a wider range of $\theta_{MOS}$, $\theta_{FD}$, and $\theta_{FB}$ than those of slower sprinters. There were no differences between the right and left lower extremities in the angles of all three parameters. These results suggest that the relative loading on the lower extremities during sprint running was not the same for the right and left extremities.

## REFERENCES

Fukunaga, T., et al. Measurement of movement power in relation to a performance of sprint run. Journal of Sports Medicine & Science, 6(2), 19-26, 1992.

Ito, A., et al. Biomechanical analysis of world top sprinters, How they ran, jumped, and threw, 31-49, H. Sasaki, et al(ed.), Baseball magazine, 1994.

# BODY SEGMENTAL FUNCTIONS IN THE SUPPORT PHASE OF SPRINTING.

K. Ogiso[1], T. Yasui[2], K. Aoyama[3] and A. Kushima[4]
1. Toba National College of Maritime Technology, Toba, Japan. 2. Aoyama Gakuin University, Atsugi, Japan. 3. Kokushikan University, Tama, Japan. 4. Miyazaki Prefectural Nursing University, Miyazaki, Japan.

## INTRODUCTION

The purpose of this study was twofold : (a) to investigate the functions of the segmental motions on the velocity of the body center of mass (CM) in the support phase of sprinting, and (b) to find out whether their functions change with the change of sprinting speed.

## METHODS

Ten male sprint runners (age 20.7±1.2yr, height 1.72±0.06m, body mass 63.3± 4.7kg) participated in this study. The subjects performed three sprints from the starting blocks. They were instructed to execute a sprinting as fast as possible. Their movement patterns were analyzed for five support phases : (S1) at 77.4±3.1% of maximum sprinting speed (MSS) during the acceleration phase, (S2) at 95.4±1.9% of MSS during the acceleration phase, (S3) at MSS, (S4) at 94.1±1.4% of MSS during the deceleration phase, and (S5) at 90.8 ±2.2% of MSS during the deceleration phase. The computation of mechanical variables was identical to that described by Ae and Shibukawa (1980). The variables are momentum, impulse and force generated by each segment. These were obtained for the right arm, left arm, trunk, free leg, and the thigh, shank and foot of the support leg.

## RESULTS AND DISCUSSION

Each body segment made a stereotyped segmental force pattern. The patterns of the thigh, shank and foot were characterized by the following : (1) The forces generated by three segments attained their maximum and minimum in order; (2) While the lower segments generated the upward forces, the upper segments generated the forward forces; (3) The forward and upward large forces weren't simultaneously generated by each segment; and (4) The adjoining segments tended to generate the inverse forces one another. These characteristics remained unchanged despite the change of the sprinting speed. The force magnitude and the real support times, however, tended to change with the speed. The force pattern generated by the foot in the second half differed from the observation above. This may be attributed to the momentum transfer from the proximal segments to the distal segments, and/or the reduction of the load on the foot due to the forward movement of the CM. These results show that the foot always uphold the whole body, and the thigh in the first half and the shank in the initial phase accelerate the CM forward. It is assumed that the inverse actions between the shank and the foot in the first half have the effect as a compensation for the loss of the sprinting speed.

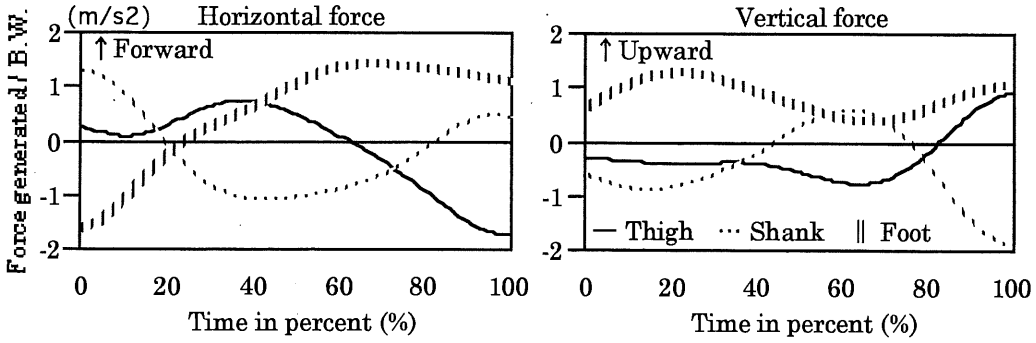



Fig.1 Mean curves of the forces generated by three segments of the support leg at S3.

## REFERENCES

Ae,M. and Shibukawa,K.(1980) A biomechanical method for the analysis of the contribution of the body segments. Japanese Journal of Physical Education, 25, 233-243.
Jacobs R. and Ingen Schenau van G.J.(1992) Intermuscular coordination in a sprint push-off. Journal of Biomechanics, 25, 953-965.

# A LONGITUDINAL STUDY ON THE DEVELOPMENT OF RUNNING MOTION OF BOYS AGED SEVEN TO FIFTEEN

K. Katoh [1], M. Miyamaru [2] and M. Ae [2]
[1] Utsunomiya University, Utsunomiya, Japan. [2] University of Tsukuba, Tsukuba, Japan.

## INTRODUCTION

Running is one of fundamental human movements and can be mastered, by the age of two years for most children unless they are handicapped. Afterwards, the quality and quantity of their running will steadily improve with maturation as well as learning until the entrance of elementary schools. Although many studies on the development of running motion and performance for children and adolescents have been done, most of them were carried out in cross-sectional methods. The present study was designed to investigate longitudinally the development of running motion, running speed, step length, and step frequency from elementary through junior high school boys.

## METHODS

Subjects were twenty healthy boys aged seven to fifteen years. Their running motion in 50 m sprint with maximal effort was videotaped with a camera operating at 60 fps. Body segment parameters of the subjects were estimated from Yokoi's body segment parameters(1985). Kinematic parameters to analyze running motion were running speed, step length, step frequency, displacement of the whole body's center of gravity (CG), average ascending angle of CG from the lowest to the highest point, and angular kinematics.

## RESULTS AND DISCUSSION

Running speed increased from 5.15 to 7.73 m/s with age. Step length, step length/body height ratio, and non-support time/support time ratio also increased, whereas step frequency decreased with the advancing age. Yearly increase of running speed were and positively correlated with those of body height observed from 10 to 12 years of age and step frequency from 9 to 15 years of age. Furthermore, yearly increase of running speed were correlated with the step length and the step length ratio at the age of 7 and 8 years and from 11 to 13 years of age respectively. However, negative correlations were found between yearly increases of running speed and those of support time from 11 to 15 years of age. The partial correlations found between running speed and the CG motion suggest that improvement in running speed with age may be resulted from shortening support time, increasing step frequency, and decreasing the bouncing movement of the CG. Figure 1 shows changes of running speed from 7 through 15 years of age for the boys studied. All the boys developed their running speed although some had fluctuations. In addition, a significant order correlation was found between running speed at the age of 7 and 15 years of age. Running speed developed by some best performers at the age of 9 years was almost same as the speed of those who were in the lowest level at the age of 15 years. The running motion of the 9-year-old best performers were high level, while that of the 15-year-old lowest ones were poor level in the same age groups. Therefore, it was suggested that the running motion could not necessarily be developed only with the maturation.

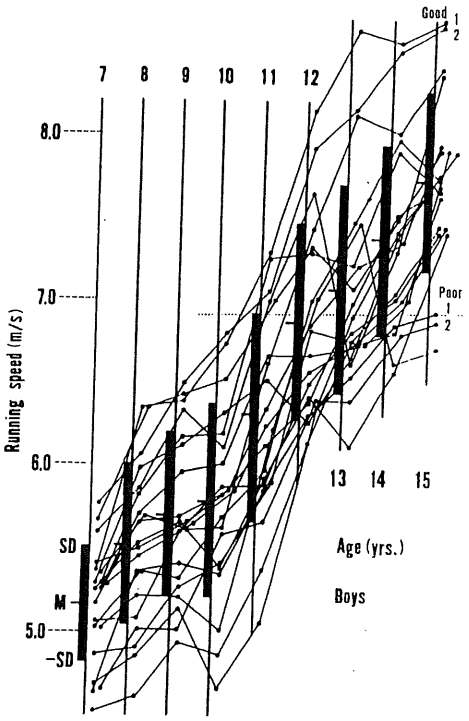



Figure 1 Longitudinal development of running speed from 7 through 15 years of age.

# CHANGES OF RUNNING SPEED, STEP FREQUENCY AND STRIDE LENGTH DURING 800M RACE.

Masaaki. SUGITA*, Akifumi. MATSUO**, Kando. KOBAYASHI*, and Michiyoshi. AE***
* Department of Sports Sciences, Graduate School of Arts and Sciences, The University of TOKYO, 3-8-1 Komaba, Megruro-ku, Tokyo 153 Japan
** Department of Physiology and Biomechanics, National Institute of Fitness and Sports, KANOYA
*** Department of Sports Sciences, University of TSUKUBA

## INTRODUCTION

Although there have been some reports about running speed, step frequency and stride lengh during actual competitive track race such as middle and long distance event, they have not been enough as a rich store of findings. The purpose of this study was to analyze running speed, step frequency and stride length during competitive 800m race including Olympic games and to investigate the mechanism of running speed changes from the view point of step frequency and stride lengh.

## METHODOLOGY

The subjects of this study were the 40 athletes who competed men's final 800m race during the '91World Championship (Tokyo), '92, '93 Japan Championship (Tokyo), '94 Asian Games (Hiroshima), '96 Olympic games (Atlanta). With the aid of four home video cameras, the difference between the pass time of each athlete for every 100m was recorded in order to calculate speed in meters per second. Four another home video cameras were installed between each mark to study further the step frequency of each athlete. From these, a periodic analysis of every ten steps made within the points between each 100m was done in order to calculate the frequency of steps per second. Stride length was calculated from running speed and step frequency.

## RESULTS AND DISCUSSION

We analyzed running speed, step frequency and step length of all athletes in these race (Figure 1).

The results were as follows,

1) The maximum speed in 800m race appeared during the first 200m in many athletes.
2) The increased running speed after 500m was due more to an increase in step frequency than to stride length in the upper rank athletes.
3)Significantly (r=0.863) correlated with the ratio of change in running speed from 600-700m section to 700-800m section that in step frequency. Therefore the factor which contributes to maintaining and increasing the running speed was the increased step frequency rather than stride length in the second half of the race and the last spurt.

## ACKNOWLEDGMENT

This study was supported by the Japan Amateur Athletic Federation (JAAF) Biomechanics research Project Team.

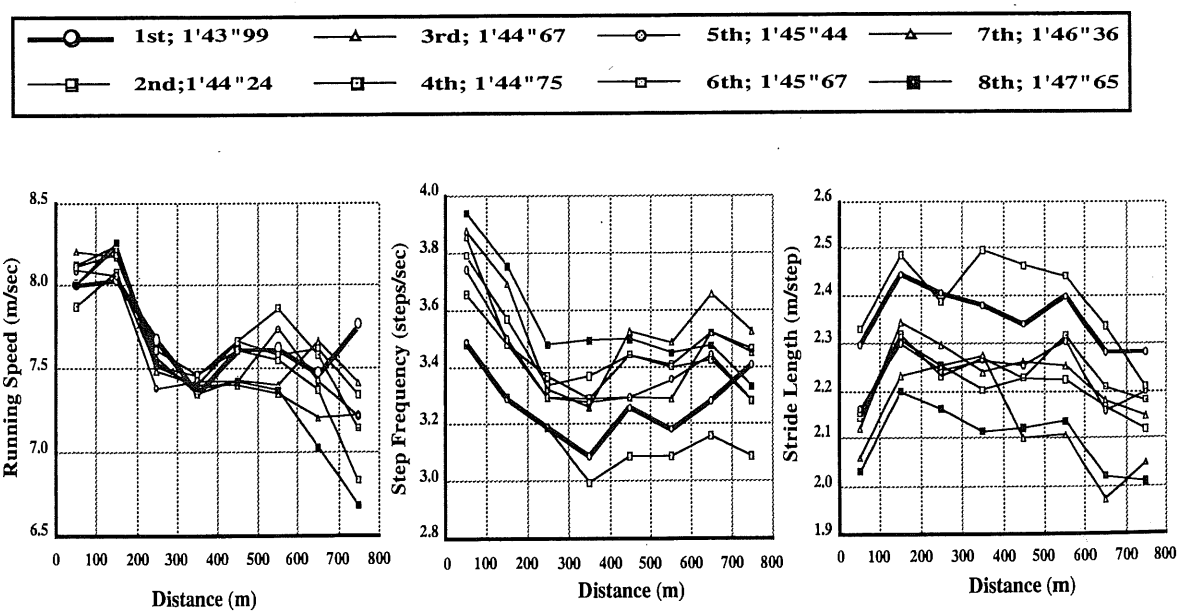



Figure 1. Running speed, step frequency and stride length at 100m interval for each athletes during '91World Championship (Tokyo).

