International Society of Biomechanics
WORKING GROUP ON FUNCTIONAL FOOTWEAR

Third Symposium on FOOTWEAR BIOMECHANICS

August 21-23, 1997
Tokyo Metropolitan University, Tokyo, Japan

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The Working Group wishes to express its sincere appreciation to the organisations who contributed to the financial support of this Symposium. Without their generosity the Tokyo meeting would not have occurred. In each case, corporate sponsorship was made possible by the efforts of one or more individuals. Special thanks therefore to Simon Lüthi, Mario Lafortune, Darcy Winslow, Yasunori Kaneko, Peter Seitz and Ned Frederick.

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The Third Symposium on Footwear Biomechanics was held as a satellite meeting of the XVI Congress of the International Society of Biomechanics. The meeting was held at the Tokyo Metropolitan University, August 21-23, 1997.

Unlike previous symposia, which have relied heavily on invited speakers, this meeting consisted almost entirely of original work that was submitted for peer review. The program included 32 presentations of original work and several extended discussion sessions. As the abstracts published here will demonstrate, the quality of the work presented at the Symposia continues to increase, reflecting the rising expectations of researchers in this field. The range of topics covered reflected broad interests from sophisticated new models and analytical methods to reports of experimental results that challenged existing paradigms. We were also challenged by the practitioners who have are trying to apply the science on a daily basis and came seeking answers to fundamental questions.

The generosity of the Symposium's sponsors made it possible to invite guest speakers, to offer travel assistance to presenters who would otherwise be unable to attend and to minimise registration fees. In addition, Adidas International and NIKE, Inc. offered research prizes for the most significant papers in applied research and basic research respectively.

At a business meeting held during the Tokyo Symposium, there was a lengthy discussion about the future of the group. After a healthy discussion of all the alternatives, it was agreed unanimously that the group should apply to the ISB for recognition as a Technical Group. At their next meeting, also held in Tokyo, the ISB Council voted unanimously to grant the group recognition as the "Technical Section on Footwear Biomechanics". We are now the ISB's third (and currently the largest) technical group.

The Symposium was opened and closed in traditional Japanese style. We enjoyed a generously catered Welcome Reception, courtesy of Mizuno Corporation which included a traditional sake barrel opening ceremony. We were also fortunate to have a tea house at the Symposium site and everyone was able to participate in a traditional Japanese tea ceremony as part of the final dinner. The violence of the breaking sake barrel and the tranquility of the tea ceremony providing contrasting bookends to an enjoyable meeting.

We now look forward to the Fourth Symposium, which will be held in 1999 as a satellite meeting of the XVII ISB Congress in Calgary, Canada.

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Yasunori Kaneko, Symposium Co-Chair  
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**Applied Research Prize**

*presented by adidas International*

The prize for the paper which, in the view of the judges, made the most significant contribution to applied research in footwear biomechanics was awarded to

Geza F. Kogler, F.B. Veer, S.E. Solomonidis, J.P. Paul  
Bioengineering Research Laboratory,  
Southern Illinois University School of Medicine, Springfield, IL, USA  
*“The influence of medial and lateral orthotic wedges loading of the plantar aponeurosis. In vitro study.”*

**Basic Research Prize**

*presented by NIKE, Inc.*

The prize for the paper which, in the view of the judges, made the most significant contribution to applied research in footwear biomechanics was shared by two presenters:

Mark J. Lake, B. Bimson, D. Gerssen, S. Nooy, N. Roberts  
Liverpool John Moores University, Liverpool, UK  
"Relating forefoot structure to dynamic plantar pressure distribution during locomotion."

Ian C. Wright, R.R. Neptune, A.J. van den Bogert, B.M. Nigg  
Human Performance Laboratory  
The University of Calgary, Calgary, Canada  
“The regulation of impact forces in running”
Thanks to Dr. Keith Williams, Dr. Bart van Gheluwe and Dr. Ton de Lange for conscientiously adjudicating the awards.

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ABSTRACTS
THE INFLUENCE OF UNANTICIPATED CHANGES IN ATHLETIC SHOE MIDSOLE HARDNESS ON LOWER LIMB MUSCULAR PREPARATION FOR IMPACT LOADING.

Mark. J. Lake and Mario A. Lafontune *

School of Human Sciences, Liverpool John Moores University, U.K.; * Nike Sport Research Laboratory, Oregon, U.S.A.

Introduction

Load on the body during locomotion has been quantified using force platforms and accelerometers to record the compressive load and segmental shock experienced on ground contact. Relatively little attention has focused on adaptation to the impact loading conditions and the neuromuscular modifications that may be associated with changes in impact severity. The relative contributions of mechanical and neuromuscular factors to the impact loads experienced remain to be determined. The time course of adaptive change is important from the perspective of injury prevention but few studies have examined adaptation effects over time. Adaptive effects due to fatigue have been examined (2) and adaptations in muscular strategies to specific mat conditions have been demonstrated for drop landings (1,8). These studies have found associated changes in impact severity alongside the neuromuscular adjustments. To determine the relative importance of neuromuscular factors, unanticipated perturbations of the conditions of impact have been employed (3,6). Such studies have established that the onset of muscle activity is planned in advance and based upon the severity of prior impacts. The aim of this study was to examine the muscular preparation for lower limb impact following an unexpected change in athletic footwear midsole density. Few studies have examined the short-term responses of the body to a modification in athletic shoe cushioning in a controlled manner. These responses might provide useful insight into the interaction between mechanical and neuromuscular factors that govern impact loading severity.

Methodology

A human pendulum apparatus was used to control the velocity (1 m/s) and lower limb posture at impact (5). The severity of impact was determined by calculating the peaks, times to peak and transient rates of wall reaction force (wall mounted force platform), shank and head shock (skin mounted accelerometers) all sampled at 1500 Hz.

Neuromuscular preparation for impact was quantified by summing the electromyographic (EMG) activity of five lower limb muscles (tibialis anterior, gastrocnemius, vastus medialis, rectus femoris and biceps femoris) 210 ms prior to impact. Ten injury-free male subjects volunteered for the experiment which involved numerous visits to the laboratory so that the subjects gained experience of impacts associated with a specific footwear condition. Two footwear conditions were used which had different midsole properties as confirmed by mechanical impact test results: 9.18 - 9.45g (‘soft’ shoe) and 10.6 - 10.8g (‘hard’ shoe). Subjects were monitored for both a soft to hard (SH) footwear change and a hard to soft (HS) footwear change. Before each shoe exchange all subjects were accustomed to a specific type of footwear by undergoing two weeks of adaptation sessions consisting of pendulum impacts and treadmill running. Frequent fake measurements of shoe insole temperature allowed the footwear transition to be totally unanticipated. Immediate and short-term changes in the variables of impact severity and pre-contact muscular activity were monitored. The short-term
adaptation period included a 10 minute bout of treadmill running, with the treadmill placed next to the human pendulum apparatus.

Results

Impact severity was increased by 5 - 20% by the soft to hard transition. In general, changes in severity occurred immediately after the transition only. There were no significant changes in the pre-contact activity of the lower extremity muscles over all the data collection periods. For the hard to soft transition, the relative adjustments in impact severity were similar to those for the soft to hard transition but in the opposite direction. Almost all the changes in impact loading variables occurred immediately after the transition only. Once again, their were no significant changes in pre-contact muscle activity across the all the data collection periods. The degree of change in impact loading variables was generally close to the difference in mechanical properties of the midsole materials already determined by the mechanical impact tester results (15%).

The results indicated that the severity of lower extremity impact was modified due to the mechanical characteristics of the shoe midsole and short-term muscular adaptation effects were minimal or not present. Pre-contact muscular activity of selected lower extremity muscles was not modified after the change in footwear or after accommodation by treadmill running.

Discussion

The response to a switch in footwear appeared to be mainly mechanical in nature and dependent upon the degree of change in midsole properties of the test shoes. The absence of change in muscular preparation prior to impact indicates that the unexpected adjustment in impact severity was either not perceived by the subjects or they deemed such a change as unnecessary. If the change in impact severity was not perceived then our ability to discern such changes is limited. Muscular preparation for impact loads, over the range experienced in this study, appears to be relatively independent of shoe midsole mechanical properties. The experimental shoes represented the range of shoe ‘hardness’ commercially available and, therefore, for in vivo comparative shoe testing an assumption of constant neuromuscular control might be warranted. This notion is supported by additional pendulum research where a similar lack of change in pre-contact muscle activity was demonstrated prior to impact on interfaces of different but expected mechanical properties. Furthermore, no significant adjustments in lower extremity muscular activation magnitude has been found during running in footwear with different midsole characteristics (4,7). When impact severity has been substantially larger, as experienced during drop jumping, neuromuscular adjustments to impacting interface hardness have been found (8).

References.

TIME-FREQUENCY ANALYSIS OF IMPACT SHOCK DURING RUNNING

A. Rakotomamonjy, M. Barbaud, M Tronel, P. Marché
Laboratoire Vision et Robotique, IUT de Bourges, France.

Introduction
Repetitive impact shocks during running are suspected to cause joint diseases. The analysis of those impact shocks has been possible owing to accelerometers mounted on the body. A classical spectral analysis based on the Fourier Transform is then used for studying the frequency components of the shock wave (1) (2). However, this mathematical tool is well adapted for signal whose characteristics do not vary over time, which is not the case of the tibial acceleration. Our aim in this study is to investigate the utility of a more adapted tool, the quadratic Time-Frequency Representation (TFR) (3) of a signal, for studying a transient signal like tibial acceleration.

Method
Experimental data have been obtained by mounting a low mass accelerometer fastened on the medial aspect of the tibia. Ten volunteers have participated to the experiment. They have been asked to run on a 20 m runaway at their preferred speed. All subject have run barefoot and with their own running shoes. The tibial acceleration has been sampled at 450 Hz and 256 points of the signal including the impact shock have been extracted for further analysis.

Quadratic Time-Frequency Representations allow to spread the energy of a signal over a time-frequency plane and permit to follow up the evolution of the time-varying spectral content of a transient signal. Among the large choice of TFR (3), we have choose the Wigner-Ville transform because of the large amount of mathematical properties that it satisfies and for its easiness to compute. The Wigner-Ville Transform WVT of a signal \( s \) is defined as:

\[
WVT(t, f) = \int_{-\infty}^{+\infty} s(t + \frac{\tau}{2}).s^*(t - \frac{\tau}{2}).e^{-j2\pi ft}.d\tau
\]

where \( t, f \) and \( s^* \) denote respectively the time variable, the frequency variable and the conjugate of the signal \( s \). Once processed, the WVT is represented, in decibel, over a time-frequency plane. The characteristics of the impact can be retrieved by examining the representation. Then, the quantitative analysis of the impact shock consists in measuring the cut-off frequency of the impact at -3 dB and -7 dB which are compared to classical parameters such as mean power frequency (MPF) or magnitude of the impact.

Results
TFR of barefoot and shod shock are presented on the Figure 1 and 2. They clearly demonstrate that during the impact, frequencies up to 50 Hz can be generated and that the major part of the energy of the signal lies within the 0-20 Hz band. Frequency contents of the active and passive phases are dissociated on the TFR, as they do not occur at the same time. Therefore, it is easy to remark that power spectral density due to the low motion of the limb is below 10 Hz.

As one may expected, wearing running shoes attenuate the magnitude and the MPF of the impact. Cut-off frequencies of the shock have also been considerably reduced (19 Hz and 54 Hz). Figure 2 illustrates the effect of the shoes on the impact. One can see that energy of the signal has been spread over the time and that the shock seems to be smoother as there is no peak on the TFR.
<table>
<thead>
<tr>
<th>Variable</th>
<th>Magnitude (g)</th>
<th>MPF( Hz)</th>
<th>Cut-Off -3 dB</th>
<th>Cut-Off -7 dB</th>
</tr>
</thead>
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<tr>
<td>Barefoot</td>
<td>7.22 ± 1.22</td>
<td>11.75 ± 1.7</td>
<td>48.58 ± 14.24</td>
<td>121.26 ± 25.83</td>
</tr>
<tr>
<td>Shoes</td>
<td>5.35 ± 1.42</td>
<td>8.54 ± 1.37</td>
<td>29.50 ± 9.12</td>
<td>67.96 ± 21.53</td>
</tr>
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Table 1: mean values and standard deviation from the 10 subjects.

Figure 1: Contour plot of barefoot shock.  Figure 2: Contour plot of shod shock.

**Discussions**

The use of TFR has revealed that frequencies of the impact shock raise higher than the bandwidth 12-20 Hz found by Shorten et al.(1)(2). This discrepancy is mainly due to an inappropriate tool. TFRs allow to distinguish the true frequencies and relative energy of the impact shock as, unlike the Fourier Transform, time information are still in the representation. Therefore, those low-energy high frequency components of the impact shock can be brought out and the effects of shoes on the shock are calculated in terms of cut-off frequency. The main advantage of TFR is its capability of giving frequency contents of the signal over the time. Therefore, further investigations can be done concerning the attenuation of the shock as, owing to this mathematical tool, correlation are possible between kinematics adaptation (e.g knee flexion) and spectral attenuation of the shock.