# THE DESCRIPTION OF THE VARIABILITY OF GROUND REACTION FORCES DURING RUNNING USING MECHANICAL MODEL PARAMETERS

Harutoshi Yukawa
Aichi University, Toyohashi, Japan

## INTRODUCTION

There is a variability of ground reaction forces during running in one subject trials, called within-subject variability, even if the same task was given. The variability is often represented by standard deviations along the time series of movement. Nevertheless, it is impossible to know how it was produced such time sequence because these values are not considered as a mechanical constraint. According to our previous study (Yukawa and Kobayashi, 1995), the vertical ground reaction forces of running are well identified by viscoelastic model (Fig.1), so it may be considered that the variability can be reflected in the model parameters. In this study, a new description of the variability of ground reaction forces was proposed.

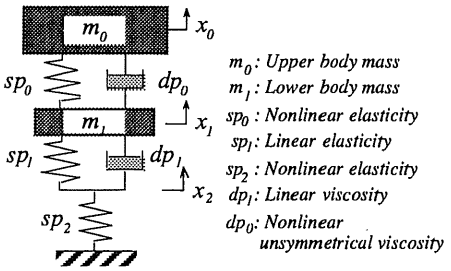



Fig.1 The viscoelastic model of the vertical component of the ground reaction force during running at a steady pace.

## METHOD

The vertical ground reaction forces of running for 20 trials were measured by running on the force platform at a steady pace (Fig.2). Simultaneously, the vertical component of the impact velocities and the rebound velocities of C.G. of the body were also measured by film analysis to use the parameter identification. A unique set of parameters of the model for each trial is obtained by nonlinear programming, reducing the sum of the relative standard error of ground reaction force and the relative error of rebound velocity. To consider the variabilities of each parameter as the variability of ground reaction forces, the mean and the standard deviation of each parameter is calculated.

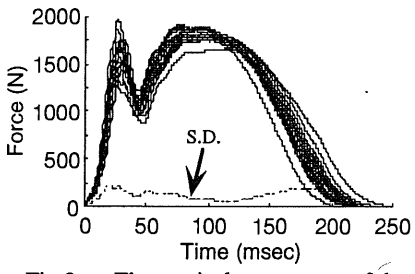



Fig.2 The vertical component of the ground reaction force and the standard deviation of 20 trials.

## RESULTS AND DISCUSSION

The nonlinear viscoelastic model in this study can estimate accurately not only the ground reaction forces but also the rebound velocities. Therefore, considering the variabilities of the parameters as that of the ground reaction forces is proper.

Fig. 3(a) shows the result of computer simulation using mean values of each parameter and (b) shows the one result of simulation using the random values of which the statistical properties, the mean and the standard deviation, is equal to that of identified parameters. These results suggest the possibility of describing the variability by a set of model parameters, if the set of parameters have a mechanical constraint to produce a proper result of simulation.

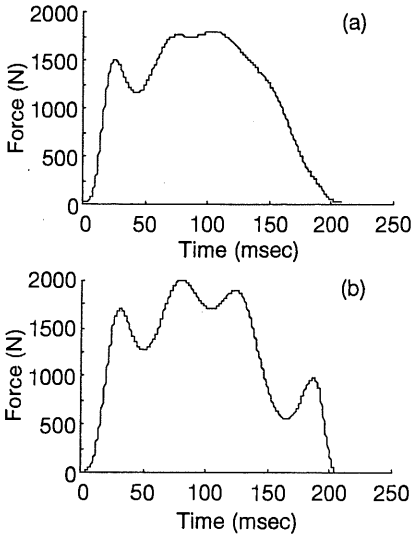



Fig.3 The result of simulation using the mean values of each parameter (a) and using the random values with statistical constraint(b).

# EFFECTIVENESS OF MECHANICAL ENERGY UTILIZATION IN LONG DISTANCE RUNNERS

Y. ENOMOTO[1], M. AE[2], and H. OKADA[1]
1)Doctoral Program of Health and Sport Sciences, University of Tsukuba
2) Institute of Health and Sport Sciences, University of Tsukuba

## INTRODUCTION

Long distance runners are required to maintain high running velocity for long time. It is important to utilize the mechanical energy effectively as well as maintaining high mechanical energy. But the number of investigations discussing the effectiveness of mechanical energy utilization for long distance runners is small, compared to those on mechanical efficiency. The purpose of this study was to investigate the effectiveness of mechanical energy utilization of long distance runners during 5000 m races.

## METHODS

5000 m races of official competitions were videotaped at 60 field/s with a VTR camera. Twenty-four male runners were selected as the subjects (race times: 13'20" - 17'20"). VTR images for each subject at 2000 m and 4000 m marks were digitized in one running cycle and smoothed to provide kinematic data. Total mechanical work ($W_{wb}$), in which energy transfers between and within segments were assumed, were calculated by equation (1) and (2).

$$E_{i,j} = m_{i,j}gh_{i,j} + \frac{1}{2}m_{i,j}v_{i,j}^2 + \frac{1}{2}I_{i,j}w_{i,j}^2 \qquad (1)$$

$$W_{wb} = \sum_{j=1}^{n-1}\left|\sum_{i=1}^{s}\left(\Delta E_{i,j}\right)\right| \qquad (2)\ \text{(Pierrynowski et al., 1980)}$$

where $E_{i,j}$ = total energy of the ith segment at time j, m = mass, g = acceleration due to gravity, h = height, v = horizontal velocity, I = moment of inertia about its center of mass, $\omega$ = angular velocity, n = total number of sample periods in one running cycle, s = total number of segments modeled, and $\Delta E_{i,j}$ = total energy change during the jth period of time for the ith segment.

To evaluate how much mechanical energy was utilized to running velocity, effectiveness index of mechanical energy utilization (EI) was calculated by equation (3).

$$EI = \frac{\frac{1}{2}MV_x^2}{\frac{W_{wb}}{CT}} \qquad (3)\ \text{(Ae and Fujii, 1996)}$$

where M = total body mass, $V_x$ = horizontal velocity of the center of mass, and CT = time for one running cycle.

And ratio of energy transfer between segments to total pseudo mechanical work ($\%T_w$) was also calculated (Pierrynowski et al., 1980).

## RESULTS AND DISCUSSION

Figure 1 shows the relationships of running velocity to EI and mean power ( $W_{wb}$ / CT / body weight ). A positive correlation was found between running velocity and EI, but no relationship between running velocity and mean power. These results indicate that effective utilization of mechanical energy is also important to maintain running velocity and to achieve good 5000 m race time. It is considered that EI is an useful index for the evaluation of the technique of long distance running. Figure 2 shows that relationship between EI and $\%T_w$. A high positive correlation between EI and $\%T_w$ suggests that runners with higher EI value transferred greater mechanical energy from segment to segment than the runners with lower EI value.

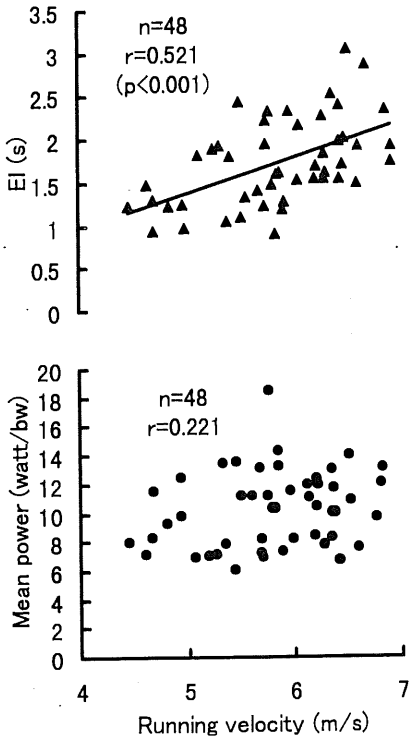



Figure 1 Relationships of running velocity to EI and mean power.

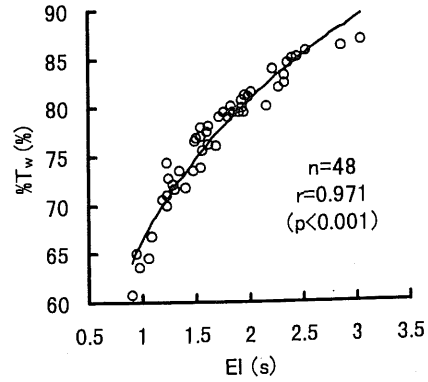



Figure 2 Relationship between EI and %Tw.

## REFERENCES

Ae, M. and N. Fujii (1996): Effectiveness of mechanical energy utilization and its index in human movement. *Bull. Inst. Health & Sport Sci. Univ. of Tsukuba* 19, 127-137.

Pierrynowski, M. R. et al. (1980): Transfers of mechanical energy within the total body and mechanical efficiency during treadmill walking. *Ergonomics* 23 (2), 147-156.

# OPTIMAL RUNNING ECONOMY AMONG ELITE RUNNERS MAY BE RELATED TO A LOW RELATIVE LEVEL OF MUSCLE ACTIVATION

Voigt,M Arnbjerg,H* Mogensen,HO* and Klausen,K*
Center for Sensory-Motor Interaction, Aalborg University. *Laboratory for Human Physiology, August Krogh Institute, Copenhagen. Denmark. Email: mv@miba.auc.dk

## INTRODUCTION

It has been demonstrated that running economy (RE) is dependent on a large variety of physiological, biomechanical and physical factors (Cavanagh and Kram 1985, Morgan and Craib 1992). However, the relationship between RE and neuromuscular factors among highly skilled runners has not been clarified in detail yet. The purpose of the present study was to compare the timing and the relative amplitude of muscle activation in lower extremity muscles during running in two groups of elite runners with significantly different RE.

## MATERIAL AND METHODS

RE was determined for twenty eight Danish male elite runners with a personal record below 33min on 10km. The six runners with the lowest RE (Ggroup) and the six runners with the highest running economy (Bgroup) were selected for EMG and kinematic analysis.

*Determination of RE*: Oxygen consumption and blood lactate was measured during steady state running at 14, 16 and 18 kmhr$^{-1}$ and at a velocity corresponding to the average velocity at their personal record on 10km (PRvel). Maximal oxygen uptake was measured at the end of the test. RE was determined as a 'RE-profile' i.e. the average of the REs from the four velocities expressed in ml min$^{-1}$ kg$^{-1}$ km$^{-1}$.

*EMG and kinematic analysis*: EMG and joint angles were recorded during steady state running at PRvel. The EMGs were recorded with surface electrodes from eight muscles of the right leg: tibialis anterior (TA), soleus (SOL), medial gastrocnemius (MG), semitendinosus (ST), vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF) and gluteus maximus (GM). The timing of the running cycle was recorded with footswitches and the ankle, knee and hip joint angles were measured with goniometers. Additionally, EMGs were recorded during a series of standardized explosive movements, in order to obtain maximal dynamic EMG output from the muscles. The signals were sampled at 1 kHz and analyzed off-line. The relative muscle activation during running was calculated by normalization of the EMG amplitude to the maximal dynamic EMG amplitudes. Integrated EMG (IEMG) over the full stride cycle was calculated. Onset and offset of the EMG activity periods in relation to the heel strike were determined by a graphic interactive method (Walter 1984). Origin-to-Insertion (OI) length-changes were calculated from the joint angle-to-OI transfer functions (Visser et al. 1990). Twenty stride cycles from each subject were analyzed.