**References**

3) Hlawatsch et al. (1992), Linear and quadratic time-frequency signal representations, IEEE SP magazine, April, 21-67.
PASSIVE REGULATION OF IMPACT FORCES IN HEEL-TOE RUNNING
I.C. Wright, R.R. Neptune, A.J. van den Bogert, B.M. Nigg
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Introduction
Impact forces during running are forces due to the collision of the foot with the ground. These forces have long been thought to contribute to injuries in running. Cushioning is the reduction of these forces, and has been of interest to the footwear industry as it has been thought that cushioning could be achieved by using soft shoe sole materials. This belief was based on material tests of shoes which showed increases in impact forces with increased shoe sole hardness. However, when people run in shoes of varying hardness this same trend in impact force magnitudes is not observed. This result may be explained by adaption or changes in the movement. These changes could be active changes resulting from changes in the initial conditions at touchdown or changes in muscle activities. This active adaptation has often been suggested to be the mechanism of impact force regulation in running, but there is not strong evidence supporting this. Passive changes in the movement are ones that arise without changing muscle activities or touchdown initial conditions. Such changes in the movement are a consequence of the differences in the forces caused by the different shoe sole hardness. This passive regulation has not been shown to account for impact force regulation.

The purposes of this study were to determine if passive mechanisms can account for impact force regulation during heel-toe running with shoes of varying midsole hardness, and if so, how does this passive impact force regulation affect internal forces. This will help us understand how shoe properties affect movement during running, and how potentially injury-causing forces are affected by impact force regulation.

Methods
A simulation model was developed by compiling data from several literature sources into the forward dynamics simulation software package DADS (version 8.0, CADSI, Coralville, IA). The three dimensional model of the lower extremity was based on an existing static simulation model and inertial properties from literature. The model had rigid segments representing the toes, foot, talus, shank, patella, and thigh of the support leg, and a rest-of-body segment. Twelve functionally independent muscle groups were included in the model, the paths of which were based on a past simulation study. The force-length-velocity characteristics of the muscles were represented by Hill-based lumped parameter models. The contact between the foot and the ground was represented by 66 contact points distributed across the underside of the foot. The visco-elastic mechanical properties of the shoe were based on published mechanical tests of running shoes. Initial conditions were based on nine recreational male runners heel-toe running at 4 m/s. Limb segment kinematics were recorded using video motion analysis, while ground reaction forces were recorded using a force platform. Positions and velocities of the body segments at heel strike were used as the initial conditions. Step functions were used for the muscle stimulations. The times of onset and the levels of stimulations were determined by minimizing the differences between the simulated and measured kinetics and kinematics for one subject.

Simulations were performed to determine if passive force regulation could account for impact force regulation. Simulations were performed for each of the nine subjects with the
initial conditions and the muscle stimulations held constant while the shoe sole hardness was varied between the soft and the hard conditions.

**Results and Discussion** Between the soft and the hard shoe sole conditions, there were differences in the ground reaction forces for each subject specific set of initial conditions. However, there were no significant differences in the peak vertical impact force between the soft and hard shoe conditions. This matched past experimental results. The rate of loading was found to be greater for the hard shoe sole condition than for the soft shoe condition for each of the nine subject specific sets of initial conditions. This too matched experimental results. This suggests that the simulation model responded like subjects to changing in shoe sole hardness. Furthermore, since for these simulations there was no opportunity for active changes, this shows that passive mechanisms may account for impact force regulation.

For the simulations in this study, it was found that the initial rate of knee flexion was greater for the hard shoe condition than for the soft shoe condition for all nine subject specific sets of initial conditions. This also matched experimental results. Since the increased rate of knee flexion decreased the amount of mass involved in the collision during the impact phase, the rate of flexion caused a decrease in the effective mass of the leg. This, therefore is a possible mechanism of impact force regulation.

An investigation of the peak forces in an arbitrary selection of muscles during the impact phase revealed that internal forces changed with this passive force regulation. For the tibialis anterior muscle, the forces were greater for the hard shoe than the soft shoe for all nine subject specific sets of initial conditions. The opposite was true for the peroneus muscle, whereas for the hamstrings muscle, there were no significant differences between the two shoe conditions. This suggests that internal forces may change with changing shoe hardness but changes in internal forces are not represented by changes in externally measured ground reaction forces.

**References**

STEP TO STEP VARIABILITY OF IMPACT FORCES AND REARFOOT MOTION DURING TREADMILL RUNNING

Kersting, U.G., Brüggemann, G.-P.
Institute for Athletics and Gymnastics, German Sport University - Cologne; Germany

Introduction
Several experimental studies and modelling approaches implicate a close association between impact forces at touch-down and rearfoot motion during running (Stacoff et al., 1988; Nigg & Bahlsen, 1988). However, recent publications have shown that changes in shoe construction designed to alter the pronatory movement in the subtarsal joint did not influence impacts substantially (DeWit et al., 1995; Hennig & Milani, 1995). Due to these contradictory results, it remains uncertain if there exists a direct relationship between impact forces at touch-down and rearfoot motion as indicated by a frontal plane model of the foot (Denoth, 1986). To give a more comprehensive explanation, this study was designed to investigate whether a relationship between impact forces at touch-down and rearfoot motion exists within subjects and between steps taken. It was assumed that variations in sagittal plane kinematics are neglegible.

Methods
Ten subjects, all recreational runners participated in this study. Their mean age, height and weight were 28.5 ± 5.3 years, 178.8 ± 1.9 cm and 75.8 ± 5.8 kg respectively. All met the criteria of being rearfoot strikers, injury free and experienced in treadmill running. Standard Asics Gel 125 ES running shoes were used for the testing sessions. During the experiment subjects were required to run on a motor driven treadmill (Gaitstar; BFTS, Cologne) at a velocity of 3.5 ms⁻¹. When a stable running pattern was attained, data collection began and continued for a period of 15 s. A strain gauge based forceplate built into the treadmill was used to collect ground reaction force data, while an in-shoe goniometer recorded rearfoot motion. In addition a hydrocell force sensor (2 cm in diameter; Paromed, Markt Neubeuern) was positioned inside the shoe under the heel and an uniaxial accelerometer (Biovision, Wehrheim) was fixed with adhesive and a rubber band on the medial aspect of the tibia. All data were A/D converted and recorded simultaneously with a sampling frequency of 1000 Hz. Prior to calculating the parameters the raw data were

Table 1: Determined parameters out of 21 ± 1 steps.

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<tr>
<th>Parameter</th>
<th>Code</th>
<th>Unit</th>
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<tr>
<td>External impact peak</td>
<td>F_p</td>
<td>N</td>
</tr>
<tr>
<td>External force rate</td>
<td>dF_p</td>
<td>Ns⁻¹</td>
</tr>
<tr>
<td>Internal impact peak</td>
<td>F_ip</td>
<td>Ncm⁻²</td>
</tr>
<tr>
<td>Internal force rate</td>
<td>dF_ip</td>
<td>Ns⁻¹</td>
</tr>
<tr>
<td>Maximum acceleration</td>
<td>a_max</td>
<td>g</td>
</tr>
<tr>
<td>Initial rearfoot angle</td>
<td>β_TD/β_pr</td>
<td>°</td>
</tr>
<tr>
<td>Minimum rearfoot angle</td>
<td>β_min</td>
<td>°</td>
</tr>
<tr>
<td>Maximum velocity of pronation</td>
<td>dβ</td>
<td>°s⁻¹</td>
</tr>
<tr>
<td>Range of pronation</td>
<td>β_range</td>
<td>°</td>
</tr>
</tbody>
</table>

Figure 1: Simultaneous recording of rearfoot angle, tibial acceleration, force/pressure under the heel and ground reaction force for one single step; arrows indicate the identified maxima/minima.
filtered using a FFT procedure with cut-off frequencies determined by the spectrum of each signal (18 - 35 Hz).

Each subject performed one trial of which 21 ± 1 steps were selected for further analysis. Pearson's correlation coefficients were calculated to check for possible dependencies between the extracted parameters.

**Results and Discussion**

When the time course of rearfoot motion with respect to first ground contact was investigated, 50 % of the subjects were found to start pronation from a slightly supinated position 10 to 40 ms prior to touch-down. The remaining subjects started the pronatory movement at the instance of first ground contact. Since the former subjects have a $\beta_{\text{min}}$ prior to touch-down their range of pronation was calculated as the difference between $\beta_{\text{pre}}$ and $\beta_{\text{min}}$; for the latter subjects the difference between $\beta_{\text{TD}}$ and $\beta_{\text{min}}$ represented the range of movement.

The results indicate an inconsistent pattern between selected parameters among subjects. In figure 2 an example of two subjects showing contradictory relations between the range of pronation ($\beta_{\text{range}}$) and the magnitude of the external impact peak ($F_p$) is presented. To summarise the individual findings for the whole sample a mean correlation coefficient was calculated ($C_{\text{mean}} = 1/10 \sum C_i$). The results of this analysis are given in table 2.

![Graph showing regression analyses of range of pronation and external impact peak for two different subjects.](image)

<table>
<thead>
<tr>
<th>Parameters</th>
<th>$C_{\text{mean}}$</th>
<th>SD</th>
<th>CoV</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_p - F_{ip}$</td>
<td>0.57</td>
<td>0.16</td>
<td>27.94</td>
</tr>
<tr>
<td>$F_p - a_{\text{max}}$</td>
<td>0.57</td>
<td>0.26</td>
<td>44.96</td>
</tr>
<tr>
<td>$F_p$</td>
<td>0.30</td>
<td>0.24</td>
<td>79.53</td>
</tr>
<tr>
<td>$\beta_{TD}/\beta_{\text{pre}}$</td>
<td>0.23</td>
<td>0.40</td>
<td>176.7</td>
</tr>
<tr>
<td>$dF_p - \beta$</td>
<td>0.72</td>
<td>0.13</td>
<td>18.48</td>
</tr>
<tr>
<td>$dF_p - a_{\text{max}}$</td>
<td>0.19</td>
<td>0.34</td>
<td>184.0</td>
</tr>
<tr>
<td>$F_p - \beta_{\text{range}}$</td>
<td>0.19</td>
<td>0.40</td>
<td>214.8</td>
</tr>
<tr>
<td>$dF_{p} - \beta_{\text{range}}$</td>
<td>0.10</td>
<td>0.38</td>
<td>372.3</td>
</tr>
</tbody>
</table>

Looking at the kinetic variables some reasonable dependencies become obvious. The external impact demonstrates moderate but significant correlations to maximum acceleration ($a_{\text{max}}$) and internal Impact ($F_{ip}$). The external force rate ($dF_p$) and $a_{\text{max}}$ demonstrate the strongest and most consistent correlation with $r = 0.72 \pm 0.13$. In an interindividual comparison by Hennig et al. (1993) this relation was also
demonstrated. Comparing kinetic and kinematic parameters no considerable relation can be observed. In particular maximum pronation, pronation velocity and total range of pronation display almost no significant correlation with the impact peak or the force rate. In addition no relationship with maximum tibial acceleration could be identified. It cannot be concluded that step-to-step variations of rearfoot motion directly influence impact forces during running. Further individual changes might be responsible for the observed inconsistencies.

References
THE MEASUREMENT AND EVALUATION OF THE CUSHIONING ABILITIES OF RUNNING SHOES
Sadayuki Ujihashi
Dept. of Mechanical and Environmental Informatics, Tokyo Institute of Technology;

Running shoe sole has an important role to absorb the external impact forces that transmitted from the running surface. Especially, the impact force occurred on the runner’s heel is about 3 times of the runner’s weight. Therefore, the purpose of in this paper is to show the method of measurement and evaluation for the mechanical characteristics of shoe sole. The impact device used in the trials, is made by considering the collision conditions between human heel and the running surface. It means the maximum impact force is about 2kN, the impact velocity is 1m/s and the duration time of load is about 20 to 40 ms. To satisfy all the conditions, we proposed a drop-weight type testing system with an accurate measuring devices. A variety of commercial running shoes were chosen randomly for the trials, to check up the validity of the testing system. From the measurements, the mechanical properties of each shoe, including the maximum impact force, maximum deformation, rate of energy absorption, average Young's modulus, etc., are extracted to evaluate the individual cushioning characteristics. At the same time, the human sensory evaluation of each shoe was done to compare with the mechanical measurements. As the result, by observing the similar tendencies of these two evaluations that we have obtained, it is obviously that the mechanical measurements have definite correlations with human senses.
THE INFLUENCE OF MEDIAL AND LATERAL ORTHOTIC WEDGES ON LOADING OF THE PLANTAR APONEUROSIS. IN VITRO STUDY.
G.F. Kogler¹, F.B. Veer¹, S.E. Solomonidis², J.P. Paul²
¹Orthopaedic Bioengineering Research Laboratory, Southern Illinois University School of Medicine, Springfield, IL, USA
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THE EFFECTS OF VISCOELASTIC INSOLES ON GAIT KINETICS
Paul Fiolkowski, Jeff Bauer
Department of Exercise and Sport Science, University of Florida

Introduction
Gait has been described as a series of collisions between the foot and the ground. A review of current footwear marketed shows that it has been a common aspect of footwear manufacturers to try to increase the cushioning of their products.

One method of increasing cushioning that is available is the use of commercially available viscoelastic insoles. These have previously been noted as decreasing accelerations in gait. There has been research to suggest clinical benefits associated with these insoles. Previous research has shown that modifying footwear can decrease the skeletal transients that result from gait. (1) Some of the research that has studied these insoles has examined accelerations, while others have studied vertical forces and kinematics (2). There have been reports that have dealt with cushioning and perception, and the results have shown that the perception does not always coincide with the measured results. (3) Therefore, it was the intent of this project to measure the forces experienced at the foot-shoe interface and how the addition of a viscoelastic insole would alter these forces. It was also a goal of this project to measure the effects, if any, on the timing of the phases of gait.

Methods
For this experiment, 8 subjects were recruited from the student population at the University of Florida. All subjects were free of any symptomatic pathology of the lower extremity and had been injury free for at least the past 6 months. After being informed of the data collection procedures and signing an informed consent form, the subjects were fitted with identical shoes in this experiment. The subjects were also fitted with a pair of Parotec pressure sensitive insoles to collect the data. The technical operating characteristics for the Parotec insole system used in this experiment is given in Table 1.

After a brief familiarization period, the subjects walked in a straight line at a self-selected ‘normal’ pace. The system collected data from the first 5 footstrikes on each foot, for a total of 10 footstrikes. The data was then downloaded into a computer for analysis. The trial was repeated with the viscoelastic insoles, the presentation of which was randomized, and the subjects were blind as to the insole condition. Three trials were collected in each condition.

Results
The viscoelastic insoles had no effect on the impulses in the rearfoot region. The control condition demonstrated a value of 29.9Ns while the viscoelastic insole condition showed a value of 26.9Ns. Further, there was no difference in the time to peak pressure in the rearfoot.

The forefoot measures showed a difference between the conditions. As shown in Figure 1, there was a significantly lower impulse in the control condition ( p<0.05 ), while the use of the insoles resulted in a significantly lower time to peak force in the forefoot ( p<0.01). There was no difference in the ground contact time between the 2 conditions.

Discussion
A considerable amount of footwear research seems to examine ways of decreasing the forces encountered as a result of foot impacts with the ground. While viscoelastic insoles are commonly used for this purpose, and their mechanical properties would suggest that they
would be effective for this purpose, the results here indicate that these insoles are not having the intended effects on force attenuation. The impulse in heel strike did not significantly change, nor did the time to peak force. This may be an indication of a decrease in afferent nerve activity, as has been proposed by Robbins (2), which may lead to an increase in the impact force. While the rearfoot values were unchanged, there was a change in the rate at which the forefoot region was loaded, as well as in the impulse in the forefoot. With the use of the insoles, the forefoot was loaded sooner and to a higher degree. It is not clear what effects these changes may have on injuries, or the prevention of injuries. If excessive rearfoot loading or motion is a possible cause, then this study suggests that the rearfoot is unloaded faster with the use of the insoles. The reasons for this change are not clear, and more research is needed to determine the causes for this change in gait.

Table 1: Technical measuring characteristics of Parotec insole system

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Measurement range</td>
<td>62.5 N/cm²</td>
</tr>
<tr>
<td>Resolution</td>
<td>0.25 N/cm²</td>
</tr>
<tr>
<td>Classification of accuracy</td>
<td>+/- 2.0% of end value</td>
</tr>
<tr>
<td>Non-linearity</td>
<td>+/- 0.42% of end value</td>
</tr>
<tr>
<td>Hysteresis</td>
<td>0.01 N/cm² at 2 (N/cm²)/h</td>
</tr>
<tr>
<td>Temperature coefficient</td>
<td>+/-0.015 (N/cm²)/K</td>
</tr>
<tr>
<td>Air humidity coefficient</td>
<td>+/-0.001 (N/cm²)/%</td>
</tr>
</tbody>
</table>


NATURAL PLANTAR TOPOLOGY AND POSTURE: 
A NEW STANDARD FOR ANATOMICAL FOOTWEAR RESEARCH 
Randall Barna 
Footform Laboratories

Footwear insole anatomy is integral to comfort and biomechanical solutions. Yet, there are few valid studies on the subject. Unfortunately, most footwear research is focused on materials, or on the body's interaction with specified materials and structures. Biomechanical knowledge and diagnostics are advanced science. Applying therapy with in-shoe anatomy or orthoses therefore, should also be a science but is considered an "art" in today's market. Lack of standards and inconsistencies in the methods for plantar modeling have prevented progress in this field. McPoil and Hunt have recently compiled results that challenge the most common theory for evaluation and treatment of the foot, and recommend, as we do, reevaluating the present approach. (1)

If, comfort and biomechanical solutions are achieved with footwear anatomy, then a standard for plantar topology evaluation exists that is logical and can determine what is "normal" to a foot. Such a standard exists, just as time existed before clocks. We simply need to view the dynamic topology of a foot during gait in a static model. Natural Plantar Topology and Posture (NPTP) is the plantar aspect as it fully functions in its natural and normal manner - without obstruction. A model of Natural Plantar Topology and a reference of its posture to level is a standard for each foot. For research and design purposes, the interspace between NPTP and level is clearly defined. In standardized footwear, the "normal" interspace will be entirely occupied by footwear or orthotic devices integrated to footwear. The study concept is quite simple. Either the anatomy in footwear matches NPTP or it does not. Any alterations to NPTP can be measured, applied to footwear, and studied for their therapeutic effects.

Application of a plantar topology standard began with criteria for a modeling device that could portray NPTP. First it must be dynamic - mimicking the body during gait as closely as possible. Second, it must be consistent - every model is created the same - interlocation and intertechnician. Third, it must be reproducible - the same result every time. Fourth, it must be indiscriminate of foot size, shape, flexibility, bodyweight, etc. Fifth, it must naturally occur without touching or positioning. Sixth, it must be non-obstructive so that flexible foot topology will not be compensated on a non-yielding surface. Finally, it must be reference measurable in easily achieved units or standards of measure.

Our device invented for Dynamic Plantar Modeling meets the above criteria and we developed a method to replicate the interspace between NPTP and level ground. The entire process is called "The Topological Approach." The foot and an impressionable thermoplastic material are placed on sand particles within a container. A subject stands on their foot in natural single limb stance. Friction between the granular particles holds the body stable. Dynamic gait-simulated motion is generated to the foot from the sand below by displacing volume, breaking friction and internally rotating the container. The impressionable material captures the topology of the foot as it progressively settles from heel to toe in a new supported position without obstruction. The Topological model is measured for its posture relative to level before it is removed from the container. To generate the in-shoe anatomy (orthoses) with a surface that replicates NPTP we use a special process utilizing visco-elastic polymer. Gravity levels the material to the desired reference position and fills the interspace between NPTP and level. This method and material also allow pre-existing footwear anatomy to be integrated to the orthoses and alterations to be performed and measured. Technology is now available to automate the fabrication of the shoe insert and use other materials.
We discovered that the NPTP models contained valuable information. Longitudinal arches, posture to level, and forefoot/rearfoot relationships revealed foot function and the pathology of biomechanical stress. Initial testers of footwear with NPTP were non-symptomatic. Their response indicated a natural feel, with benefits of comfort and reduced fatigue. Objective in-shoe pressure analysis of one subject recorded that dynamic pressure patterns were equal with or without NPTP orthoses in footwear. This confirmed that our invention was producing the NPTP natural standard and we began applying alterations to symptomatic cases.

We have developed protocol for alterations, increments of measure, and a database format for records and continue to conduct subjective efficacy studies. A study was done in a clinical setting on 793 randomly selected subjects with a variety of symptoms. Response was voluntary to a mail survey designed to assure they were using the orthoses correctly and to record their outcome and perceived benefits. Response came from 52% of the recipients. A second study was conducted among 50 non-responders to the initial survey. This study was the same as the first, and also voluntary. Response from the second survey was 50%. The results of survey one; 93% claimed to be better when they wear their orthoses. In survey two, the non-responders, 94% claimed to be better. In these surveys all positive responders listed additional benefits with the most common being Comfort (79% in study one) and (72% in study two). A third study was conducted among the 793 subject group that had zero alterations to NPTP in their orthoses. This group comprised 39% of the responders. 94% of this group claimed to be better with 80% citing comfort as an additional benefit. We were pleased to find that 39% of symptomatic cases needed no alterations to NPTP and had equally positive results. We also found that a small alteration (.05") was markedly noticeable. The studies show that The Topological Approach works.

The results of The Topological Approach study clearly show the potential and need to apply well-designed anatomy to footwear is enormous. There is a large variance between in-shoe anatomy of stock footwear and NPTP - justifying the application. Also, the testing of footwear structures and materials will be more objective if the tester's feet are standardized with NPTP orthoses. For medical research, valid objective studies using in-shoe pressure analysis, computerized gait analysis, and The Topological Approach are achievable. Apply alterations to NPTP and objectively test for their therapeutic effects to increase comfort, performance and prevention.

Introduction

Runners are susceptible to injuries in the lower extremities. Excessive foot pronation, excessive tibial rotation, cavus feet or limb length discrepancies have been associated with an increase of overuse injuries such as patello-femoral disorders, shin splints, Achilles tendinitis, plantar faciitis and stress fractures (1,2). Furthermore, it has been proposed that many running related injuries caused by excessive foot eversion and excessive internal tibial rotation (1,2). Orthotic shoes, specific shoe construction features, materials and/or shapes of shoe inserts are commonly assumed to be beneficial with respect to comfort and/or overuse injuries (3,4,5). Consequently, shoe inserts have often been constructed to reduce foot eversion and/or internal tibial rotation (3,4,5,6). However, a systematic approach outlining the relationship between static and dynamic foot and leg characteristics and appropriate insert characteristics is still missing. The purposes of this investigation were:

- to quantify the effect of subject specific anthropometric characteristics on foot eversion and tibial rotation,
- to quantify the effect of systematic changes in material composition of one specific shoe insert, specifically constructed to reduce foot eversion and tibial rotation, on foot eversion and tibial rotation during running and
- to quantify the possible influence of individual foot characteristics on changes in foot eversion and tibial rotation due to the used inserts.

Methods

Twelve physically active male subjects were recruited for the study. The subjects ran on a 30 m indoor runway at a running speed of 4.0 ± 0.2 m/s and the movement was quantified for one foot fall in the middle of the runway. All running trials used one standard running shoe. Six conditions were analyzed in this study, one condition in which the movement was performed with the test shoe without any insert and five conditions with specific inserts with different material properties. The inserts had a bilayer design using two different material hardnesses at the top and bottom of the inserts. Seven running trials were collected for each condition.

The spatial positions of the lower leg, foot and shoe were defined using six spherical reflective markers attached to the subjects leg and shoe and two flat round reflective markers attached to the subjects foot with adhesive tape. A hole was cut in the back of the left shoe to view the two flat circular markers. The three-dimensional positions of all markers while running were recorded with four high speed video cameras at 200 Hz. The functional kinematic variables examined in this study were the foot-leg in-eversion angle, $\beta$, and the leg-foot tibial rotation, $\rho$. Specifically, these variables were determined for the time immediately before ground contact and for the maximal amplitude.

Each subject was classified with respect to arch height, arch stiffness and range of motion of the ankle joint complex. Arch height was quantified with the subjects standing in upright posture with the ankle joint in a neutral position on a raised platform. The highest point along the soft-tissue margin of the medial plantar curvature was measured with a modified Mitutoyo digital caliper. Two measurements were taken, one in a full weight bearing postion and another in an unloaded position. Relative arch deformation, RAD, was a measure of the stiffness of the arch of
the foot. RAD was defined as the ratio between the unloaded and loaded arch heights. Active range of motion of the left foot relative to the leg was evaluated using an apparatus which allowed independent measurement of three rotational degrees of freedom: (plantarflexion-dorsalextension, in-eversion and ab-adduction) and three translational degrees of freedom as described earlier.

The statistical analysis selected was a two way repeated measures MANOVA (within trials and inserts).

Results and Discussion

The average group changes due to the studied inserts in total shoe eversion, total foot eversion and total internal tibial rotation were typically smaller than one degree if compared to the no insert condition and the differences were statistically not different. The measured ranges of total foot eversion for all subjects were smallest for the softest and about twice as large for the hardest insert construction. Thus, the soft insert construction was more restrictive, forcing all feet into a similar movement pattern while the harder combinations allowed for more variation of foot and leg movement and did not force the foot into a pre-set movement pattern.

The individual results (Table 1) showed substantial differences between subjects and a trend: subjects which showed generally a reduction of tibial rotation with all tested inserts had typically a flexible foot. However, subjects which showed generally an increase of tibial rotation had typically stiff feet.

The results of this study suggest that subject specific factors, such as static, dynamic and neuro-physiological characteristics of foot and leg, which are not fully understood today are important to match specific feet and shoe inserts optimally.

Table 1 Summary of effects of inserts on total foot eversion and total tibial rotation for the individual subjects. The numbers in the table represent the different subjects.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Inserts Decreased All Results</th>
<th>Inserts Increased All Results</th>
<th>Inserts Increased and Decreased Results</th>
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</thead>
<tbody>
<tr>
<td>$\Delta \beta_{\text{foot}}$</td>
<td>2,4</td>
<td></td>
<td>1,3,5,6,7,8,9,10,11,12</td>
</tr>
<tr>
<td>$\Delta \rho$</td>
<td>1,5,7,9,11,12</td>
<td>4,6,10</td>
<td>2,3,8</td>
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References

Introduction
The location of the gait line (centre of pressure line) has often been used to quantify abnormal foot function, as well as to assess the influence of orthoses or shoes on the dynamic function of the foot\(^1\). Information about the application of forces acting on the plantar surface of the foot can thus be obtained. However, measurement of the velocity of the gait line should provide additional information about the function of the foot during forward motion. This study was carried out in order to investigate the relationship between the walking velocity and the average velocity of the gait line, as well as to investigate the effect of different foot types, orthoses and foot wear on the dynamic velocity profile of the gait line.

Methodology
The subjects walked on a treadmill at controlled speeds of between 1.5 and 5.0km/h. The plantar pressure distribution was measured inside the shoe at 0.5km/h intervals using the pedar mobile in-shoe system. Each subject was measured whilst wearing their own selected ‘fashion’ shoes and trainers. Some subjects were also measured with and without orthoses. Commercially available software (novel-win) was used to calculate the velocity of the gait line. The instantaneous velocity of the gait line for each frame recorded during the dynamic roll-over (measured at 50Hz) was calculated and plotted against time. A dynamic velocity profile of the gait line was thus obtained, a typical curve of which is shown in Figure 1. A range of barefoot pressure distributions were also analysed.