## RESULTS

REs were 187.3±4.5 and 177.9±3.7 ml $O_2$ kg$^{-1}$ min$^{-1}$ km$^{-1}$ in Bgroup and Ggroup respectively after the selection ($p < 0.0001$). Average values for both groups were: age 28±4 years, height 1.80±0.05m, weight 69.3±4.2 kg, PRvel velocity 19.2±0.8 km hr$^{-1}$, maximal oxygen uptake 4.72±0.33 l min$^{-1}$ and no significant differences between the groups were found. As indicated by fig.1 no differences in the timing of the EMGs and in the OI-velocity profiles were found. However, a two way ANOVA revealed that Ggroup had a lower general IEMG in the muscles than Bgroup ($p < 0.03$) which also is reflected by fig.1, but no differneces between the individual muscles were located.

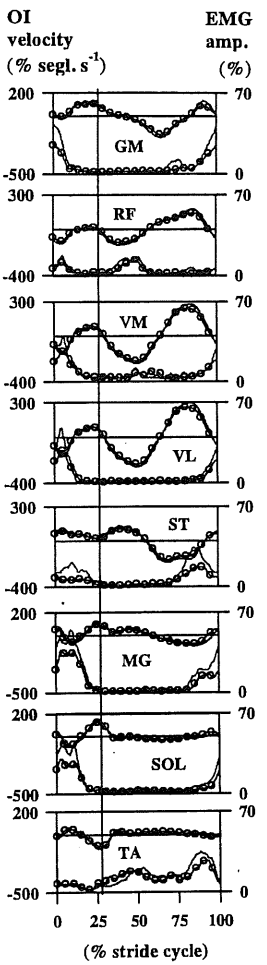



*Fig.1 Average EMGs recorded at PRvel (bottom curves) and OI-velocties (symbols = Ggroup). Vertical line = toe off.*

## DISCUSSION

The two groups of elite runners observed in this study were homogeneous in all the variables obtained, except for the difference in IEMG. A possible relationship between a low level of muscle activation and an optimal running economy is not straight forward to explain. If we assume that the runners in the two groups, due to the homogeneity, on average are performing the same amount of mechanical work and that the lower extremity muscles are generating the mechanical work over the same contraction distance and with the same average velocity and muscle force, then the lower IEMG level in Ggroup indicates that this group of runners might have had stronger muscles than Bgroup. Then the significant difference in running economy should be explained by differences in mechanical efficiencies of the muscle work performed at steady state at different relative contraction levels due to e.g. different loads on the aerobic and anaerobic energy producing systems. However, there were no significant differences between the groups in the lactate levels and the R-values. A higher degree of synchronous muscle activation (Milner-Brown 1975) in Bgroup could also explain the difference in IEMG between the groups as well. If this is the case then there should be no difference in the general muscle force between the groups either. The observations in the present study raises several questions, but firstly it should be investigated if there is a relationship between RE and the strength of the lower extremity muscles.

## REFERECES

Cavanagh and Kram (1985) Med Sci Sports Exerc 17:304-308
Morgan and Craib (1992) Med Sci Sports Exerc 24:456-461
Milner-Brown et al.(1975) Electroencephalogr Clin Neurophysiol 38(3):245-254
Visser, J.J. et al (1990) Eur J Appl Physiol 61:453 -60
Walter,C (1984) Hum. Mov. Sci. 3:155-162

## ACKNOWLEDGEMENTS

This study was made possible by Tha Danish Elite Sports Association and The Danish National Research Foundation

# THREE DIMENSIONAL ANALYSIS OF THE HIP JOINT TORQUES AND POWERS FOR JAPANESE FEMALE HURDLERS

**Norihisa FUJII, Michiyoshi AE, and Ken MIYASHITA**
Institute of Health and Sport Sciences, University of Tsukuba, Tsukuba, JAPAN

## INTRODUCTION

The patterns of joint torques have been investigated in various human movements, i.e. walking, jogging, and sprinting. However, there is very little information on the joint torques and joint torque powers of the lower extremities during sprint hurdling. The purpose of this study was to compute and analyze joint torques and joint torque powers of the hip for Japanese female hurdlers.

## METHODS

Seven female hurdlers and multi-event athletes performing sprint hurdling were videotaped with two high-speed cameras (NAC Inc.) which were panned. Ground reaction forces during the take-off phase were also measured with a Kistler force platform. Three dimensional kinematic and kinetic analyses were carried out to estimate angular velocity and joint torque (JT) of the hip for the trail leg. The angular velocity and the joint torque were converted to the segment coordinate system fixed to the thigh, as shown in Figure 1. Joint torque power (JTP) was calculated as the product of each component of the angular velocity and JT in the segment coordinate system.

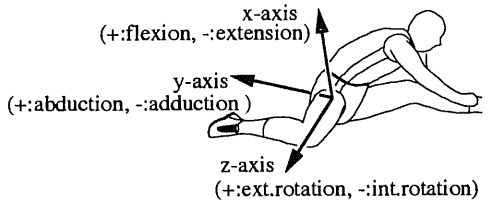


Figure 1 Segment coordinate system fixed to the thigh

## RESULTS AND DISCUSSION

Figure 2 (a) shows an example of JTs of the hip for the trail leg, which were taken between the third and the fourth hurdles. Figure 2 (b) and (c) show the JT and the JTP patterns from the instant of the take-off (0.72 s) to the toe-off of the lead leg (1.15 s). The time of touchdown of the lead leg was 1.06 s, as indicated with a vertical broken line.

The flexion torque (positive torque in x-axis) at the hip was generated at the instant of the take-off, and then changed to the extension torque (negative torque in x-axis) before the touchdown of the lead leg. The adduction torque (negative torque in y-axis) and the external rotation torque (positive torque in z-axis) were generated during the hurdling phase (0.72 to 1.06 s).

In the first half of the hurdling phase, only flexion torque (x) produced the positive power used for lateral swing of the trail leg, while the powers of the adduction torque (y) and the external rotation torque (z) were negative. In the latter half, the external rotation torque (z) produced the positive power which was used for swinging back of the trail leg.

Although the patterns of the JT and JTP in Figure 2 show some similarities with those of the male hurdlers (Fujii et al., 1996), the peak adduction torque (y) and the external rotation torque (z) for the females appeared earlier than those of the male hurdlers. It is inferred that the differences in the timing of the torque exertion are resulted from the difference in hurdle height (0.840 m for female and 1.067 m for male), and also from the difference in the hurdling time (0.34 s for female and 0.42 s for male on the average).

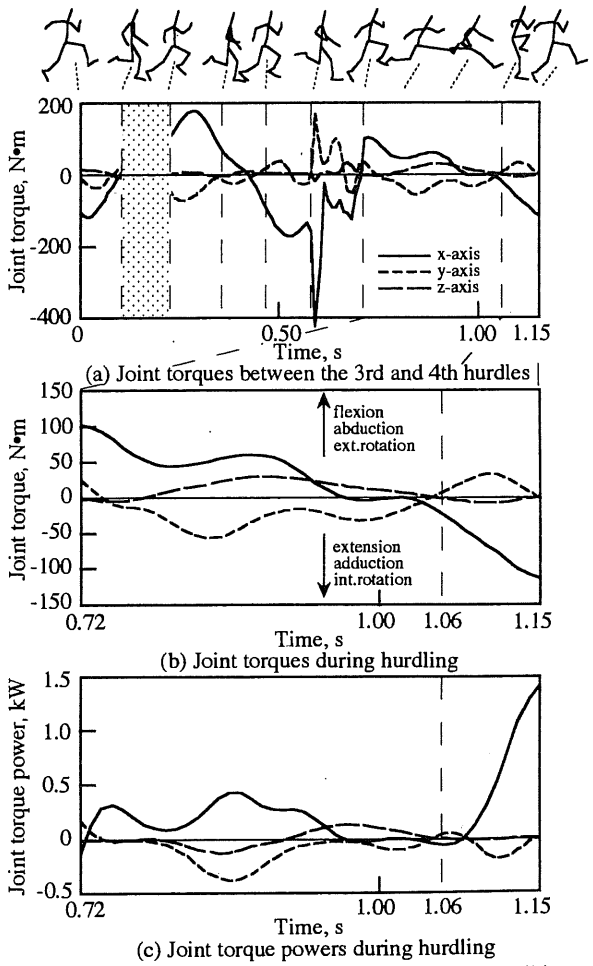


Figure 2 Joint torques and powers of the hip for the trail leg

# ANALYSIS OF THE RACE-PATTERNS OF MEN'S 400M HURDLES
## —The races of the XXVI th OLYMPIAD in Atlanta, 1996—

Y. Morioka[1], M. Ae[2], M. Sugita[3], A. Matsuo[4] and K. Kobayashi[3]
1)Sports Science Lab., Japan Amateur Sports Association, Japan
2)Institute of Health and Sport Sciences, University of Tsukuba, Japan
3)Department of Life Science, University of Tokyo, Japan
4)National Institute of Fitness and Sports in Kanoya, Japan

## INTRODUCTION

Biomechanics Project Team of the Japan Amateur Athletic Federation(JAAF) had opportunity to videotape the races of the men's 400m hurdles of the XXVIth OLYMPIAD in Atlanta, 1996. The final race was high level in which the four finishers broke 48.00s.

The purpose of this study was to analysis of the race patterns of the men's 400m hurdles and to obtain the characteristic of race pattern of the world class hurdlers.

## METHODS

The race of the men's 400m hurdles was videotaped with several video cameras panning from start to goal at 60fields/s. After time indication was superimposed on each field of the VTR images, the time indication at touchdown immediately after hurdling was read. Using the flash of starter's gun recorded on VTR image as start, each touchdown time from the start were obtained. Initial times for each hurdle to hurdle interval were calculate as difference between the successive two touchdown time and divided by interval distance to obtain the section velocity.

## RESULTS AND DISCUSSION

Figure 1 shows section velocity curves and number of steps during the section (NS) of R. Adkins (bold line: the first place in the final, 47.54s) and R. Robinson (narrow line: the 5th place in the final, 48.30s). Both Adkins and Robinson attained maximum velocity at 1-2section(between 1st hurdle and 2nd hurdle) and gradually decelerated from this section to last 10-Gsection(between 10th hurdle and Goal). Decline of section velocity for Robinson larger than Adkins after 5th hurdle.

Figure 2 shows change in velocity between the successive two sections( $\Delta V$) for Adkins and Robinson. $\Delta V$ for Robinson remarkable declines at 5-6 and 7-8section in comparison with Adkins. Decrease in section velocity with increase in NS was -0.23m/s for Adkins (6-7 section) and -0.29m/s for Robinson (7-8 section).

These results show that Adkins was more superior than Robinson in ability to maintain section (running) velocity during the latter half of the race and to restrain the deceleration with increase in NS.

This study was supported by JAAF Biomechanics Project team.

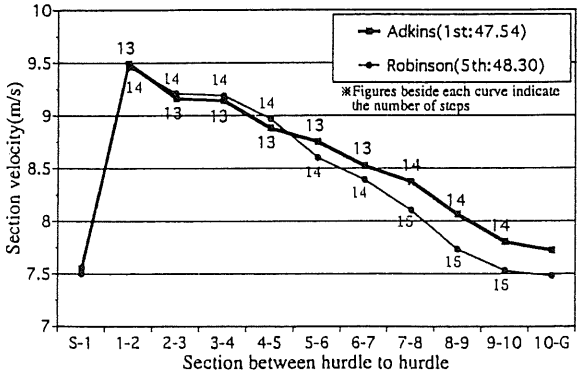



**Figure1 Section velocity curves for Adkins and Robinson**

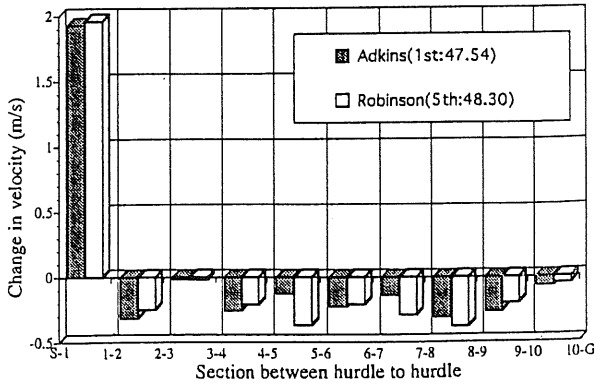



**Figure2 Change in running velocity between the successive sections for Adkins and Robinson**

## THE ANALYSIS OF RACE IN 400M HURDLE FOR WOMEN

Toshifumi Yasui[1)] ,Kazuyuki Ogiso[2)],Kiyohide Aoyama[3)] Yasuo Sekioka[1)] Jun Nagai[1)]
Ken Miyashita[1)]Mitsugi Ogata[1)]
[1)]Tsukuba University, Tsukuba city, Ibaraki, JAPAN, [2)]Toba National College of Maritime, Toba city,Mie,JAPAN,[3)]Nihon University, Tokyo, JAPAN,

### INTRODUCTION

The purpose of this study was to discuss factors related to performance analyzing the women's 400m hurdle race.

### METHODS

We videotaped the races with 60f/sec speed in 1994-1996. The video images were dubbed with a digital timer and each interval time was calculated and analyzed. The total number of races and the total number of samples were 65 and 383 respectively. The interval time was defined as the time from starting point to the first touchdown after hurdling , and between each touchdown thereafter. The samples were categorized into 5groups based on performance time. The range of performance time was 58.79-70.81 sec.

### RESULTS

High correlation between each interval and performance time was found in all samples. The correlation coefficient between 9th interval time and performance time was the highest among all results. Performance time was correlated with 2,5,6,9,10th interval time in group A(mean performance time=58.79-59.98sec,n=49). Performance time was correlated with all interval times except for 11th interval time in group B(mean performance time=60.02-61.99sec,n=103). Performance time was correlated with 2,35-11th interval time in group C(mean performance time=62.06-63.98sec,n=109). Performance time was correlated with 2-8,10,11th interval time in group D(mean performance time=64.03-65.97sec,n=73). Performance time was correlated with all interval times except for 11th interval time in group E(mean performance time=66.07-70.81sec,n=49).

**Tab1 The Equations of Model Interval Time in all samples**

| INTERVAL | EQUATION | CORRELATION COEFFICIENT | S.E |
|---|---|---|---|
| Int.1 | y=7.182x+11.478 | .684 *** | .392 |
| Int.2 | y=9.812x+15.654 | .730 *** | .470 |
| Int.3 | y=9.498x+15.852 | .733 *** | .451 |
| Int.4 | y=9.964x+12.539 | .764 *** | .431 |
| Int.5 | y=9.448x+13.739 | .819 *** | .340 |
| Int.6 | y=8.665x+16.353 | .877 *** | .244 |
| Int.7 | y=7.958x+18.730 | .898 *** | .200 |
| Int.8 | y=7.097x+22.133 | .898 *** | .178 |
| Int.9 | y=6.472x+24.264 | .909 *** | .152 |
| Int.10 | y=5.544x+29.014 | .864 *** | .166 |
| Int.11 | y=3.759x+36.935 | .695 *** | .199 |

***: $p<0.001$

Averaged velocity of second interval in each group was highest. Group A was the highest in particular. After the second interval averaged velocity was gradually decreased to 10th interval. The pattern of velocity was similar in all groups. Averaged stride frequency of second interval in each group was the highest, and theirs of group B-E gradually descended to $10^{th}$ interval. But stride frequency of group A descended to $6^{th}$ interval. After that, it ascended to $7^{th}$ interval and continuously descended to $10^{th}$ interval. Average stride length of the second and third interval in each group was higher than other stride length. The stride length in each group gradually descended to $10^{th}$ interval. But the difference among each group was higher in the latter half of race than the first half. The change of stride length in group A was minimum.

### DISCUSSION

Since average velocity in each group reached to the maximum in second interval and there were differences in all intervals among all groups, the higher velocity in second interval may be very important to improve performance time. Performance time was significantly correlated with 2,5,6,10th interval time in all groups. Furthermore, performance time was correlated with 7-9th interval time in 4 groups. Therefore, it may be suggested that performance time correlates highly with the latter half of the race. Still, stride length was more highly correlated with performance time than stride frequency, especially the stride length in the latter half of race. In conclusion, higher velocity in second interval and improving performance time in the latter half of the race can be the important factors to improve overall performance time.

### REFERENCES

1.SUSANKA,P.,MISKOS,G., et al: TIME ANALYSIS OF THE 400m HURDLES.,SCIENTIFIC REPORT ON THE II WORLD CHAMPIONSHIPS IN ATHLETICS ROME 1987:I.A.A.F,1987

# ELECTROMYOGRAPHIC ACTIVITY AND BIOMECHANICS OF THE TRANSFER OF HORIZONTAL SPEED INTO ANGULAR VELOCITY DURING THE TAKEOFF IN THE LONG JUMP

W. Kakihana and S. Suzuki
Department of Sports Sciences, School of Human Sciences, Waseda University, Tokorozawa, Japan

## INTRODUCTION

A major task of a jumper at takeoff is to obtain vertical lift, while maintaining horizontal velocity. The purpose of this study was to determine the characteristics of the electromyographic activity (EMG), kinematics, and kinetics involved in the transfer of horizontal speed into angular velocity during the takeoff by comparing the performance of two jumpers.

## METHOD

Two male long jumpers, TM (the more accomplished performer) :175 cm, 67 kg, 7.63 m (greatest distance), and YS :177 cm, 67 kg, 6.80 m (greatest distance), performed long jumps at different approach speeds by increasing the step numbers from 1 to 9. Surface EMG activity was recorded from selected muscles of the takeoff leg. Kinematics were determined with a high speed video (250 f/s). The ground reaction forces (GRFs) included the vertical (Fz), forward-backward (Fy), and side-to-side (Fx) components.

## RESULTS

Kinematics: The angle of takeoff of TM, was greater (0.41 - 0.59 rad) than that of YS (0.25 - 0.29 rad). In general, the horizontal velocities of the center of gravity (CG) at the instant of touchdown [TM: 7.37 m/s; YS: 7.50 m/s for a 9-step approach (SA)], and those at takeoff (TM: 6.85 m/s; YS: 7.61 m/s for a 9-SA) were both increased as the approach step number was increased. However, for TM the instant of touchdown and the vertical velocity of the CG were less than that for YS (-0.10 m/s vs -0.25 m/s for a 9-SA), the upper extremity was more slanted backward (1.68 vs 1.45 rad for a 9-SA), the height of the CG was lower (0.91 vs 1.00 m for a 9-SA), and the joints of the knee and ankle were more extended (knee: 2.83 vs 2.58 rad, ankle: 2.11 vs 1.94 rad for a 9-SA). The maximum velocity of the knee and ankle joint at the contact phase was greater (knee: -10.73 vs -5.78 rad/s; ankle: -8.92 vs -7.19 rad/s for a 9-SA). The velocity of backward rotation of the thigh during the early contact phase (E2 phase) was less. At the takeoff, the vertical velocities of the CG were greater (3.01 m/s vs 2.09 m/s for a 9-SA).

Kinetics: For TS, on the Fz component of the GRF, the peak value of the impact force (6.89 vs 4.14 kN), the impulse at the contact phase (319 vs 215 N·s), and the impulse of the Fy component of the GRF (-80 vs -30 N·s) were larger for the 9-SA.

EMG activity: For TS, the pattern of EMG for all muscles (biceps femoris, rectus femoris, vastus medialis, gastrocnemius, soleus and tibialis anterior) revealed an earlier onset of activity relative to the cyclical phases of the joint angles as the approach speed increased. Considerable overlapping of the EMG bursts was seen between the gastrocnemius and the tibialis anterior, and co-contraction could be observed just after the contact phase (E2 phase) at an approach speed beyond a 5-SA. The biceps femoris EMG for YS increased during the transition between the E1 and E2 phases, when it contributed to hip extension as the approach speed increased. However, the biceps femoris EMG for TM was moderate during the E1-E2 transition, and increased during the early E3 phase as the approach speed increased.

## DISCUSSION

The characteristics of EMG, kinematics, and kinetics transfer of horizontal speed into angular velocity during the takeoff was primarily seen in the EMG activity of the biceps femoris and the Fz component of the GRF, which was reflected in the kinematics recorded during the takeoff.

## REFERENCE

Kakihana, W., Yamanouchi, T. and Suzuki, S. (1995) Electromyographic activity and biomechanics of long jumps with different takeoff angles. Med. Sci. Sports Exercise 27 (Suppl. 5), S171.

# A THREE-DIMENSIONAL ANALYSIS OF ANGULAR MOMENTUM IN THE ASIAN TOP DISCUS THROWERS

**T.Miyanishi** 1, **S.Sakurai** 2 and **A.Wakayama** 3
1 Sendai College, Miyagi, Japan. 2 Nagoya University, Nagoya, Japan. 3 Institute of Sports Medicine and Science, Aichi, Japan.

## INTRODUCTION

Dapena (1993a,b) has pointed out that the increase of the horizontal and the vertical speeds of the discus were closely linked with the angular momentum of the discus thrower about the vertical and anterior-posterior axes (the direction of the throw), respectively. However, he did not examine the influence of the motion of the discus thrower on the components of the angular momentum of the thrower. Our purpose was to clarify the angular momentum of Asian top discus athletes during the second single support phase (SS2) to the delivery phase (DV) by using a three-dimensional (3D) link-segment analysis, and to examine the relationship between the angular momentum and the motion in the thrower.

## METHODS

Top six finalists and two Japanese throwers (rank 10 and 11) of the 12th Asian games in Hiroshima 1994 were filmed with two high-speed video cameras (NAC) at 200 frames/s. The average distance of the top six was 56.44±1.84 m. 3D landmark coordinates of 15 body segments were obtained by using the DLT method. The locations of the center of gravity (cg) of the segments, the angular momentum (H) about its cg of the thrower (include the angular momentum of the discus prior to release), the velocity of the discus and the absolute angle for the upper trunk were computed based on previous study (Miyanishi et al., 1997). All angular momentums (local term) about the three orthogonal axes passing through its cg of each segment were computed, except those about the longitudinal axis of segment which are the head, hands, forearms, upper arms and feet, were assumed to be zero. The right-handed orthogonal inertial reference system with the signs for the angular momentum were shown in **Figure** at the right side.

## RESULTS AND DISCUSSION

Two different patterns were found in the change of the angular momentum of the thrower during the delivery phase (see **Figure**). One was the pattern of the gold-medalist which rapid increase of the positive $H_Z$ with the negative $H_X$ and the negative $H_Y$ decreasing. The other was that of the bronze-medalist which marked increase of the negative $H_X$ and the negative $H_Y$ with the positive $H_Z$ dropping. The horizontal speed of discus at the release was larger in the bronze-medalist (19.55 m/s) than in the gold-medalist (18.30 m/s). The vertical speed of discus at release was larger in the gold-medalist (13.05 m/s) than in the bronze-medalist (12.50 m/s). The angle of the backward lean (posterior direction) for the upper trunk at the instant of the ground contact by the right foot (Ron) was larger in the gold-medalist (127°) than in the bronze-medalist (101°). Time used from Ron to the release of discus was short in the gold-medalist (0.30 s) than in the bronze-medalist (0.44 s). These results indicated that the manner to obtain the velocity of the discus in the delivery phase is dependent on the difference of the motions of the thrower. That is, as shown in the stick-pictures in **Figure**, the motion of the gold-medalist with the body leaning consistently backward and right ward (throwing arm side), which will be called the motion of "speed dominant type". On the contrary, the bronze-medalist was without consistent backward, and body rotating forward and left ward (non-throwing arm side), "force (strength) dominant type". Therefore, we may say that the motion of "speed dominant type" obtains not only the horizontal speed of the discus by mainly using rotation about the longitudinal axis of the body, but also the vertical speed of the discus at the same time. On the other hand, "force dominant type" increase the horizontal speed of discus by using both rotations about the frontal and the longitudinal axes of the body, and the vertical speed by rotation of the sagittal axis of body. [This project was supported by Japan Amateur Athletic Federation (JAAF).]