Results and Discussion
The average velocity calculated from the velocity profile showed the most consistent trends and had the strongest correlation with the walking speed. A linear correlation coefficient of greater
or equal to 0.98 existed for all conditions, although it was thought that factors such as the shoe type, the foot length, the cadence, the step length, the foot structure as well as the gait line length affected the exact nature of the relationship.

The dynamic velocity profile varied between subjects, but for each subject was similar for all walking speeds, although the velocity peaks became more defined with increased walking speed. This indicated that the general form of the velocity profile was not dependent on the walking speed. Small variations in the relative timing of the velocity peaks were observed in the profiles recorded for each subject wearing different orthoses and shoe types, but again, the overall form of the velocity curves for each subject remained very similar, thus suggesting that the overall velocity profile was not sensitive to the type of shoe or orthotic. However, the form of the velocity profile could vary considerably for different subjects.

Barefoot dynamic velocity profiles for a range of feet were thus analysed further in order to assess the relationship between the velocity profile and the dynamic function of the foot. The velocity profile calculated from barefoot measurements taken from each individual were found to be similar to those calculated from in-shoe measurements, but the profile could vary considerably between individuals.

A general form of the velocity profile was identified for ‘normal’ subjects which is illustrated in Figure 1, although small variations occurred in the relative heights of the peaks and their exact timings for different individuals. This general form could also be identified for children’s feet (2-8 years). However, this form was very different, for example, from that which was typical for a diabetic foot, suggesting, therefore, that the function of the foot in determining forward motion is significantly altered in the diabetic foot. This could have implications for the quantitative assessment of various foot disorders.

**Conclusion**

The dynamic velocity profile appears to be insensitive to the walking speed, the shoe type, and barefoot compared to in-shoe walking, but does appear to be dependent on the individual function of the foot. The velocity of the gait line could therefore be a useful quantitative parameter in the assessment of dynamic foot function, either barefoot or in-shoe.

**References**

A COMPARISON OF TWO FRICTION MEASURING METHODS
Craig Wojcieszak, Peixing Jiang, & E. C. Frederick
Exeter Research Inc. and Converse Inc.

Introduction

Although force plates are considered the “gold standard” for measuring frictional forces between shoe and surface, they are not usually feasible for in situ testing, or testing of surfaces such as sod or loose soil. We developed a manually actuated, portable friction tester, similar to a device described by Valiant (1), to test surfaces we could not bring into the laboratory. We compared the kinetic friction coefficients collected simultaneously with the portable friction tester and a force plate to observe any differences between these methods.

Methods

Prior to testing, we cleaned the outsoles of eight basketball shoes (identical model and size) with alcohol, abraded them with 100 grit sandpaper, cleaned them again with alcohol, and allowed them to dry. Via the portable friction tester (S2T2; Exeter Research, Inc.), we subjected each shoe to a static vertical load of 696 N as it was manually pulled across a polyurethane-varnished maple floor surface bolted to an AMTI force plate.

System analysis revealed the difference between the shear force acting on the force plate (F_{FP}, which is also equal to friction force) and the portable friction tester (F_{DFI}), and therefore the difference between the friction coefficients \( \mu_{FP} \) and \( \mu_{DFI} \), was directly related to acceleration (2). Newton’s second law (see Equation 1) defines these relationships, and dictated we pull each shoe with minimal acceleration in order to minimize frictional differences. We accomplished this by pulling each shoe 120 mm across the floor sample at an aggregate mean velocity of 7 mm/s (+/- 1 mm/s).

We simultaneously collected force data at 200 Hz for all three force-components from the force plate and the force output of the portable friction tester. Kinetic friction coefficients were calculated (see equations 2 & 3 below) for each shoe from the data subset in which the shear force remained relatively constant for the longest period of time (at least one second). The average difference between the maximum and minimum shear force within these data subsets was 14 N, +/- 5 N. Force plate friction coefficients were calculated using the measured shear (F_X & F_Y) and normal (F_Z) force components. Portable friction tester coefficients were calculated using F_{DFI} and a static vertical load of 696 N.

\[
\text{Equation 1: } \sum F = ma, \text{ specifically... } F_{DFI} - F_{FP} = m_{\text{shoe}} \cdot a_{\text{shoe}}
\]
\[
\text{Equation 2: } \mu_{FP} = \frac{\left( F_X^2 + F_Y^2 \right)^{1/2}}{F_Z}
\]
\[
\text{Equation 3: } \mu_{DFI} = \frac{F_{DFI}}{696 \text{ N}}
\]

Results

Force plate and portable friction tester mean \( \mu_k \) values for the eight shoes were 0.74 (+/- 0.04) and 0.76 (+/- 0.05), respectively. A paired t-test indicated the differences between these means were significant (P < .01).
Discussion & Conclusions

The results of this study indicate a significant difference between determining $\mu_k$ with the force plate and portable friction tester methods. However, looked at from a practical viewpoint the mean difference between $\mu_{FP}$ and $\mu_{DFI}$ is relatively small ($\Delta \mu_k = 0.02$). Additionally, the inter-shoe variability measured with either method (standard deviation for $\mu_{FP}$ and $\mu_{DFI} = 0.04$ and 0.05 respectively) was larger than the difference between the measuring methods. Inspection of $\mu_k$ vs. time curves also demonstrate that $\mu_{DFI}$ closely mirrors $\mu_{FP}$. The largest difference between $\mu_{FP}$ and $\mu_{DFI}$ for any single data pair was 0.04. Thus, the portable friction tester can be a useful substitute for the force plate in determining friction coefficients when the following statements are true.

- Acceleration of the shoe sample remains at or near zero.
- The researcher deems the difference between $\mu_{FP}$ and $\mu_{DFI}$ small enough to be inconsequential.

References

The purpose of developing this measuring system was the strong need to get information about horizontal movements of the foot compared to the shoe in order to get any idea about the foot in the shoe in sports with high sideward, stopping and turning movements (e.g. all indoor sports, tennis). These information help to quantify these diverse situations, help to realize unphysiological and stressing movements of the foot and thus give valuable information of possible sources of injuries. Till today it was only possible to measure vertical forces (and pressures) of the foot in the shoe which is unsatisfactory and insufficient to get complete information of the foot compared to the shoe in sportive situations. The system is based on contactless sensing elements where the receiving sensors are fixed to the foot and the transmitting sensors are fixed at the transition from the outsole to the midsole. It is possible for the sensors to measure the movements in all three dimensions (both horizontal and the vertical direction). Moreover it is possible to realize high time and place resolution.

The Hardware Solution:

- Foot Applicable Sensor (FAS) for position detection (taped to the foot, using the application tool for positioning)
- Hand Trigger Unit (HTU) to start measurement (using the button of the joystick to start measurement)
- Data Logger Module (DLM) (micro processor controlled data logger, setting start/stop window for measuring procedure, interface to adapt data format to RS 232 standard or PCMCIA)
- Output Signal Transfer (data input/output based on industrial standards, devices adaptable to different requirements, connection via RS 232, COM 1/2 to computer
- Software (SM) based on WIN 95/ or WIN NT)

Technical Data:

- Number of sensors:
  - LPSS 2000/5P: ≤ 5 sensors per shoe
  - LPSS 2000/7P: ≤ 7 sensors per shoe
- Smallest distance between two sensors: 10mm
- Sample rate per sensor: ≥ 500 Hz
- One battery set can drive the system for: ≥ 30 min
- Resolution: better 1mm
- Repeatability: better 1mm
- Measuring range (max. shift of foot in shoe): ± 15mm
- Warning for low battery: buzzer
- Dimensions which can be measured: 2 (forward/backward, lateral/medial)
- Data logger for all data: ≥ 30 min
- Data exchange to evaluation PC: via PCMCIA or RS 232
- Evaluation of data: with external PC
Skin is known to influence the measured tibial acceleration in skin mounting technique. Due to the low resonant frequency of the accelerometer attachment on the tibia, recorded impact shock values are prone to error. It has been pointed out that skin mounted accelerometers give higher peak and later time-to-peak acceleration values than bone mounted transducers. Besides, spectral analysis of the impact shock can be distorted due to the overlapping of the skin mounted accelerometer attachment response and the frequency response of the impact shock. Despite all this limitations and measurement errors, skin mounting technique is still preferred because of its easiness of using in routine measurement owing to his non-invasive character. Therefore, methods for removing or reducing skin effect have to be found in order to obtain an accurate estimation of the tibial acceleration with skin mounted transducers.

We propose a method based on the wavelet shrinkage of the skin mounted signal. The wavelet decomposition allows to represent a signal on basis of orthogonal function. Unlike the Fourier transform of the signal, each coefficient issue from the decomposition is localised in time and in frequency. The technique of wavelet shrinkage is a technique that permits to recover a true signal from indirect or noisy data by reducing or setting to zero all coefficients below a particular threshold.

The algorithm that we use for recovering the true tibial acceleration follows those three steps:
- wavelet transform of the skin mounted tibial acceleration,
- shrinkage of coefficient based on the power spectral density of the measured acceleration,
- inverse wavelet transform in order to obtain an estimate of the true tibial acceleration.

Results obtained with simulated impact peak signal and Ziegert’s model of skin effect will be presented and discussed.
RADIOGRAPHIC AND ULTRASONIC METHODS TO STUDY FOOT STRUCTURE
Erez Morag* and Peter R. Cavanagh
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INTRODUCTION
Although foot structure is a feature of critical interest to the footwear industry, it has traditionally been defined simply by gross anthropometric and/or by a subjective description of the medial longitudinal arch. The mechanical properties of the heel pad have also been studied by various methods (1,2). The purpose of the present study was to introduce (1) a protocol to study the bony structure of the foot from standardized radiographs, and (2) a new method to study soft tissue properties \textit{in vivo}. When combined, these two methods provide important information regarding the bony structure of the foot and the mechanical properties of its soft tissue.

METHODS
The radiographic method involved the collection of standardized lateral and dorsi-plantar weight bearing plain radiographs of the foot and ankle. Issues such as X-ray settings, subject posture, foot positioning, calibration, and analysis have been previously discussed (3,4,5). The films were scanned with a video camera and captured with a frame grabber card, using Radius Videoviewer software onto a Macintosh computer. This technique allowed the measurement of bone dimensions and orientation as well as the thickness of compressed soft tissue during weight bearing, using the NIH Image program (Figure 1). The means, standard deviation, and radiographic reliability (ICC) of selected measurements are presented in Table 1.

![Figure 1. Schematic diagrams of radiographic measurements from lateral (left) and A-P (middle and right) views](image)

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Code</th>
<th>View</th>
<th>Units</th>
<th>Mean</th>
<th>s.d.</th>
<th>ICC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calcaneal inclination</td>
<td>6</td>
<td>lateral</td>
<td>degrees</td>
<td>21.3</td>
<td>6.4</td>
<td>0.97</td>
</tr>
<tr>
<td>Sesamoid height</td>
<td>1</td>
<td>lateral</td>
<td>mm</td>
<td>7.1</td>
<td>1.5</td>
<td>0.87</td>
</tr>
<tr>
<td>Intermetatarsal 1-5</td>
<td>1</td>
<td>A-P</td>
<td>degrees</td>
<td>24.3</td>
<td>3.4</td>
<td>0.92</td>
</tr>
<tr>
<td>Morton's index</td>
<td>9</td>
<td>A-P</td>
<td>mm</td>
<td>3.2</td>
<td>2.5</td>
<td>0.98</td>
</tr>
</tbody>
</table>

The ultrasonic method (Figure 2) allowed simultaneous monitoring of the force applied to the tissue and the ultrasonic images of the underlying bone during a quasi-static computer-controlled compression. A cylindrical Plexiglas rod was attached to the end of a 7.5 MHz ultrasound probe. A force transducer was mounted in series with the probe, and the Plexiglas rod was slowly advanced into the tissue (at 15 mm/s) until the applied pressure reached approximately 400 kPa. The ultrasonic images of the foot and the magnitude of the applied force were continuously recorded throughout the experiment on S-VHS video tape using a split-screen mixer (Figure 3). These data were then used to construct force-displacement...
curves (Figure 4). The day-to-day reliability of selected soft tissue properties was assessed (table 2) and data from more than 50 healthy individuals were collected (3).

**Figure 2.** A schematic diagram of the ultrasonic method to study soft tissue properties

**Figure 3.** A sample frame showing the ultrasonic image of the calcaneal tuberosity (right) and the force level (left)

**Figure 4.** A sample force-displacement curve during heel compression. Fitted curve is presented as a solid line

**Table 2.** Selected soft tissue properties, their means (N=52) and ICCs (N=7)

<table>
<thead>
<tr>
<th>Variable</th>
<th>Units</th>
<th>Mean</th>
<th>s.d.</th>
<th>ICC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heel pad thickness (unloaded)</td>
<td>mm</td>
<td>14.3</td>
<td>1.90</td>
<td>0.70</td>
</tr>
<tr>
<td>Displacement at 50N (relative to thickness at 5N)</td>
<td>mm</td>
<td>3.97</td>
<td>0.93</td>
<td>0.79</td>
</tr>
<tr>
<td>Displacement at 105N (relative to thickness at 5N)</td>
<td>mm</td>
<td>4.6</td>
<td>0.90</td>
<td>0.86</td>
</tr>
</tbody>
</table>

**APPLICATIONS**

These methods were used to characterize differences in foot structure across the adult life span, and to identify structural factors that determine peak pressure under different foot regions during walking. It has been found, for example, that the heel pad becomes stiffer with age, and that radiographic measurements are associated with foot pressure under all studied foot regions.