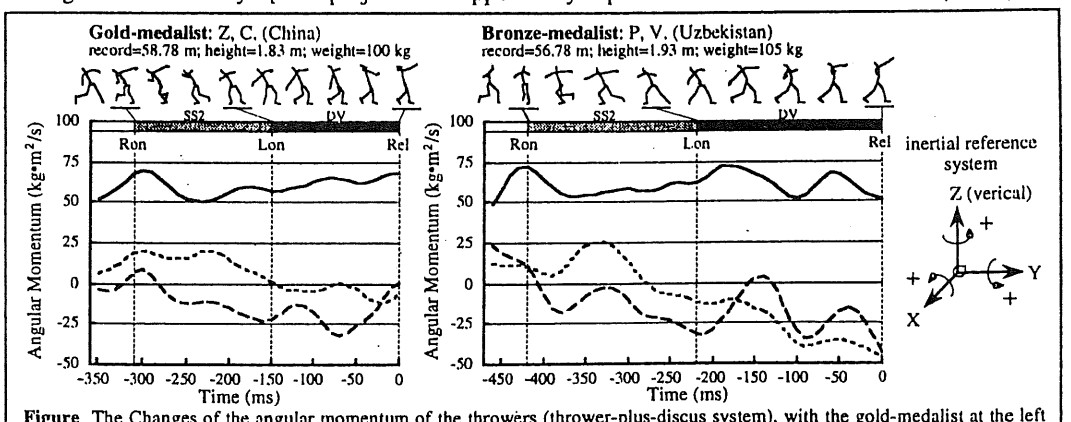



**Figure** The Changes of the angular momentum of the throwers (thrower-plus-discus system), with the gold-medalist at the left side and the bronze-medalist at the right side, respectively. Solid line represent the angular momentum about the Z-axis, dotted X-axis and dashed Y-axis (the direction of the throw) in the inertial reference system. Three vertical dotted lines represent the instants of the ground contact by the right foot (Ron), by the left foot (Lon), and of the release of the discus (Rel), respectively.

## REFERENCES

Dapena, J. (1993a) New insights on discus throwing. *Track Technique* 125, 3977-3983.
Dapena, J. (1993b) An analysis of angular momentum in the discus throw. In *XIVth Congress of the Int. Socie. Biomech. (Abstracts I)*, pp.306-307, Paris, France: International Society of Biomechanics.
Miyanishi, T. et al. (1997) A three-dimensional analysis on mechanical energy flows of torso and arm segments in baseball throw. *Jpn. J. Phys. Fitness Sports Med.* (in press)

# THREE-DIMENSIONAL ANALYSIS OF POLE MOMENT AND JOINT TORQUES IN ELITE POLE VAULTERS

J. Takamatsu[1], M. Ae[2] and N. Fujii[2]

1 Doctoral program in Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan

2 Institute of Health and Sport Sciences, University of Tsukuba, Tsukuba, Japan

## INTRODUCTION

Since a fiberglass vaulting pole was developed and used widely, the world record of pole vault has been changed rapidly. Therefore, not only vaulters and coaches but also researchers have been interested in advantages of the fiberglass pole and forces exerted on the pole by the handgrips of a vaulter. If the forces applied to the pole become clear, particularly in elite vaulters, it will be useful for improving and understanding the mechanisms of pole vault and vaulting techniques.

The purpose of this study was to investigate three-dimensional joint torques of the vaulter and the moment exerted on the pole by the vaulter, henceforth pole moment.

## METHODS

Three-dimensional segment endpoint data were obtained by digitizing 16mm films and video tapes of pole vaulters participated in 1991 World Championships in Tokyo and 1994 Asian Games in Hiroshima. 16mm films and video tapes were recorded with two 16mm cine cameras (100 frames/s) for 1991 W.C., and two video cameras (60 fields/s) for 1994 Asian G.. Thirteen vaults of elite pole vaulters were selected for analysis, and kinematic data were computed.

Three-dimensional joint torques and pole moment were computed using a link segment model consisted by foot-shank, thigh, head-trunk, right upper arm and forearm. Left arm was ignored and the effect of forces exerted by the left handgrip influenced on the pole moment was replaced by a couple acting at the right handgrip, based on the method of McGinnis and Bergman (1986).

## RESULTS AND DISCUSSION

Fig.1 shows changes in pole moment and joint torques for the vaults that were normalized by the time from the takeoff to the pole-straight, and then averaged. MPB means the maximal pole-bend. Positive value indicates the bending moment (flexion torque) and negative value for the straightening moment (extension torque). The pole-bending moment was exerted from the takeoff to 30% time, and the straightening moment was applied from the instant of MPB before the pole-straight.

McGinnis and Bergman suggested that the pole-straightening moment causes the vaulter to rotate into an inverted (upside-down) position, and the vaulter may have difficulty swing-up motion if the bending moment was too long in duration. In this study, the straightening moment in swing-up phase (pole-bending phase) was very small while the shoulder and hip torques were relatively larger than that appeared in McGinnis and Bergman (1986). It seems that the elite vaulters in this study were able to swing up the body using shoulder and hip torques rather than exerting the pole-straightening moment in pole-bending phase. If bending moment is maintained throughout this phase, larger elastic energy may be stored on the pole and returned to the vaulter as larger gravitational potential energy (C.G. height) in pole-straightening phase. Therefore, vigorous shoulder extensor and hip flexor would be important to swing up with this pole-bending technique.

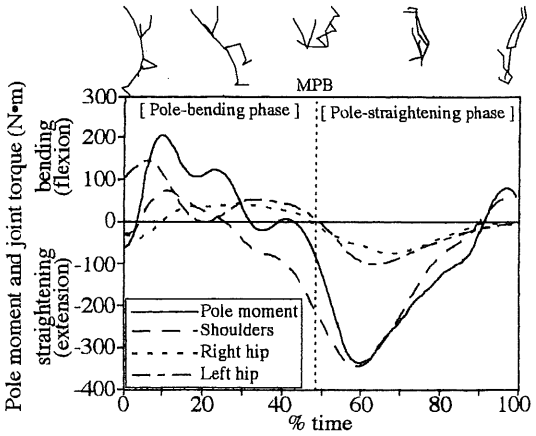



Fig.1 Pole moment and joint torques normalized by the time and averaged

## REFERENCES

McGinnis, P. M. and Bergman, L. A. (1986) An inverse dynamic analysis of the pole vault. *Int. J. Sport Biomechanics* **2(3),** 186-201.

# A THREE DIMENSIONAL ANALYSIS OF WHEELCHAIR BASKETBALL SHOOTING IN QUADRIPLEGIC PLAYERS

Hiroyuki NUNOME, Wataru DOYO, Sinji SAKURAI,
Yasuo IKEGAMI and Kyonosuke YABE

Research Center of Health, Physical Fitness and Sports,
Nagoya University, Nagoya JAPAN.

## INTRODUCTION

Wheelchair basketball has become a popular sports in disabled peoples. However, it is likely that quadriplegic peoples have a difficulty to join the game by their lower throwing ability. This study was therefore designed to characterise the shooting motion in quadriplegic players and to explore several factors limiting their throwing ability from a biomechanical point of view.

## METHOD

Six disabled (quadriplegia ) and six abled players were participated in this study. All the subjects sat on the wheelchair and shot the ball to the official height goal (3.05m). Their single hand shooting motion were sampled by two high-speed video cameras (Nac HSV400) at 200fps. The direct linear transformation (DLT) method was used to obtain 3-D space coordinates of the upper limb and the ball center. Following seven joint angle changes were obtained; 1) abduction/adduction at the shoulder joint, 2) horizontal abduction /horizontal adduction at the shoulder joint, 3) internal/external rotation at the shoulder joint, 4) flexion/extension at the elbow joint, 5) dorsi/palmar flexion at the wrist joint and 6) ulnar/radial flexion at the wrist joint. Initial velocity and projection angle of the ball were calculated.

## RESULTS AND DISCUSSION

The ball could not be reached to the goal in all the disabled subjects. Both the initial velocity (5.37 vs. 6.10 m/s) and the projection angle (52.5 vs. 63.9 deg) in disabled group were significantly smaller ( $p < .01$) than those of able group. As for the shooting motion, smaller ranges of angular motions at the wrist joint (dorsi flexion and ulnar flexion) were observed in disabled group (shown in Figure 1). In contrast, larger ranges of angular motions at the shoulder joint (abduction and horizontal abduction) and at the elbow joint (extension) were also observed in disabled group. These results suggested that the wrist joint motion may play an important role in basketball shooting. It should be considerd that the lower throwing ability observed in quadriplegic players may depend on dysfunction of the wrist joint.

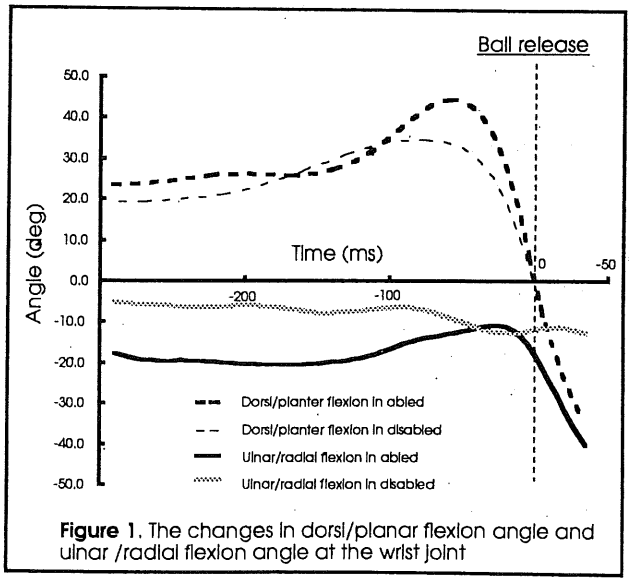



**Figure 1.** The changes in dorsi/planar flexion angle and ulnar /radial flexion angle at the wrist joint

# CHANGES IN JOINT TORQUES AND MOMENTA OF THE BODY SEGMENTS BY THE PRACTICE IN A MEDICINE-BALL BACKWARD THROWING

Yasuyuki Kubo[1)], Michiyoshi Ae[2)], and Norihisa Fujii[2)]
[1)]Doctoral program in Health and Sport Sciences, University of Tsukuba.
[2)]Institute of Health and Sport Sciences, University of Tsukuba.

## INTRODUCTION

The purpose of this study was to investigate the changes in momenta of the body segments and joint torques by the practice in a medicine-ball backward throwing as an example of whole body movement which aimed maximum velocity of the distal segment.

## METHODS

Seventeen males were asked to throw a medicine-ball (mass 3kg) backward in overhead manner with a maximal effort. Twelve unskilled throwers were selected on the basis of the throwing distance, and participated in the practice session of two days. Throwing motion was videotaped with a high-speed camera and ground reaction forces were measured with a force platform. A 8-link segment model was used to compute kinematic and kinetic variables. To examine the changes in joint torques due to the practice, two indexes were calculated. DAI (difference of angular impulse) indicates the changes in the magnitude of the joint torques. Pattern correlation coefficient (r) indicates the similarity of the joint torque patterns between pre and post practices. These indexes were calculated by the following equations, respectively.

$$DAI = \int |T_{post}|dt - \int |T_{pre}|dt \quad , \quad r = \frac{\sum (\tau_{pre} - \bar{\tau}_{pre})(\tau_{post} - \bar{\tau}_{post})}{\sqrt{\sum (\tau_{pre} - \bar{\tau}_{pre})^2}\sqrt{\sum (\tau_{post} - \bar{\tau}_{post})^2}} \quad ,$$

where, $T_{pre}$ and $T_{post}$ represent joint torques of pre and post practices. $\tau_{pre}$ and $\tau_{post}$ represent time-normalized torques. $\bar{\tau}_{pre}$ and $\bar{\tau}_{post}$ represent mean values of $\tau_{pre}$ and $\tau_{post}$.