**REFERENCES**


**ACKNOWLEDGMENTS**

The authors would like to thank David Lemmon, Ph.D., Janice Derr, Ph.D., and Doug Tubbs for their analytical and technical contribution to this study.
FOOTWEAR CONSIDERATIONS FOR THE NEUROLOGICALLY
IMPAIRED SUBJECTS
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INTRODUCTION
For the Human being to walk on the hind limbs is a gift of the nature. While the major
movements of the body are controlled by the nervous system the physical ability and pressures
of the individual also is seen as a factor. Any disorder in the nervous system could disturb
many aspects in the life form including the walking and speech pattern. There are several
neurological problems contributing to the balance and gait dysfunction. These problems
include Parkinsonism, Diabetic Neuropathy, Joint sense impairment and other related
etiologies. While medical treatment is rising to the occasion, it is also equally important on the
part of footwear/accessories manufacturers to come up with ideas and designs to assist the
differently abled people.

METHODS
A subjective study as a part of another kinetic and kinematic movement analysis were
conducted on three selected subjects, who were suffering from acute impairment, with these
conditions. The subjects had an abnormal gait and while they were walking they had a
tendency to trip, shuffle, freeze and fall. An attempt was made to facilitate these subjects with
proper footwear support to overcome these problems. Two types of footwear with different
material combinations were fabricated. One was the normal, casual shoes and the other was a
mule. The soles were made of leather, micro cellular rubber (MCR) and ethyl vinyl acetate
rubber (EVA). The mule shoes sported a single vamp piece that is lasted till the waist. A
synthetic, then a grain leather, and finally a flesh side leather in-sock was used.

RESULTS AND DISCUSSION
The casual shoe was a classic slip-on with no elastic or components. The shoe had a single
vamp with a back seam. The top line was folded and stitched along with the lining. This was
later modified and the top-line was stitched with a French binding. This gave a better look to
the wearer when a contrast colour was used for the binding piece. The leather used for the
upper was soft cow upper leather and lining was soft goat lining leather. Despite the best efforts
to make the shoes comfortable for the subjects, they experienced some pain and swelling in
their legs after continuous wear. This could be due to two reasons; one is that the shoe was
designed in such a way that it encloses the feet from all the sides. This was acting some force
on the feet from the sides. Another reason is the pressure from the bottom component. The
soling material used in the first case was vegetable tanned sole leather. This, however, was
found to be stiff, hard and acted some force on the feet from below. The other material MCR
gave the much needed comfort for the subject. On the other hand the material was felt to be a
trifle heavy. This led to the subject dragging the feet while walking. The MCR sole was
substituted with a more lighter and cushioning material, EVA. Although usage of soft EVA
soles did bring down the force at the bottom, it was necessary to improvise on the material used
as upper. Hence, the soft cow upper leather was substituted with a softer sheep skin leather.
Another modification to the slip-on casual shoe was in the form of using a PU foam coated
with fabric for the lining. This was found to give a greater comfort since the impression of the
leather was not found on the subject's feet after prolonged wear. As for the upper material, the
cow upper leather was substituted with softer sheep skin to give a much better feel. The
modifications made with the different types of material did not yield the required results anticipated. So, a change in the basic design of the footwear was considered. The customary slip-on shoe was replaced with a mule shoe. The mule is a slip-on model without any back support at the Achilles region. The vamp area is covered fully and does not have any other component. The feet should be inserted and worn like a sandal. The upper material used was cow upper leather and the same leather was used as the top sole cover. There was some discomfort for the subject because of the slipper nature of the grain leather in the top sole. In addition to the grain leather, a half sock was attached with the flesh portion onto the top sole cover. This ensured that the flesh portion of the leather, when in contact with the subject's feet, gave the necessary hold. The feet did not slip out of the mule shoe while walking in this prototype. The soling material tested for fabricating the mule shoes were vegetable tanned leather, MCR and then EVA. While the leather sole still gave the hard and stiff feeling to the subject, the MCR sole was heavy for the open mule design. EVA sole was found to be good because of its light weight and soft, cushioning substance. The slip-on shoes were found to comfortable for two of the subjects. It could be easily worn without stretching out to tie up the laces. The shoe gave a stability to the subjects while walking. There were some drawbacks, like discomfort due to the swelling of the feet after continuous wear, that could not be eliminated by using the same materials and style. Hence, a different design was adopted and the same materials were used in different combinations. The mule shoes with flesh side half sock lining and EVA soles were found to yield the best results. Subjective stability during gait seems to improve with higher heels. However, more studies have to be conducted to substantiate the findings reported in the current work.
FRICTIONAL PROPERTIES OF THE SKIN AND BLISTER FORMATION
Wunching Chang and Ali A. Seireg
Department of Mechanical Engineering, University of Wisconsin-Madison, USA

Introduction

Based on available empirical data, the authors developed an equation to describe the frictional properties of the skin. The coefficient of friction is considered to be a function of the ambient temperature and the time of thermal exposure. The effect of sweating on friction is also considered. The allowable number of rubs before the formation of blisters is consequently developed as a function of the localized pressure and the coefficient of friction between the skin and the contact surface. The developed equations can be combined with the pressure predicted from a musculoskeletal foot model for guiding footwear design in order to minimize any adverse effects due to repetitive rubbing.

Coefficient of Friction of the Skin

Modeled on the study of Naylor (1), the effect of sweating on the skin's coefficient of friction, µ, can be represented by the following equation:

\[ \mu = 1.1 - 0.6e^{-0.0033t \Delta T} \] (1)

where t (Min.) is time and \( \Delta T \) (°C) is the ambient temperature increase from the comfort temperature level of 20 °C. It can be seen that under room temperature (20 °C) without sweating, the skin's coefficient of friction, \( \mu \), is 0.5 and it does not change with time. When sweating occurs due to temperature increase, the coefficient of friction of the skin, \( \mu \), increases with time and finally reaches a constant value of 1.1. The above equation can be used to predict the shear stress applied on the skin, \( \tau \). Figure 1 shows estimated values of \( \mu \) for different ambient temperature increase above the comfort temperature level. The temperature increase is expressed in 5°C increments.

Criterion for the Formation of Ulcers from Friction

The allowable number of rubs, \( N_{all} \), under an average frictional force, \( \tau_f \) (KPa), was derived from Naylor's study (2) as

\[ N_{all} = \left( \frac{243.945}{\tau_f} \right)^{2.564} \] (2)

Because the frictional force, \( \tau_f \), is caused by pressure, it is written as

\[ \tau_f = \mu P \] (3)

where \( \mu \) is the coefficient of friction of the surface in contact with the skin. Ulcers can be avoided if the number of applied rubs is less than the allowable number of rubs corresponding to \( \tau_f \). Figure 2 shows the criteria for ulcer formation with different frictional conditions. The coefficient of friction, \( \mu \), is shown in 0.1 increments from 0.5 to 1.1. It can be seen that for large number of repetitive rubbing a small reduction in pressure or \( \mu \) can significantly increase the number of rubs for ulcer formation.

Pressure on the Foot

Pressure on the foot can be predicted by the foot model developed by Seireg (3). In this model, there are thirty three segments combining the pelvis and both legs and feet. They are assumed to be rigid bodies with six equilibrium equations each. Additional equations also
account for the equilibrium between the muscle forces that wrap around a segment and the consequent reaction force at that point. Forces in the muscles and ligaments as well as joint reaction are also considered. In all, there are 344 equations for 1008 unknown variables. Because there are less equations than unknown variables, a merit function is used which minimizes physiological work and ligament forces. This model was used to predict pressure distribution under foot normal and pathological conditions.

Discussion

This study proposes a general method for the predictions of friction induced ulcers on the foot. The preliminary analysis presented in this paper suggests that the increase in friction due to sweating can significantly reduce the number of rubs for ulcer formation when the foot is subjected to many cycles of load. In order to predict the skin's response to friction more accurately, more empirical studies are needed for more accurate evaluation of the frictional properties of the skin in contact with different materials and under different thermal conditions.

References


![Effects of Sweating on the Skin's Coefficient of Friction](image1)

![Allowable Number of Rubs for Different Surfaces](image2)

Figure 1 Effects of Sweating on the Skin's Coefficient of Friction.
Introduction
Fatigue fractures in the bones of the lower extremities are prevalent among runners and military trainees (1,2). It has been hypothesized that using custom made shoe insoles might contribute to lowering bone strains and therefore lower fatigue fracture risk. In a previous prospective study on the effect of a shock-absorbing non-custom shoe orthotic on the incidence of fatigue fractures among military recruits, a decrease in the incidence of femoral fractures was found only among recruits with high arch feet, while the overall incidence of fractures was not affected (3).

Material and Methods
In the present prospective randomized clinical trial, 404 infantry recruits were assigned randomly to groups given biomechanical custom made orthotics made of either soft polyurethane (Eshed Advanced Orthopedics, Bnei-Brak, Israel) or semirigid polypropylene (ProLab Orthotics, San Francisco, CA), and a control group with either a prefabricated flat shoe insole or no orthotic. During 14 weeks of basic infantry training the recruits were reviewed every two weeks in the field. Those suspected for fatigue fractures were evaluated with Tc99 MDP bone scintigraphy and graded on a 0-4 scintigraphic scale (4). Recruits who found their orthotics uncomfortable after the 1st two weeks period were allowed to discontinue their use and considered dropouts. Orthotic comfort was evaluated at the end of the study, from 1=not satisfactory to 4=excellent. Data was analyzed by: 1) Univariate analysis: affected/non-affected with fatigue fractures by type of orthotic using the chi-square test. 2) Multivariate analysis of several factors related to fatigue fracture risk, by conditional odds ratios.

Results
Of the 404 recruits who participated in the study, 196 (49%) completed it. Among those, the overall incidence of fatigue fractures was 17.9% for the whole group, 16.0% for recruits who trained with semirigid orthotics, 10.7% for recruits who trained with soft orthotics and 26.8% for the control group (p = 0.037). Table 1 shows the breakdown of fractures by anatomical site and orthotic type. Forty-four percent of the dropouts were dissatisfied with the orthotics. Among the users of orthotics, the largest incidence of dissatisfaction was among those using the semirigid (37%) and the lowest among those using the soft (9%). Table 2 shows the mean comfort score given to the various orthotics.

Conclusions
In this study, a successful attempt was made to significantly lower the incidence of fatigue fractures among infantry recruits. In the Army setting (short duration of training, better short run tolerance), the soft biomechanical orthotic would seem to be preferable to the semirigid biomechanical orthotic. In a civilian setting when there is more time to accommodate to the device and less intense physical training, or when long term durability of the orthotic is required, the semirigid device may be preferable. We conclude that among trainees such as infantry recruits or sports people at high risk for fatigue fractures, prophylactic use of a...
semirigid polypropylene or a soft polyurethane shoe orthotic with neutral hindfoot posts fabricated from neutral subtalar position casts would seem to be warranted.

Table 1: Incidence of Stress Fractures according to Group and Anatomical Site

<table>
<thead>
<tr>
<th>Stress Fracture Type</th>
<th>Biomechanical Orthotics</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td>n=50</td>
<td>n=75</td>
<td>n=53</td>
</tr>
<tr>
<td>All Types of S.F.</td>
<td>8 (16.0%)</td>
<td>8* (10.7%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>13 (24.5%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6 (33.3%)</td>
</tr>
<tr>
<td>Femoral S.F.</td>
<td>5 (10.0%)</td>
<td>4** (5.3%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>6 (11.3%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>4 (22.2%)</td>
</tr>
<tr>
<td>Tibial S.F.</td>
<td>7 (14.0%)</td>
<td>6*** (8.0%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>12 (22.6%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2 (11.1%)</td>
</tr>
<tr>
<td>Metatarsal S.F.</td>
<td>0 (0.0%)</td>
<td>0 (0.0%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1 (1.89%)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0 (0.0%)</td>
</tr>
</tbody>
</table>

* Chi square Group 2 vs. Group 3, p = 0.037 ; Group 2 vs. Group 4, p = 0.015.
** Fisher exact test Group 2 vs. Group 4, p = 0.043.
*** Chi square Group 2 vs. Group 3, p = 0.019.

Table 2: Orthotic Comfort Scores of Recruits according to Assigned Orthotic Group

<table>
<thead>
<tr>
<th>Orthotic Type</th>
<th># of Soldiers</th>
<th>Mean</th>
<th>S.D.</th>
<th>Duncan’s Group *</th>
</tr>
</thead>
<tbody>
<tr>
<td>Semirigid</td>
<td>98</td>
<td>2.24</td>
<td>1.35</td>
<td>B</td>
</tr>
<tr>
<td>Soft</td>
<td>102</td>
<td>2.74</td>
<td>1.23</td>
<td>A</td>
</tr>
<tr>
<td>Shoe Insoles</td>
<td>100</td>
<td>1.66</td>
<td>0.93</td>
<td>C</td>
</tr>
</tbody>
</table>

* Duncan’s Multiple Range Test

References
SPORTS SHOES AND INJURY
WISH AND REALITY OF PREVENTING INJURIES THROUGH SPORT-SHOES
Stefan Grau
Institut fuer Sportwissenschaft, University of Tuebingen, Germany.