## RESULTS AND DISCUSSION

Figure 1 represents the relationships between pattern correlation coefficient (r) and DAI for the shoulder, hip, knee and ankle joints. There were high correlations between pattern correlation coefficient and DAI at the hip ($r=0.863$, $p<0.001$) and knee torques ($r=0.737$, $p<0.01$). This result indicates that the subjects who showed greater changes in the patterns of the hip and knee joint torques decreased the magnitudes of these torques, and vice versa.

Subjects who changed the patterns and decreased the magnitudes of the hip and knee joint torques, increased the horizontal momentum of the legs so as to obtain a greater horizontal momentum of the trunk. However, others who changed no remarkable patterns and increased the magnitudes of these torques, increased the relative horizontal momentum of the trunk to the legs. It was found from the detailed analysis of torque patterns and throwing motion that the forward lean of the lower leg in the first half of throwing motion was one of the important factors to change torque patterns and the momentum of the legs.

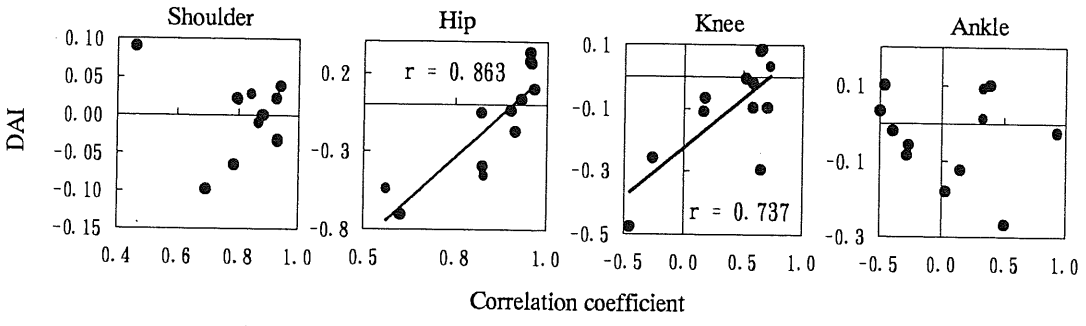



Figure 1. Relationships between pattern correlation coefficient and DAI.

# ENERGY CONTROL AT CONTACT PHASE OF BIOMECHANICAL SYSTEM WITH BALL

Galina P. Ivanova
Lesgaft Academy of Physical Culture, St.Petersburg, Russia

## INTRODUCTION

Contact phase of biomechanical chain with ball was studied mainly for tennis strokes. Presumably, these studies by Brody (1987), Grabiner et.al. (1983), Hatze (1976) were aimed at the development of stroke theory necessary for improvement of tennis rackets [1-3]. Besides, there are some diversified papers by Watanabe et.al. (1979), and Elliott (1982), devoted to the influence of grip firmness of rackets [3-5]. They are particularly interesting from the viewpoint of the mechanism of energy transformation and organisation of striker's mass. The abovementioned issues serve as subjects of the current investigations.

## METHODS

The projection of the force work on the motion axis at the impact phase was experimentally studied. We used the method of graphical integration of the square under the curve of ball deformation as a function of motion path. Deformation force was determined from ball squeezing based on rigidity coefficient equal to 110 $Ncm^{-1}$. This parameter was verified by experimental measurements. Joint ball motion and squeezing were calculated from system coordinates, which were read-off from impact filming taped on 35mm width film at electronically calibrated frequencies from 800 to 2,000 fps. Error of coordinate reading did not exceed 1% of the maximum value; velocity error was within 1.5%. Calculations of ball and biosystem energies after the impact were performed on the basis of the developed mathematical model:

$E_{u,1} = (E_{v,1} + E_{u,2})*K + A - E_{u,2}$;

$m_2 = (m_1U_1^2 - 2A + Km_1V_1^2)/(KV_2^2-U_2^2)$;

where symbols "1" and "2" denote ball and biosystem parameters respectively; $E_u$ and $E_v$ are energies before and after the impact; K is restitution coefficient; V and U are velocities before and after the impact; m is mass and A is total work force.

## RESULTS

Analysis of 10 tennis services provided the following parameters of contact phase:

$V_2 = 30 \pm 2.1$ $ms^{-1}$; $m_2 = 0.15 \pm 0.1$ kg; $X = 0.12 \pm 0.02$ m; $A = 22.0 \pm 1.2$ Nm;

where X is joint motion path. The following Table shows similar characteristics for strokes in volleyball:

| Parameter | Slight ball deformation | Considerable ball deformation |
|---|---|---|
| $m_2$, kg | 0.7 | 0.15 |
| $V_2$, $ms^{-1}$ | 26 | 22 |
| $U_1$ , $ms^{-1}$ | 35 | 26 |
| X, m | 0.27 | 0.22 |
| $A= A_L + A_{UL}$ | 13.65= 3.65+ 10 | 36 = 10 + 26 |

where $A_L$ and $A_{UL}$ are force works during loading and unloading phased correspondingly.

## DISCUSSION

1. Possibility of energy control during the contact phase depending on aims of strokes and properties of bodies in contact is experimentally proved.
2. Force work on ball deformation as a function of it's rigidity is determined.
3. Mathematical model for calculation of striker's mass in account of work of deformation forces is proposed.
4. Due to extrapolation program based on sportsman experience, the force, applied to ball on the path of their joint motion is being developed by biosystem 90-120 ms before the stroke.

## REFERENCES

1. H. Brody. *Models of Tennis Racket Impacts.* International Journal of Sports Biomechanics. V.3, N.3, p.293-6, (1987).

2. M. Grabiner, J. Groppel, K. Campbell. *Resultant Tennis Ball Velocity as a Function of Off-Center Impact and Grip Firmness.* Medicine and Science in Sports and Exercise. v.15, n.6, p.542-4, (1983).

3. H. Hatze. *Forces and Duration of Impacts and Grip Tightness During the Tennis Stroke.* Medicine and Science in Sports and Exercise. v.8, n.2, p.88-95, (1976).

4. T. Watanabe, Y. Ikegami, M. Miyashita. *Tennis: the Effect of Grip Firmness on Ball Velocity After Impact.* Medicine and Science in Sports and Exercise. v.11, n.4, p.359-61, (1979).

5. B. Elliott. *The Influence of Grip Tightness on Reaction Impulse and Rebound Velocity.* Medicine and Science in Sports and Exercise. v.4, n.5, p.348-52, (1982).

# RELATIONSHIP BETWEEN LEFT FOREARM MUSCLE ACTIVITIES AND ADDITIONAL TORQUE ABOUT GRIP OF JAPANESE BOW IN RELEASE

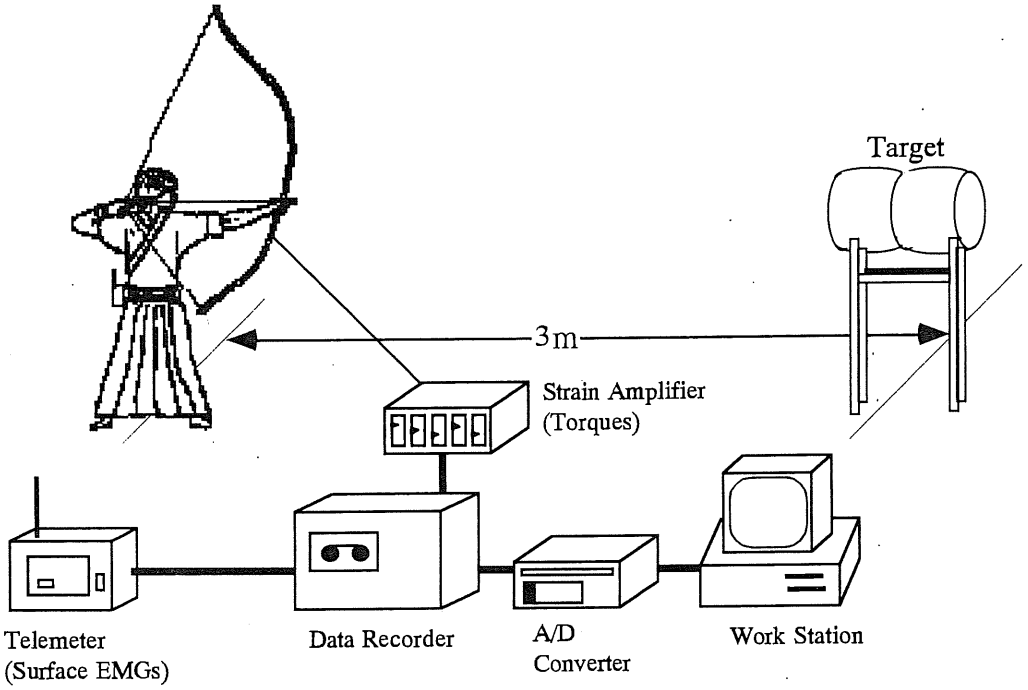
Satoshi HOSOYA, Morihiko OKADA, Chikara MIYAJI, Keigo OHYAMA BYUN
University of Tsukuba, Tsukuba, Ibaraki, JAPAN

## INTRODUCTION

KYUDO using Japanese bow is a traditional sports. Since Japanese bow is very complicated in the shape and structure, additional torque about grip must be applied in horizontal plane (NEJIRI) and in sagittal plane (UWAOSHI) to hit the target[1][2]. The purpose of this study was to investigate the relationship between left forearm muscle activities and these torque in releasing a Japanese bow.

## METHODS

Surface EMGs of four muscles (extensor carpi radialis longus, extensor digitorum, extensor carpi ulnaris, flexor carpi ulnaris) and torque about grip in release were collected in 10 male subjects shooting a Japanese bow. Cross talks between EMGs obtained from adjacent muscles were evaluated by using cross-correlation function. Absolute values less than 0.3 of cross-correlation coefficient was considered as a criterion for specific EMG signals.




## RESULTS

Analysis of correlations between the relative EMG magnitude and impulses of torque about grip just before release revealed that extensor digitorum and extensor carpi ulnaris muscles had positive relationships to the torque in the horizontal plane. On the other hand, activities in the extensor carpi radialis longus, extensor carpi ulnaris and flexor carpi ulnaris muscles had negative relationships to the torque in the sagittal plane.

## REFERENCES

[1]Hosoya. S., Shinzaki. S., Mori. T., Takahashi. M., A mechanical study of NEJIRI-technique in KYUDO -Computer simulation analysis-, JAB Vol.26 No.3, pp31-40,1994.

[2]Hosoya. S., Miyaji. C., Kobayashi. K., Computer simulation of restitution of Japanese bows -Two dimensional analysis with a link segments model-,JJSSI Vol.5 No.2, pp25-33,1995.

# GROUND REACTION FORCES OF JAPANESE DRUM PLAYING

N. Yasuda and H. Yamamoto
Kanazawa University, Kanazawa, Japan

A. Azuma
Fukui Health Administration Association, Fukui, Japan

M. Fukushima
Morimoto Elementary School, Kanazawa, Japan

## INTRODUCTION

The study of playing dynamics is essential for a biomechanical profile of Japanese drum (*wadaiko*) playing that aims at quantifying the specific movements involved in rhythmic exercises. This study examined the force exerted to the ground as the drummer attempted to hit a given sound while performing certain musical tasks. Therefore, the purpose of this study was to investigate the effects of tempo and drumming pattern of Japanese drum playing (JDP) upon ground reaction forces.

## METHODS

Subjects for this study were seven virtuosi male Japanese drummers (age 29.9±5.5 yr). In order to record three-components of the forces; vertical, anterior-posterior, and left-right force, each subject performed to play a Japanese drum with a pair of large wooden sticks (0.53 kg) for five minutes, standing on a strain-gauge type force plate. And also each subject was instructed to play the drum at three tempos (90, 120, and 150 beats/min) by different drumming patterns (single sticking, S; "*Futatsu-uchi*" , F; "*Mitsu-uchi*" , M) as well as to keep a sound level (120 dB). The sound level was measured using a digital sound level meter, and was immediately displayed to the subjects in order to keep a constant sound level. The order of all trials was randomized and drumming tempo was indicated by a metronome. The basic components of ground reaction forces for all trials during JDP are defined as maximum vertical (F1), minimum vertical (F2), anterior peak (F3), posterior peak (F4), right peak (F5), and left peak (F6) forces.