The main function that is expected of athletic footwear is injury prevention and protection of
the athlete. Since the beginning, the focus of research has mainly been in two areas: over-
pronation control and cushioning. Later new fields of research were pursued, like torsion,
pressure distribution etc. It has been assumed that different parameters within these areas lead
to possible overloading and thus to injuries of the athlete. The results of this research leads to
completely different solutions of the "problems" with sport-shoes. Nevertheless the amount of
injuries has not decreased. Moreover, it seems that many injuries occur because of those "injury
preventing solutions" that have been developed over the years.

The purpose of the Ph.D thesis has been firstly the evaluation of the relevant research
literature of the last 20 years in order to find out the different research areas and parameters
which were supposed to have a connection to injuries. For this purpose almost 400 articles in
nearly 60 periodicals and about 20 books were evaluated. The parameters were classified and
evaluated by a meta-analysis. It has been possible to show which parameters may have a
connection to injuries (correlate highly) and which and why others don't. In a second step
future research perspectives and directions are given.

Introduction
There is no verification/proof that
• reduction of pronation will lead to a decrease of injuries
• reduction of high impact loads will lead to a decrease of injuries
• allowance of torsional movements will lead to a decrease of injuries
• for the dissertation important that there is no proof/verification that
• (over-) pronation will lead to assumed injuries (e.g. achilles tendon injury)

Literature
Increase of injuries in the last 20 years (in relation to amount of people doing sports)
If one has a look at the injuries of the lower extremities:
-> 30 % achilles-tendon injuries
-> 25 % knee injuries (patella)

Goal of Study
To find out if there is a relation between commonly used - pronation describing parameters and the
clearly defined achilles-tendon injury (achillodynie).

Hypothesis:
There is a significant difference in the pronation describing parameters between people with
achillodynie and people with healthy feet (without achillodynie).

Methods
• 1389 cases of video analysis of different injuries and healthy people of the last 8 years
• 156 cases of clearly defined achilles - tendon injuries (achillodynie)
• 156 cases of people with healthy feet
• treadmill with constant speed 8 km/h (5 mph)
• 4 cameras (2 back, 1 side, 1 front )
• 25 frames/second
Parameters (barefoot and shod conditions):
Minimum calcaneus- floor angle $\gamma$ ( static stance)
Minimum calcaneus- floor angle $\gamma$ (dynamic stance)
Relative calcaneus- floor angle (dynamic stance - static stance)
Achilles - tendon angle $\beta$ (touch - down)
Maximum achilles - tendon angle $\beta$ (static stance)
Maximum achilles - tendon angle $\beta$ (dynamic stance)
Relative pronation (dynamic stance - static stance)
Total pronation (dynamic stance - touch down)

Results

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Test for equality of means</th>
<th>Levene’s Test for Equality of Variances</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>t</td>
<td>df</td>
</tr>
<tr>
<td>Barefoot, relative calcaneus -floor angle</td>
<td>2.069</td>
<td>310</td>
</tr>
<tr>
<td>Barefoot, calcaneus – floor angle static stance</td>
<td>-2.261</td>
<td>310</td>
</tr>
<tr>
<td>Barefoot, calcaneus – floor angle stance</td>
<td>-3.798</td>
<td>310</td>
</tr>
<tr>
<td>Barefoot, achilles tendon angle static stance</td>
<td>-2.203</td>
<td>310</td>
</tr>
<tr>
<td>Shod condition calcaneus – floor angle static stance</td>
<td>-2.684</td>
<td>307</td>
</tr>
<tr>
<td>Shod condition calcaneus – floor angle stance</td>
<td>-3.357</td>
<td>308</td>
</tr>
</tbody>
</table>

Conclusions
There are no significant differences between the two groups  (it appears to be significant, but the significance is caused by the high number of 156 cases).
Nevertheless it is obvious that there is a difference between the groups and that there is a higher risk of getting achilles-tendon injuries with greater angle $\beta$ and lower angle $\gamma$. The future is to find out the exact relationships between the two groups and and possible achilles-tendon injury describing parameters.
A 3D KINEMATIC EVALUATION OF FOOTWEAR STABILITY IN LATERAL MOVEMENTS
Bin Xia and John Robinson
Nike Sport Research Lab, Beaverton, USA

Introduction
Stability is an important characteristic of footwear, particularly as athletes perform lateral maneuvers during sporting events. Sport shoes should provide stability during the support phase of a lateral maneuver because of the large forces transmitted to the ground as players transition from lateral braking to push-off (1). Any instability during the support phase could lead to acute injury. Lateral instability has been cited as a cause for commonly occurring injuries like ankle sprains. Epidemiological studies have shown that between 38 - 45% of all sports injuries are ankle sprains (2). Although rearfoot motion associated with lateral stability has been studied extensively (3, 4), there is a paucity of information regarding forefoot kinematics. Researching forefoot motion during lateral maneuvers may help reduce the incidence of ankle injuries. The purpose of this study was to investigate forefoot function as athletes perform lateral maneuvers, leading to a 3 dimensional protocol for evaluating lateral stability of sports footwear.

Methods
Eleven basketball players wearing US men’s size 9 basketball shoes were recruited in the study. Based on their performance level, 8 of them were selected in the final data collection. A pair of prototype shoes were constructed by modifying a typical basketball shoe model, completely segmenting midsole and outsole, and bonding these components to the uppers only at the perimeter. A pair of unmodified basketball shoes were used as control. The prototype shoes were subjectively evaluated as more unstable than the control shoes. Each subject repeated 5 trials of lateral shuffle wearing each pair of the unstable prototype shoes and the control shoes. The foot and ankle structure was defined as a 4-segment model: forefoot, mid-foot, rearfoot, and shank. A marker triad was placed on each of the 4 segments. The foot marker triads were mounted on the shoes, while the shank marker triad was directly mounted on the skin. A Motion Analysis System and related software was used to record and process kinematic data, while a Kistler force plate was used to record foot-floor contact time. The relative motions between defined segments and within the lab coordinate system during foot-floor contact phase was investigated.

Results
Rearfoot inversion was significantly less in the unstable prototype shoes when compared with control (LSMean ± standard error: 24.2°±1.6° vs. 31.7°±1.4°; p=0.0011) (Figure 1). However, all the subjects evaluated the prototype shoe as more unstable. Shoe torsion (relative movement between forefoot and rearfoot along the longitudinal axis) and forefoot position in the X-Z plane of the lab coordinate system during support phase (arbitrarily defined as 20%-80% of foot contact phase) were identified as additional factors to describe lateral stability. The prototype shoes showed larger shoe torsion (LSMean ± standard error: 23.0°±1.3° vs. 6.9°±1.2°; p=0.0126) (Figure 2) between forefoot and rearfoot during support phase. The forefoot of the prototype shoes rotated more than the control shoes in X-Z plane of the lab coordinate system (LSMean ± standard error: -3.35°±3.9° vs. 3.56°±2.9°; p=0.0004) (Figure 3).
Discussion and conclusion

Rearfoot inversion alone may not adequately define lateral stability of sports footwear as shown by the results in this study. Subjects subjectively evaluated the prototype shoe as unstable compared to the control basketball shoe, however, there was less rearfoot inversion in the prototype shoe. Forefoot motion data clearly demonstrated instability of the prototype shoes.

Even though shoe torsion may not correlate with foot torsion inside the shoe, excessive shoe torsion may indicate that the prototype lacked lateral stability. It seems logical that the plantar surface of the forefoot should remain parallel with the ground during the support phase as an athlete performs a lateral maneuver. Any over-rotation beyond the horizontal plane should be considered instability. When the shoe rotates beyond the horizontal plane then either the forefoot has slid laterally beyond the midsole/outsole, or the foot itself has over-rotated. Either situation may lead to excessive inversion of the whole foot causing ankle injuries.

In summary, forefoot motion is an important parameter to consider when studying lateral stability. It is particularly pertinent in any protocol designed to evaluate the stability of sports footwear. Shoe torsion and forefoot position in the lab coordinate system in addition to rearfoot motion should all be considered when evaluating lateral stability.

References

GENDER, SIDE AND REGIONAL PRESSURE DIFFERENCES DURING SPIKING
Timothy J. Eng & Jonathan B. Fewster
NIKE Sport Research Lab, Beaverton, Oregon, USA

Introduction
Most kinetic research examining volleyball spiking has focused on measuring ground reaction forces. For footwear research, we are also interested in determining load distribution over the different regions of the foot. Although some data using discrete sensors has been reported, no data has been published which used pressure sensitive insoles to measure this distribution during spiking. To facilitate future research and product design such measures could also be used to assess differences between gender, side, and regions. Therefore, the purpose of this investigation was to assess gender (G), side (S), and regional (R) load and plantar pressure differences during volleyball spike jumps and landings.

Methodology
Twelve subjects, 6 women and 6 men of similar stature (mean heights = 1.75 m and 1.76 m, respectively) and mass (mean masses = 76.2 kg and 72.1 kg, respectively), volunteered to participate in the study. Each of the subjects were required to have greater than 4 years of competitive volleyball experience, spike proficiency, and sample-sized feet (US size 9.0 men, 10.5 women). While it was not a selection criterion, all subjects spiked with their right hand.

After the subjects warmed up, they practiced spiking a volleyball held by a spike trainer (American Athletic, Inc., USA), a device which suspended the ball by elastic cords. Following the warm up, the subjects acclimated to spiking the ball with the F-Scan (Tekscan, Inc., USA) pressure measurement wires and cuffs attached. Insoles were placed inside the shoe and plantar pressure data were collected at 165 Hz for 4 s while subjects performed 10 maximal effort spikes.

The F-Scan insoles had been calibrated at 4 pressures (180, 350, 520, and 700 kPa) using an air bladder (Novel gmbh, Germany). Individual conversion coefficients were thus derived for each sensor on each insole and were used to convert experimental data. The insole was divided into toe (T), forefoot (FF) midfoot (MF), and rearfoot (RF) regions. The toe gap was identified from a peak pressure plot as the space separating the T and FF regions. The remaining length of the insole was then subdivided into FF (0-33%), MF (34-59%), and RF (60-100%) regions.

The data were separated into jumping and landing phases. A mixed effects ANOVA model was used to assess differences between gender, side, and among regions. Initially this model included trials to detect any trends. Load was normalized by summing the pressures over a region and dividing by body weight. The maximum load was labeled the normalized peak load (NPKLD). Peak pressure (PKP) was the highest pressure within a selected region. The medial-lateral location of the PKP on the insole was also measured (NPKX).

Results and Discussion
The ANOVA model estimating trials effects showed significant differences among trials for NPKLD during the jumping and landing phases. Since PKP did not decrease over trials, we examined the maximum contact area (MAXAREA), calculated by summing the maximum number of active sensors and multiplying by the area of a sensor. The results showed that MAXAREA decreased for both jumping and landing, suggesting that sensors may be failing over trials. A Duncan Mean Separation procedure on both variables revealed that the
first 4 trials were not significantly different from each other; therefore, subsequent analyses were performed on these data.

See Table 1 for the statistically significant ($\rho < 0.05$) main effects and interactions. Note the significant gender (20% difference) and side (31% difference) effects for NPKLD during jumping. Regional effects were revealed for all dependent variables. A gender-by-side-by-region interaction for PKP appears to have been caused by differential gender responses in the toe regions between left and right sides with the female mean PKP exceeding that of the male mean. In the MF and RF regions, the female and male PKP means were similar.

### Table 1: Statistically Significant ($\rho < 0.05$) Effects and Interactions

<table>
<thead>
<tr>
<th>Jumping</th>
<th>Landing</th>
</tr>
</thead>
<tbody>
<tr>
<td>NPKLD (BW)</td>
<td>NPKX (column)</td>
</tr>
<tr>
<td>Male &gt; Female</td>
<td>Male &gt; Female</td>
</tr>
<tr>
<td>Right &gt; Left</td>
<td>Male &gt; Female</td>
</tr>
<tr>
<td>FF, RF &gt; Toe &gt; MF</td>
<td>Toe, FF &lt; RF &lt; MF</td>
</tr>
<tr>
<td>Gender<em>Side</em>Region</td>
<td>Gender*Region</td>
</tr>
<tr>
<td>Side*Region</td>
<td>Side*Region</td>
</tr>
</tbody>
</table>

The ANOVA revealed a significant gender difference for the NPKX; however, there was a side-by-region interaction. A simple effects ANOVA showed that this interaction was caused by a side effect in the MF region. The right side’s NPKX was oriented more laterally than the left side’s indicating a shift in the mean peak pressure location towards the spiking arm side, possibly caused by asymmetrical upper body kinematics or foot morphology. Landing produced a similar interaction for the NPKX.

The only non-region effect during landing was the gender effect for NPKLD (39% difference). A significant gender-by-region interaction was found for both NPKLD and PKP. A simple effects ANOVA showed a significant gender effect in the MF and RF regions for NPKLD and PKP, a significant gender effect for the RF region, and a trend for the MF region. In these cases the male means exceeded those of the females. These results may have been caused by landing from higher jump heights (mean ball height = 2.6 m for males and 2.8 m for females).

### Conclusions and Implications For Footwear Research

Results from this experiment provided valuable regional loading information and gender differences. The analyses revealed that gender, side, region and their interactions should be taken into account when conducting research on volleyball spiking. Although NPKLD was concentrated in the FF and RF, PKP was located in the T and FF, indicating a potential need for pressure distribution. The PKP location was confirmed by previous literature. The higher male loads during the takeoff and landing phases may be related to increased jump heights. The interaction between gender and region during landing revealed differences in the MF and RF regions. The spiking-side foot experienced higher forces during jumping which could have been related to a shift in the body’s center of mass towards that side. Another plausible explanation might be that since the subjects were trying to maximize the height of the spiking hand, they developed greater propulsive forces on that side. Future studies linking kinematics and kinetics should increase understanding of the laterality differences.