## RESULTS AND DISCUSSION

The force time curves during JDP showed cyclic patterns with clear peaks appearing in vertical, anterior-posterior, and left-right components. ANOVA revealed that the maximum and minimum vertical (F1 and F2) and the anterior-posterior peak (F3 and F4) forces were significantly different between drumming patterns, and that F1 showed significant difference between tempos ($p<0.05$, Fig. 1). It was suggested that the difference of drumming patterns during JDP would effect the displacement of the body's center of gravity in both the vertical and the anterior-posterior directions.

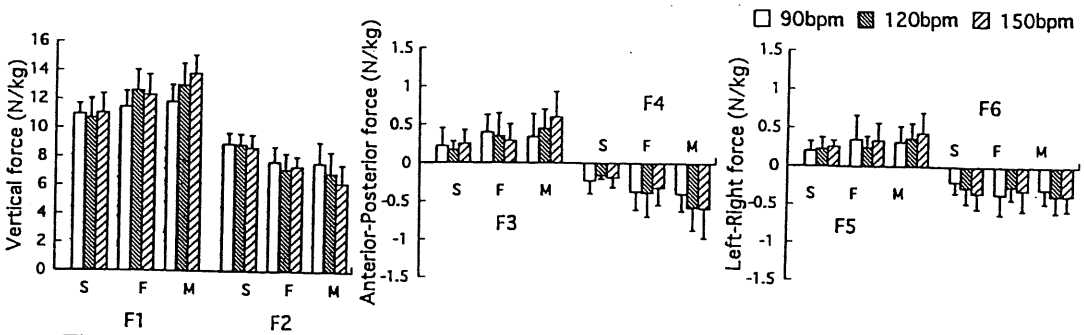



Fig 1. Ground reaction forces of each parameter during five-minute Japanese drum playing exercise (Mean±SD of cycles, S: Single sticking, F: "*Futatsu-uchi*" , M : "*Mitsu-uchi*" )

# HOW TO QUANTIFY POSTURAL CONTROL WITH FORCE PLATE DATA?

Annica Karlsson
Systems and Control group, Uppsala University, Sweden

## INTRODUCTION

Different measures based on the ground reaction force (F) vector and the centre of pressure (COP) are often used to characterise a persons postural control[1]. These measures quantify the magnitude of the body sway, but do not tell us *how* the person controls the sway. A model-based method, which needs force plate data only as input, can be used to determine to what extent the ankle strategy is used to control the sway[2]. The basic idea of the method is to compare the acceleration of a single inverted pendulum's centre of mass (COM) with the same length and mass as the subject, denoted $\ddot{x}_{model}$, with the subject's actual acceleration of the COM, denoted $F_x/m$. The aim of the study was to compare measures based on the pendulum model with two more commonly used measures - the standard deviation (SD) of the F and the COP.

## METHODS

20 normal subjects, two stroke patients, and a MMC child took part in the study. Three measurements in normal standing (30s) on a force plate were made on each subject. Only the sagittal plane was considered in the analysis. The following measures were studied:

- **SD(Fx/m)** - the standard deviation of the antero-posterior F divided by the body mass.
- **SD(COPx)** - the standard deviation of the antero-posterior COP (highpass filtered to remove drifts).
- **r** - the correlation coefficient between $\ddot{x}_{model}$ and $F_x/m$.
- **q** - the ratio $\frac{SD(\ddot{x}_{model}-F_x/m)}{SD(F_x/m)}$, was analysed to study if the two accelerations have the same amplitude.

A person using the ankle strategy corresponds to a correlation coefficient close to 1 and a ratio close to 0. See Fig.1.

## RESULTS

The acceleration of the COM for two of the persons is shown in Fig. 1. One is using the ankle strategy to a large extent (left-hand diagram, **r** = 0.97 and **q** = 0.24), and the other is using it to a lesser extent (right-hand diagram, **r** = 0.89 and **q** = 0.54).

The results for the control group and the three patients are summarised in Table 1. The results for the control group showed a rather large variability. Despite this we found that all four measures for the MMC patient deviate from the control group. The **SD(Fx/m)** and the **SD(COPx)** for the stroke patients are in the same order as the control group's, while the correlation coefficient, **r**, and ratio, **q**, are not.

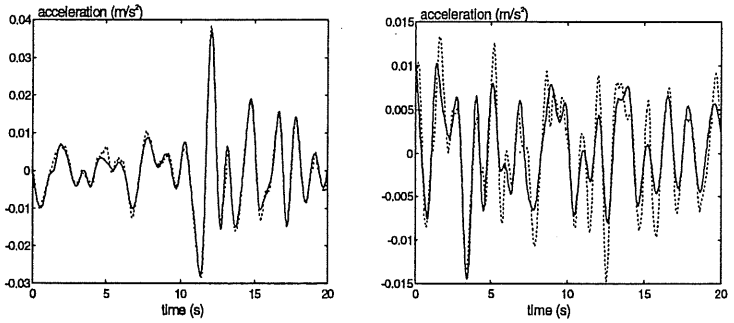



*Fig. 1. The true acceleration, $F_x/m$ (solid line), and the acceleration predicted by the model, $\ddot{x}_{model}$ (dotted line), for two subjects.*

| | **SD(Fx/m)** ($m/s^2 \cdot 10^{-3}$) | **SD(COPx)** ($m \cdot 10^{-3}$) | **r** (%) | **q** (%) |
|---|---|---|---|---|
| Mean ± S.D. | 11.5 ± 2.8 | 2.6 ± 0.6 | 91.8 ± 3.0 | 40.6 ± 6.8 |
| Stroke 1 | 9.7 | 2.4 | 96.2 | 28.3 |
| Stroke 2 | 12.7 | 2.2 | 97.7 | 21.8 |
| MMC | 29.7 | 5.8 | 96.0 | 30.9 |

*Table 1. The results for the control group is summarised in row 1 (mean and standard deviation between 20 subjects). The following three rows shows the results for the three patients.*

## DISCUSSION

The model measures quantify the movement strategy while the 'standard' measures quantify the amplitude of the body sway. Even though a patient does not sway more than normal he or she might use a different strategy. Our conclusion is that both kind of measures are of importance if the aim is to characterise a subjects postural control.

## REFERENCES

[1] P.A. Goldie, T.M Bach and O.M. Evans. Force platform measures for evaluating postural control: reliability and validity. Arch. Phys. Med. Rehab. 70 (1989) 510-517.

[2] A. Karlsson and H. Lanshammar. Analysis of postural sway strategies using an inverted pendulum model and force plate data. Gait & Posture, 5/2, 1997 (to appear).

# STRUCTURAL AND PARAMETRICAL VARIATIONS OF THE VERTEBRAL COLUMN MODEL FOR THE DYNAMICAL LOADS ASSESSMENTS

V.A.Sholukha [1], A.V.Zinkovsky [1], K.J. van Zwieten [2], P.L.Lippens [2], A.A.Ivanov[1]
[1] St. Petersburg State Technical University, St. Petersburg, Russia
[2] Limburgs Universitair Centrum, Diepenbeek, Belgium

## INTRODUCTION

Investigation of human skeletal-muscular apparatus (SMA) reaction on dynamics loads is especially actual for multilink joints chains, for example, for lower back and neck parts of the spine (Croft et al.,1992). Main investigation methods are based on analysis of acceleration and velocity measurements for shock loads, or on finite-element models of vertebrae interaction. In this work the problem is considered from the point of view of synthesis of goal-oriented motions of a system of bodies (Zinkovsky et al., 1996). This approach allows to estimate the dynamic loads on the spine for normal and extreme motion conditions of the SMA.

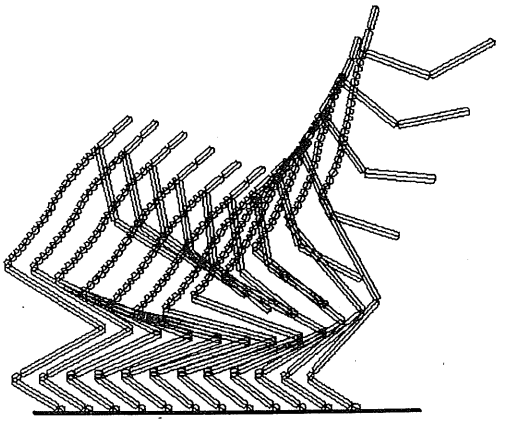


Fig.1. Kinematic scheme of the 30 links model motion.

## METHODS

The main investigation method is based on computer modelling of human motions with preset values, form and period of action of external loads, provided by means of non-stationary dynamic constraints. This approach allows to variate structure, form and visco-elastic parameters of links and joints conserving the kinematics of motion. For modelling of visco-elastic interactions of the SMA with different objects, shock loads are imitated according to the winkler model.

## RESULTS

In Figs. 1-4 there are presented results of calculations for a 30-element model. For the motion simulation components of the ground reaction force and movement of the toe were preset. There are presented graphics of bending moments and forces acting in the model spine, consisting of 21 element with concentrated at the cylindrical joints visco-elastic characteristics. Calculations have been made for different forms of the spine and types of joints in the saggital and frontal planes. Variations of their form and parameters allowed to assess critical loads for motions of different types. For the impact phase of the motion rotation and displacement model of the joints was used. It gives more smoothed results for the first impact impuls after landing at the visco-elastic surface.

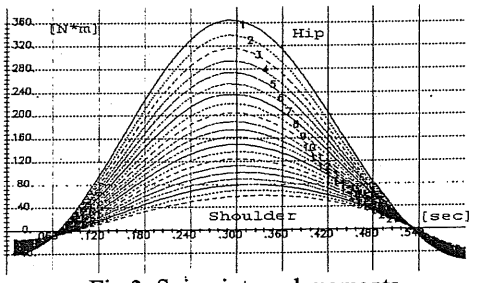



Fig.2. Spine internal moments.

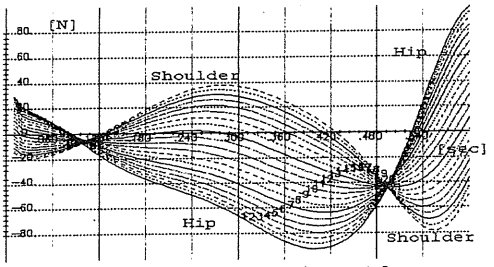



Fig.3. Spine internal horizontal forces.

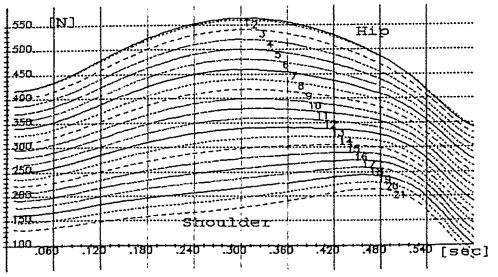



Fig.4 Spine internal vertical forces.

## DISCUSSION

The suggested method of modelling is based on solution of the motion synthesis problem. This, from one hand, leads to necessity of realization of accurate procedure of integration of systems of implicit differential-algebraic equations, but, on the other hand, allows to variate structure and parameters of the model conserving its integral characteristics of motion and goal-oriented kinematics, usual for most of professional motions.

## REFERENCES

Croft, A.C. (1992) The cervical acceleration/deceleration syndrome. In Steigerwald DP, Encinitas, CA: Keiser Publishing Co.

Zinkovsky, A.V., Sholuha, V.A., Ivanov, A.A. (1996) Mathematical Modelling and Computer Simulation of Biomechanical Systems. World Sci. Publ. Co. Ltd., London, UK, 212 pp.