### Reference

FOREFOOT ABDUCTION IN VARIOUS SPORT SITUATIONS AND
ITS APPLICATION TO SPORT SHOE DESIGN

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Introduction

Freychat et al (1995) presented a new method to record the forefoot ab/adduction from static
and dynamic footprints. An abducted forefoot was called an « open foot », while an adducted
forefoot was a « closed foot ». Correlations between the amount of forefoot ab/adduction and
ground reaction forces suggested that the open foot could be associated to a flexible-spring foot
while the closed foot was associated to a stiffer-spring foot. The purpose of the present study
was first to check the repeatability of such footprint measurements and then, to quantify the
forefoot ab/adduction in various sport situations. As an application, a new sport shoe design
will be proposed.

Methods

Three set of footprints were recorded by means of a photosensitive paper, in the following
conditions : Walking (30 subjects, walking uphill, level and downhill, with a 15 kg backpack
loading, at a mean velocity of 1.5±0.2 m .s⁻¹), Running (20 subjects, after 10', 20' and 30’ of
barefoot level running on the grass, at a mean velocity of 3.1±0.5 m .s⁻¹) and Jumping (30
subjects in impulse, landing and side-shuffle action). Five trials were performed for each
walking and jumping situation, instead of only one trial for running situations. Five static
footprints were also recorded for each subject.

On each footprint, the rearfoot and forefoot axis were identified according to the method
described by Freychat et al, 95 (1). The forefoot ab/adduction angle (α) was then measured
between the two axis (Fig. 1). A positive α was interpreted as forefoot adduction while a
negative α related a forefoot adduction.

In order to check the intra-experimenter repeatability of this footprint method, three
experimenters were asked to analyse 5 times a randomly selected set of 5 footprints. Average
SD for 5 measurements of the forefoot ab/adduction angle (α) on the same footprints were 1.6°,
1.5° and 1.1°. In order to check the inter-experimenter repeatability, a set of 30 footprints was
measured by each experimenter. Mean and SD for each experimenter were 3.5°±5.8°,
2.3°±6.2°, and 2.4°±5.9°. No statistical difference was found between these 3 experimenters.

For each sport situation, the five trials were then averaged and two parameters were calculated:
the relative change in forefoot abduction (α_{dynamic} – α_{Static}) and the intra-subject absolute
asymmetry (|α_{right foot} - α_{left foot}|).

Results

Consistent relative changes in forefoot abduction were observed in sport situations, especially
in downhill walking (from -2.4°±5.0° for the right foot to -4.3°±5.4° for the left foot), running
after 30’ (-3.4°±7.6° for the left foot to -3.7°±6.8° for the right foot) and impulse (-3.2°±6.3°
for the right foot, to -7.2°±4.8° for the left). Relative changes in forefoot adduction were
observed in side-shuffle situation (from +3.4°±7.6° for the left foot, to +4.2°±5.8° for the
right).
Maximum changes in forefoot abduction ran from -5.6° (side shuffle) to -19.7° (impulse). Maximum changes in forefoot adduction ran from +1.5° (impulse) to +15.4° (side shuffle). As a consequence, maximal changes were observed in Jumping situations. Consistent mean asymmetry to 3.4°±2.6° to 6.8°±4.3° was always observed between the right and the left feet (from 0° for the minimal to 22° for the maximal intra-subject asymmetry).

Discussion & Application
These results confirm those of Freychat et al. (95) by the range of forefoot ab/adduction and asymmetry, and by the fact that a tendency to greater forefoot abduction was observed when a flexible foot was required (higher impact loading due to impulse, fatigue or downhill walking).

It now appears that various sport situations need the foot to adapt it’s shape by means of consistent forefoot ab/adduction. The first consequence will be a dramatic change in foot shape that may induce some conflicts with the shoe. In addition, because of foot asymmetry, the same pair of shoe cannot fit one foot as well as the other.

It is therefore suggested to adapt the design of the shoe in order to let the forefoot ab/adduction act during the sport situation. A technical solution is presented as a two parts-construction of the shoe, where the rearfoot and the forefoot act independently and are gathered by means of a soft extensible textile on the upper part, while a rigid plastic bone, encapsulated in a soft flexible box, links the rearpart and the forepart of the sole unit. Such construction has been controlled by means of 3D video camera (Motion Analysis) recording the displacement of 7 markers putted on the upper and sole unit of the shoe, while two subjects walked downhill, level and uphill. Relative changes in shoe-forefoot ab/adduction (from -2° to +9°) were recorded and confirmed that such construction of the shoe may better respect the natural dynamic changes of the foot shape.

References
INJURY DRIVEN CHANGE TO THE FUNDAMENTAL DESIGN PARAMETERS OF THE AUSTRALIAN RULES FOOTBALL BOOT.
Simon J. Bartold
University of South Australia, Adelaide, Australia.

The direct cost to the Australian community of injury through sport is approximately $298 million. Of this over one third can be attributed to Australian Rules Football (Egger 1992). As a result of this study a need was identified to change the long standing design of the Australian Football boot. Prior to any fundamental design change an extensive needs survey of the sport was undertaken, examining:
- the demands of the game and the changes occurring over the past 10 years
- the injury patterns of the game
- the players
- the coaches
- the medical staff
- the current footwear.

The Game
Australian Rules Football is a unique sport with many features that considerably increase the risk of injury to the athlete.

Football is primarily a speed sport with players averaging 25-30 kilometres per game. The game is divided into four “quarters” with a short break between each “quarter”. The level and intensity of activity is therefore arguably the greatest of any weightbearing sport. It involves sudden, short changes in direction and cutting manoeuvres. Most players are of high body mass and therefore the risk and indeed incidence of anterior cruciate ligament injury is amongst the highest for any sport in the world. The cleat or “sprig” pattern has been implicated as a possible contributor to this high A.C.L. injury rate by reducing the release rate of the boot from the playing surface and therefore allowing unacceptable torque to be transmitted to the knee joint.

The game also involves kicking the ball distances of 60 metres or more with a very high degree of accuracy and jumping to catch the ball.

Football is a winter sport and is often played in very wet conditions.

As a result of this study radical changes to boot design have been implemented. These changes are expected to have a measurable effect on the incidence of injury in the sport.

This paper outlines the results of the study and the changes to Australian Rules Football Boot design. The changes are expected to have a flow-through effect to other sports, especially soccer and rugby.
INFLUENCE OF THE SHOE MECHANICS ON THE LOAD OF THE FOOT DURING THE PUSH-OFF PHASE
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novel Biomechanics Lab, Munich, Germany

A large number of special shoe constructions and materials were developed to reduce the load on the foot during the roll-over process, e.g. special cushioning to decrease the impact during initial heel contact. The aim of this study was to investigate how the shoe mechanics influence the load on the foot during the push-off phase.

For measuring the local loads on the plantar surface of the foot inside the shoe, the pedar pressure distribution measurement system was used (99 sensors per insole, scanning rate 50 frames/second). For examination of the pressure distribution on the dorsal surface of the foot a dorsal pad (24 sensors; 50 frames/second) connected to the pedar box was developed. For measuring the ground reaction force outside the shoe a Kistler platform inserted in the walkway and synchronized with the pedar system was used. Two different kinds of shoes (oxford and moccasin type) were used for testing.

During normal walking (speed 1.5 meters/second) with the oxford shoe the maximum values of the first peak of the vertical force curves showed no significant differences for measurements inside and outside the shoe. The second peak of the force curves measured inside the shoe was approximately 50% higher compared to the vertical force maximum measured outside of the shoe. The reasons for the additional forces inside the shoe are the following:
- The insole of the shoe is acting as an elastic membrane during bending. The surface tensions of this rather stiff membrane result in an additional force acting on the plantar surface of the foot.
- The creases on top of the shoe are also pressing on the dorsal side of the foot, therefore increasing the load on the plantar foot surface.

These facts were proven with measurements inside the soft moccasin shoe. The increase of the force inside the shoe during push-off was only 10-15% compared with the vertical force measured outside the shoe. This occurred because the insole of the shoe was more elastic (less surface tension) and the creases were much softer.

This demonstrates that certain kinds of shoes due to their shoe mechanics - are still increasing the force acting on the foot during a portion of the roll-over process.
CALIBRATION AND ANALYSIS METHODS FOR THE F-SCAN SYSTEM
Jonathan B. Fewster & Timothy J. Eng
NIKE Sport Research Lab, Beaverton, Oregon, USA

Introduction:
Many researchers have demonstrated limitations of the F-Scan in-shoe pressure measurement system (Tekscan, Inc., Boston, USA) such as time-dependent errors, hysteresis, non-linearities, and variability between sensors on an insole. At the same time, the system provides high spatial resolution in a non-intrusive, low-cost package. To utilize these benefits and minimize time-independent limitations, we developed several calibration and data analysis procedures to improve the validity of the quantitative data.

Methods:
The manufacturer recommends calibrating an F-Scan insole with either the subject standing on one insole or with an optional bladder calibration system which was not tested in this study. With the first protocol, the F-Scan software uses the subject’s body weight to calculate a single calibration factor for all sensors on the insole. However, non-linearities have been observed at high pressures. Furthermore, individual sensors are not consistent across the insole. To improve calibration, some researchers have used an air bladder to provide a load at uniform pressure over the whole insole.

To address the issues of non-linearity and inter-sensor variability, we developed a calibration procedure and its associated software. We calibrated each insole at 4 pressures (180, 350, 520 & 700 kPa) using an air bladder (Novel Gmbh, Munich, Germany). By measuring each sensor on the insole at each pressure, we developed an array of calibration factors for the insole. In subsequent conversions, experimental values were linearly interpolated between the four calibration values. At pressures above the highest pressure and below the lowest pressure, the nearest calibration factor was utilized to convert the data. After calibrating with a pressure bladder, an area compensation factor was necessary to account for the non-sensing portions of the insole. Thus to achieve total force, measures were multiplied by 1.8063.

During our testing with subjects, we observed individual sensors spiking – sometimes from baseline to saturated values. These spikes lasted no longer than several samples before returning to lower, more realistic values. Because the single sensor measured such a high magnitude while the adjacent sensors were much lower, it was doubtful that such measurements were real. While the cause of these spikes was uncertain, their affect on data was problematic. Several options were considered to minimize or remove these spikes. The method selected was a post-collection software algorithm we called a Spike Protection Factor (SPF) which compared the value of an individual sensor to the values of the adjacent sensors. If the sensor value was greater than 100 kPa and greater than the SPF times the average of the adjacent sensors, then the sensor value was changed to the average of the adjacent sensors.

Results and Discussion:
The shape of the force curve was compared to that measured by a force plate; with two subjects and a total of 12 landing trials, the Pearson Product Moment correlation was 0.928. Regressions between the F-Scan and force plate data yielded slopes between 1.09 and 1.29 – all more than two standard errors above the optimal value of 1.00. F-Scan was underestimating force, perhaps due to the sensors not being co-planar with the force plate, shear or loading non-
perpendicular to the sensors or a decrease in pressure magnitude with repetitive loading of the sensors.

A range of SPF values were evaluated in their ability to minimize spikes without attenuating true data. Two landing trials with many spikes had the following results:

Note that for subject 1, while the peak pressure of 819 kPa was erroneously high, the high SPF values had already removed a spike of 1668 kPa. The SPF value must be chosen carefully for each experiment. The selection may be facilitated by graphs such as those above, but observation of trials with problematic spikes before and after spike removal is the best way to confirm the ideal SPF for a given experiment.

After selecting an SPF of 3.0, with 10 subjects, a total of 194 trials, approximately 70 frames per trial, approximately 590 sensors sampled per frame, only 2127 sensor-frames were changed, averaging 10.96 sensors per 70-frame trial or less than 0.03% of total sensor-frames. With this low number we believe that true data was not attenuated but peak pressure values were more realistic with the majority of these spikes removed.

**Conclusion:**

These methods of calibration and data processing attempted to compensate for sensor non-linearity, inter-sensor variability and remove erroneous spikes, allowing greater quantitative utilization of the F-Scan system for force and pressure measurement. Further efforts may improve total force measures and address time-dependent errors and reliability of the system.

**References:**

METATARSAL LOAD AS A FUNCTION OF SHOE CHARACTERISTICS
P. Westblad, T. Hashimoto, N. Murphy, A. Lundberg.
Dept. of Orthopaedics, Karolinska Institute at Huddinge Hospital, Stockholm, Sweden

The incidence of metatarsal stress fractures among Swedish Marines has increased. Several environmental conditions have been altered; most notably the previously used stiff army boot was replaced by a new boot with a more flexible sole. This pilot study was undertaken to clarify the influence of boot type on external load on the metatarsal region at conditions similar to Marine basic training.

Four subjects participated in the study. A Tekscan F-Scan sensor was placed inside an outer sock on the right foot. Each subject performed walking trials with a 40 kg backpack under each of three conditions: barefoot, with the stiff (older) army boot and with the new boot. The footprint was divided into six regions for which the force impulse was calculated and expressed as a percentage of the total impulse.

Under barefoot conditions a considerably smaller proportion of the load was carried by the heel region (30% vs. 44 and 54%), corresponding to a larger load on the forefoot. There were no significant differences between the loading conditions for the boots; however, considerable differences between individual steps was observed.