# LOW BACK FORCES AND MUSCLE ACTIVITY IN NEW ZEALAND AND AUSTRALIAN TRADITIONAL SHEEP SHEARING TECHNIQUES

R.N. Marshall and A. Burnett
The University of Auckland, New Zealand,
and The University of Western Australia, Australia

## INTRODUCTION

Traditional sheep shearing techniques require the shearer to adopt a stooped position while manipulating a sheep positioned between the feet and using mechanical shears to remove the fleece as quickly and efficiently as possible. Removing the fleece from a sheep typically takes between 45 and 200 seconds, depending upon the shearer's expertise and type of sheep. During this time the shearer is active, although the trunk generally remains approximately horizontal. A professional shearer will shear between 150 and 300 sheep in each ten-hour day. Sheep shearing has the potential to generate substantial low back stress. Gmeinder (1986) noted approximately 20% of all claims for compensation from sheep shearers were due to back injury, however, due to the debilitating nature of low back pain, it was found that compensation payouts far outweighed these statistics. In a preliminary analysis of the National Worker's Compensation Statistics Database by Worksafe Australia in 1995, sheep shearing was found to be one of the most hazardous of all industrial occupations.

The aim of this work was to describe the compressive and shear forces in the L5/S1 joint of the lower back as well as the electrical activity of major trunk and lower limb muscles for Australian and New Zealand traditional shearing techniques.

## METHODS

Five experienced male shearers were used as subjects. Seven points in the shearing action were examined, and these were blows from the crutch, tail, neck, shoulder, long blow, front leg/whip side and last leg. Each subject sheared two sheep using each style. Three video cameras operating at 60 Hz were used to record the shearer's actions. For each subject trial, the left and right ankles, knees, hips, shoulders, elbows and wrists were digitised from the videotapes using an Ariel Performance Analysis System (APAS) and the data were combined to provide three dimensional coordinates of the digitised points.. Two seconds of electromyographic data from eight trunk and lower limb muscles were recorded and synchronised with the videotapes.

The joint coordinates were then used to calculate 3-D segment and joint angles for the lower leg, thigh, trunk, knee and hip. The forces acting at the L5/S1 joint were evaluated using the University of Michigan 3-D Static Strength Prediction Program™ (1993).

The muscle activity was normalised to selected body postures in order to permit comparisons between subjects. Data were compared within and between subjects and shearing styles.

## RESULTS

None of the sampled body positions showed L5/S1 compression forces over the NIOSH Design Limit (3329 Nm), although all were above 2000 Nm, with average values (± 1 SD) of 2529 (241) Nm for the Australian technique and 2640 (287) Nm for the New Zealand technique. These values are 5.75 and 6 times the compressive forces calculated for erect standing with the same hand forces as in shearing. It was shown that compressive, shear and torsional forces at the L5/S1 joint are proportional to increases in body size and mass, although it also appears that increasing mass has a larger effect than increasing height. An analysis of horizontal forces at the hands indicated dangerous force levels may be generated when both hands are used to push in the same direction while in a stooped position.

Limited muscle activity from the erectore spinae at L3 level and hamstring muscles suggested the shearers used ligamentous restraints or intrinsic muscle stiffness to maintain pelvic and trunk positions in the stooped postures. The erectore spinae muscle at T9 displayed high levels of activity over most of the shearing action. These results raised concern over the control of hip and trunk flexion. We surmise that the lower back ligaments may be progressively lengthened over the day, with a consequent progressive reduction in their ability to provide support and control for the spine. This may increase the potential for injury when stressful movement are performed later in the day. A second concern relates to the long term effects of these postures on the integrity of the spine.

Gmeinder, G.E. (1986). Back complaints among shearers in Western Australia: A pilot study. Australian Journal of Physiotherapy, 32, 139-144.

# RANGE OF MOTION AND MUSCLE ACTIVITY DURING ACTIVE TRUNK ROTATION OF TRACTOR DRIVERS AND OFFICE WORKERS.

A. Bodén and K. Öberg.
Department of Agricultural Engineering, Swedish University of Agricultural Sciences, Uppsala, Sweden

## INTRODUCTION

When driving a tractor in the field using an implement at the rear, the driver often has to sit twisted looking backwards to overview and control the implement. Twisting of the trunk has been identified as a risk factor in the incidence of low back pain (Riihimäki et al., 1989). The purpose of this investigation was to measure the range of motion and muscle activity and find out whether there are any differences between tractor drivers and office workers that might be due to the twisted posture when driving the tractor.

## METHODS

The exertion was made from neutral position to the maximum twisting angle, both clockwise (cw) and counter-clockwise (ccw). To be able to interpret the EMG data the subjects also performed maximum voluntary contractions (MVC). These were done with the shoulder frame locked at -40 degrees (direction of action: ccw) and 40 degrees (direction of action: cw, figure 1).

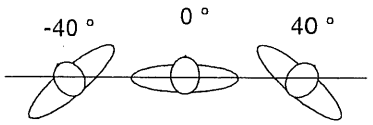



Figure 1. MVC was done with the shoulder frame locked at -40 degrees and 40 degrees.

Ten healthy male tractor drivers and ten healthy male office workers volunteered for the study. Surface EMG electrodes were applied to the skin in a bipolar configuration. The muscles measured were left and right erector spinae, left and right external oblique and left and right rectus abdominis. Description of the measurement equipment has been published earlier (Öberg et al., 1994). The EMG-signal was sampled at 500 Hz, band-pass filtered (10-500 Hz), digitised, rectified and low-pass filtered at 3 Hz.

## RESULTS

The maximum twisting angle is 55 degrees (both directions) for the tractor drivers and 54 degrees (cw) and 51 degrees (ccw) for the office workers. The differences are not significant. The external oblique muscles and erector spinae muscles show the alternating pattern when doing the exertions in the two directions. This pattern can not be seen for the rectus abdominis muscles. The data concerning muscle activity are to be analysed further.

## DISCUSSION

The differences in range of motion between the work groups are small. There may be several explanations of this. First, there might not be any differences according to twisting angle. Second, there might be differences, but they are hidden by the effects of whole-body vibrations. Seidel and Heide (1986) suggest a two-phase development of long-term effects from whole-body vibrations. First the range of motion increases because of muscular insufficiency and height loss of intervertebral spaces. This increase in mobility leads to a high strain during whole-body vibrations, and this long-term strain causes degenerative changes resulting in a decrease of mobility (second phase). Third, the sample of 20 subjects might be too small to show whether there are differences or not. However, the results are valuable when continuing the work concerning the twisted posture during tractor driving. Together with investigations of trunk resistance and trunk strength, this will be a contribution to useful information in order to improve the drivers working conditions.

## REFERENCES

Öberg, KET. Bodén, A. Hansson, P-A. (1994) EMG and torque measurements of trunk rotation when sitting. In *Proceedings of the 12th Triennial Congress of the IEA.* Toronto, Canada.

Riihimäki, H. Sakari, T. Videman, T. Hänninen, K. (1989) Low back pain and occupation. *Spine* **14,** 204-209.

Seidel, H. and Heide, R. (1986) Long-term effects of whole-body vibration: a critical survey of the literature. *Int Arch Occup Environ Health* **58,** 1-26.

# TRUNK MUSCLE STRENGTH IN ECCENTRIC AND CONCENTRIC LATERAL FLEXION

Qiang-Min Huang and Alf Thorstensson
Departments of Neuroscience, Karolinska Institute and Human Biology, University College of Physical Education and Sports, Stockholm, Sweden

## INTRODUCTION

Eccentric actions for the trunk muscles are common in daily life activities, such as lowering a load. Such actions may increase the risk of back injury, particularly in asymmetric loading situations. It is therefore important to characterize also the eccentric strength capacity of the trunk muscles. Recent studies have demonstrated higher forces in lowering than lifting (Cresswell and Thorstensson 1994, Huang *et al.* 1996). However, in lateral, asymmetric, lowering, the force curves showed a different position dependency than in lifting. The primary aim of this study was to characterize the eccentric strength performance during lateral trunk flexion, particularly the influence of trunk position and speed of movement.

## METHODS

Six healthy male volunteers were investigated. Their mean age, body mass and height were 28 ± 4 years, 76 ± 6 kg and 1.80 ± 0.05 m, respectively. The Swivel Bench (Thorstensson and Nilsson 1982) was combined with the Spark machine (Seger *et al.* 1988) to allow for eccentric trunk muscle strength to be measured, in addition to concentric and isometric. The subjects were positioned supine with the upper body on the movable part and the lower body on the rigid part of the Swivel Bench, with the pivot point at the $L_5$-$S_1$ level. Force was measured with a load cell and angular displacement of the trunk with a potentiometer. The produced strength (torque) of the trunk muscles on the left side was calculated over the whole range of motion (50 - $60^0$). Both concentric and eccentric actions were performed at the set angular velocities of 15, 30, 45 and 60 $^0$/s. In addition, maximal isometric actions were carried out at five different angles of lateral trunk flexion: 20, 10, 0, -10 and $-20^0$ ($0^0$ = straight body).

## RESULTS

The torque versus trunk angle relationship was similar, irrespective of mode and speed of muscle action. The position for peak torque ranged 9-$14^0$ to the contralateral (right) side under both eccentric and concentric conditions (Fig. 1). Also, the highest isometric torque occurred at 10 and $20^0$. Eccentric strength was higher than the isometric, which, in turn, exceeded the concentric strength at corresponding trunk angles. Changes in velocity did not significantly effect the torque output in eccentric actions (Fig. 1). In concentric actions, on the other hand, torque values decreased with increasing velocity (Fig. 1).

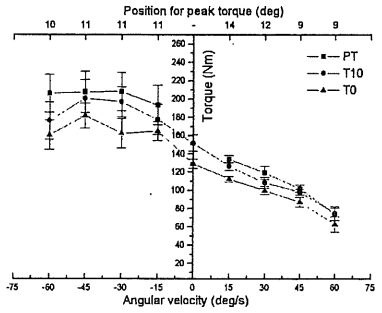



**Fig 1.** Peak torque (PT) and position specific torque at 10 and $0^0$ (T10 and T0) versus angular velocity. The top axis shows the mean positions for each PT value.

## DISCUSSION

The position dependency of the lateral trunk flexor strength appears to be mainly related to mechanical factors, such as length of the muscles and their lever arms. The type and speed of muscle action seem to play a minor role. This is in line with earlier results for single muscle groups, such as the knee extensors (Westing and Seger 1989). However, when performing a lateral pull with a hand-held load in standing, peak eccentric pulling force occurred in a position where the lateral trunk muscles were shortened (Huang *et al.* 1996). Thus, other factors, such as shoulder or finger strength and/or balance requirements, determine the performance in that situation. The speed dependency of the torque curve resembles that earlier obtained for the knee extensors. The lack of an increase in torque with eccentric speed could result from an inhibitory mechanism preventing the utilization of the full force producing capacity in eccentric muscle actions (cf. Westing *et al.* 1990).

## REFERENCES

Cresswell, A.G. and Thorstensson, A. (1994) Changes in intra-abdominal pressure, trunk muscle activation and force during isokinetic lifting and lowering. *Eur J Appl Physiol* **68**, 315-321.

Huang, Q-M., Sato, M., and Thorstensson, A. (1996) Pulling force in lateral lifting and lowering. *Ergonomics* (submitted).

Seger, J. Y., Westing, S. H., Hanson, M., Karlson, E. and Ekblom, B. (1988) A new dynamometer measuring concentric and eccentric muscle strength during accelerated, decelerated, or isokinetic movements. *Eur J Appl Physiol.* **57**, 526-530.

Thorstensson, A. and Nilsson, J. (1982) Trunk muscle strength during constant velocity movements, *Scand J Rehab Med.* **14**, 61-68.

Westing, S.-H. And Seger, J.Y. (1989) Eccentric and concentric torque-velocity characteristics, torque output comparisons, and gravity corrections for the quadriceps and hamstring muscles in females. *Int J Sports Med* **10**, 175-180.

Westing, S.H., Seger, J.Y. and Thorstensson, A. (1990) Effects of electrical stimulation on eccentric and concentric torque-velocity relationships in man. *Acta Physiol Scand* **140**, 17-22.