Current projects include load distribution analysis with a larger number of subjects and direct metatarsal deformation analysis.
A clinical assessment was performed on 15 marathon runners together with measurement of in-shoe plantar pressures 3 days prior to running the London Marathon (1997). The clinical assessment was performed by a podiatrist using a computer program used for the prescription of orthoses. In-shoe plantar pressures were measured using the pedar system with each athlete running on a treadmill at their marathon pace and wearing their marathon shoes. Every runner complained of an injury or pain. The injuries presented were diverse and the in-shoe pressure measurements showed high variability in the loading patterns between runners and often between the feet of the same runner. In the opinion of the podiatrist, orthoses would have been required in some cases but in others, selection of an appropriate running shoe may have sufficed. The high incidence of injuries in marathon runners dictates that there may be a need to provide footwear that are designed to accommodate an assistive device prescribed by a clinician.
DEVELOPMENT OF A RUNNING SHOE FROM DIGITIZED FOOT SHAPES
Gordon Valiant and Robert Borchers
Nike, Inc.

Introduction

Lasts are the forms over which shoes are made1 and are essential in running shoe construction. They also provide the internal shape that accommodates the morphology of a broad segment of a running population, thereby significantly influencing fit and comfort, as well as function.

It was the purpose of this study to develop running shoes with enhanced fit and comfort by developing lasts that more closely approximate the shape of the male and the female foot.

Methods

Plaster casts of the right foot of 30 male subjects (foot length = 263.5 ± 2 mm) and 27 female subjects (234.0 ± 2 mm) were obtained under 1 body weight (BW) load. Casts were scanned with 2 CCD cameras located medially and laterally, each of which recorded laser light beams reflected off the plaster (Cyberware®). Coordinates defining surface detail of the plaster feet were obtained. Medial and lateral views were aligned and overlapping points removed2. Coordinates were reduced to frontal plane cross sections spaced 2 mm apart, smoothed, and expressed relative to a horizontal axis. Coordinates for all male and all female foot casts expressed relative to similar axes were averaged to define three dimensional shapes. A numerically controlled milling machine cut models of each 3D foot shape.

The models were used by a pattern maker to create uppers, a model maker to make anatomically contoured midsole moulds, and a shoemaker to assemble shoes. These shoes were tested by groups of male and female runners. Shortcomings in fit and comfort were addressed by length increases, shape changes in the front tip, addition of toe spring, and toe height increases. These changes satisfied requirements for foot dynamics during running. Last shape changes, shoe construction, and subsequent running feedback occurred in iterative steps until last shapes were finalized. The process lead to gender specific lasts for producing foot shaped shoes.

Results

When compared to running shoes lasted with midsoles, upper patterns, and materials that were equivalent but with shapes dictated by conventional lasts, foot shaped running shoes were preferred by male testers. The greatest differentiation was improved arch fit and support. Also, heel fit was substantially snugger, with a wider and more comfortable toe and forefoot. The forefoot region of the foot shaped last was 5-10 mm wider (Figure 1) but 1-13 mm shorter (Figure 2) than a conventional running shoe last. However, differences in girth of the frontal plane cross sections were relatively smaller (Figure 3). The increased width of the foot shaped last was offset by reduced height to achieve nearly equivalent girths in some regions.

In general, foot shape lasted shoes were not preferred by female testers.

Discussion

This shoe development process provided a static, rigid, weight bearing shape. Ignoring foot shape changes while loads increase at a fast rate from 0 to > 2 BW, then decrease, was a
Figure 1. Width of parallel sections projected in a horizontal plane of the foot shaped last (solid) and a conventional last (dashed) plotted from heel to toe.

Figure 2. Width of parallel sections projected in a sagittal plane of the foot shaped last (solid) and a conventional last (dashed).

Figure 3. Girth of parallel sections in a frontal plane of the foot shaped last (solid) and a conventional last (dashed) plotted from heel to toe. Girth and height differences at the opening for the ankle result from arbitrarily selected heights.

Adjustability and adaptability of fit in running shoes designed for a broad segment of the running population is desirable. Generally, women’s running shoes are modifications of shoes originally designed for men. For this study the women’s running shoe developed from data specific to them did not enhance fit, even though data agreed with previously reported sexual dimorphism of the human foot.

For example, the final shape either fit too tightly in the forefoot or not tight enough. The slightly greater variability in average foot shape compared to men may perhaps support a need for greater adjustability in women’s running shoes.

References
Dual density midsoles provide medial stability when performing a running movement. We analyzed two different dual density midsoles: one had a medial side harder in the forefoot only while the other had the medial side hard on the entire length of shoe. This study involved twelve subjects. As the subject ran on a treadmill at 3.8 m/s we performed high speed video analysis at 265 frames per second. Two markers were placed respectively on the Achilles Tendon and Gastrocnemius area of each subject. Two markers were also placed on the heel area of the shoes. The results indicate that dual density midsole construction in the forefoot only, is sufficient to provide good medial/lateral stability.

Introduction
Evaluating the medial/lateral stability of footwear for the most part has involved varying the heel construction of running shoes. (Frederick 1984, Nigg 1986, Cavanagh 1990). These construction techniques included midsole hardness, heel flare, heel counter hardness and dual density midsole construction, where some medial aspect of the midsole consists of a harder material than the lateral aspect.

These researchers showed through varying the construction in the rearfoot of running shoes an effect on the pronation movement (Rate of Pronation, Maximum Pronation, Range of Motion(RoM)) could be obtained. However, the effect of forefoot construction has yet to be thoroughly investigated.

Also, we have speculated because of past adidas studies which showed high pressure under the area of the first and second meta-tarsal heads and the time when maximum pronation occurs, well after forefoot contact, that forefoot construction can have an effect on both range of motion and maximum pronation. To validate this theory a pilot study was performed on two pairs of shoes. One with a dual density forefoot and one with a single density forefoot. This pilot study showed that a difference did exist between the two shoes.

From this result we hypothesized that to limit the range of motion and maximum pronation values dual density construction was only needed in the forefoot and not in the heel area of the shoe. Therefore no difference should exist in the two shoes which are being evaluated for this study.

Methodology
In this study we analyzed two pair of running shoes with varying dual density midsole construction. Shoe 1 has a complete dual density midsole. With a medial aspect of 60° Shore FWC and a lateral aspect of 40° Shore FWC. Shoe 2 was constructed the same as shoe 1 in the forefoot while the heel was a single density.

Twelve subjects took part in the range of motion portion of this study while 10 subjects were involved in the maximum pronation portion. Each subject ran on a treadmill at 3.8 m/s and were filmed from the rear following Clarke et al, for frontal plan evaluation, figure 1.

The camera system which was used recorded at 265 frames per second.
The range of motion and maximum pronation values were determined by the Achilles Tendon angle as shown in figure 2, and maximum pronation is reported as a negative value. All values were calculated by dropping the high and low of the initial five trials determined by the frame when forefoot contact occurred and averaging the remaining three trials together.

**Results**

The following are the results for the two shoes involved in this study.

<table>
<thead>
<tr>
<th></th>
<th>Shoe 1</th>
<th>Shoe 2</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>RoM</td>
<td>15.4°</td>
<td>15.5°</td>
<td>0.98</td>
</tr>
<tr>
<td>Max. Pronation</td>
<td>-7.8°</td>
<td>-8.1°</td>
<td>0.36</td>
</tr>
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</table>

**Conclusions/Recommendations**

As the data indicated no difference appeared between the two shoes. Showing that shoe 1 and shoe 2 performed in a similar fashion in regards to RoM and maximum pronation.

Granted the experimental design for this study may not have been complete because no dual density rearfoot only shoe was tested. However, the results of this study as well as the past adidas study referenced in this paper show that forefoot construction can be an important part of footwear construction as it relates to RoM and maximum pronation. Therefore, additional testing needs be performed to obtain a more well rounded understanding of forefoot construction and it’s effect on performance.

Also, this study brings into question the way in which the pronation movement is evaluated. Presently the foot is evaluated as a single unit in regards to pronation. However, is this the correct way to understand the foot/shoe interaction during the running stance phase? Possibly it would be more beneficial to evaluate the heel and forefoot as two separate units and/or their effect upon each other.

**References**


THE INFLUENCE OF RESTRICTED REARFOOT MOTION ON IMPACT FORCES DURING RUNNING
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Institute for Athletics and Gymnastics, German Sport University - Cologne; Germany

Introduction
In the literature high passive forces and excessive pronation have been closely linked to injuries in distance running. Many attempts have been made to reduce both, the initial impact peak and the range or velocity of pronation (Nigg & Morlock, 1987; Nigg et al., 1988; DeWit et al., 1995). However, analytical approaches suggest that these cannot be minimised at the same time (Denoth, 1986). Experimental studies focussing on this relation show contradictory results and different individual adaptation strategies as reaction to changed footwear have been discussed (Cavanagh, 1990). To ascertain if restriction of rearfoot movement influences the magnitude of initial impact peak without changing the shoe itself, different possibilities of external movement limitation were applied.

Methods
Nine recreational runners participated in this study. Their mean age, height and weight were 29.4 ± 5.9 years, 179.3 ± 1.6 cm and 75.6 ± 4.8 kg respectively. All of them were rearfoot strikers, injury free and experienced in treadmill running. Two shoe types were used during the testing sessions: Asics Gel 125 ES running shoe (R) and Asics Sport Track II Sport Sandal (S). The sandal was used to enable the subject to maintain a heel-toe running style without influencing the movement of the calcaneus. Each shoe was tested for four different conditions: no restriction (NR), taped (TP) in a supinated position via standard tape (Leukotape®), wearing an Aircast® ankle brace (AC) and the combination of splinting and taping (AT). During the experiment subjects were required to run on a motor driven treadmill (Gaitstar; BFTS, Cologne) at a velocity of 3.5 ms⁻¹. When a stable running pattern was reached, data collection began and continued for a period of 15 s. A strain gauge based forceplate build into the treadmill was used to collect ground reaction force data, while an in-shoe goniometer recorded rearfoot motion. Additionally an hydrocell force sensor (2 cm in diameter; Paromed, Markt Neubeuern) was positioned inside the shoe under the heel and a uniaxial accelerometer (Biovision, Wehrheim) was fixed with adhesive and a rubber band on the medial aspect of the tibia. All data were A/D converted and recorded simultaneously with a sampling frequency of 1000 Hz. The subjects were videotaped in a sagittal view with a 50 Hz video camera (Sony CCD, S-VHS video system) and ankle and knee angles were analysed by commercially available software (PEAK Motus, Englewood CO). Prior to identifying the listed parameters (table 1) the raw data were filtered using standardised procedures.

Each subject performed one trial for each testing condition of which 10 steps were selected for further analysis. Prior to each test the goniometer offset was determined with the subject standing still without shoes in a standardised position. The rearfoot angle was negative for

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Code</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td>External impact peak</td>
<td>(F_p)</td>
<td>N</td>
</tr>
<tr>
<td>Internal impact peak</td>
<td>(F_{ip})</td>
<td>Ncm⁻²</td>
</tr>
<tr>
<td>Internal force rate</td>
<td>(dF_{ip})</td>
<td>Ncm⁻²s⁻¹</td>
</tr>
<tr>
<td>Maximum acceleration</td>
<td>(a_{max})</td>
<td>g</td>
</tr>
<tr>
<td>Initial rearfoot angle</td>
<td>(\beta_{TD})</td>
<td>°</td>
</tr>
<tr>
<td>Minimum rearfoot angle</td>
<td>(\beta_{min})</td>
<td>°</td>
</tr>
<tr>
<td>Range of pronation</td>
<td>(\beta_{range})</td>
<td>°</td>
</tr>
<tr>
<td>Knee angle at touch-down</td>
<td>(\gamma_{TD})</td>
<td>°</td>
</tr>
<tr>
<td>Sole angle at touch-down</td>
<td>(\delta_{TD})</td>
<td>°</td>
</tr>
</tbody>
</table>
The supination at touch-down and the range of motion are displayed in figure 1. The foot strikes the ground in a more supinated position when taped (TP). Using the splint (AC) the foot orientation appears less supinated than in unrestricted running (NR). The AT condition was comparable to NR as it had the combined effect of TP and AC with regard to foot orientation at touch-down. The range of motion is comparable for NR/TP and AC/AT. The absolute values show only slightly greater values for sandals. These data demonstrate clear effects of the applied restriction to rearfoot movement in running.

The sandal produces greater tibial acceleration (a_{\text{max}}) values, but no considerable differences between restriction conditions can be seen. Impact peaks are only slightly different in magnitude for running shoes and sandals (fig. 2). For the running shoes the peaks are greatest for AT and TP which is in contrast to the hypothesis of pronation as an impact-reducing mechanism (Denoth, 1986). The sandals display a different pattern; AT and TP lead to lower impacts. The analysis of the other parameters (table 1) also shows only slight and not systematic variations. The internal impacts show a comparable pattern to the external forces and only minimal changes are found in knee and sole angle at touch-down.

From the current data it can be concluded that differences in pronatory movement are not directly coupled to the magnitude of impact shock at touch-down in heel-toe running and that formerly discussed parameters cannot be used to explain the differences (Nigg & Bahlsen, 1988).

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Motion During Foot Contact in Running. Journal of Applied Biomechanics 11: 395 - 406
RELATING FOREFOOT STRUCTURE TO DYNAMIC PLANTAR PRESSURE DISTRIBUTION DURING LOCOMOTION
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Introduction
The pads of fat under the heel and ball of the foot compress at ground contact during locomotion. This helps to cushion and protect the musculo-skeletal structures of the body from damage. Pad thickness under the heel and metatarsal heads has been estimated using ultrasound techniques and there is evidence to suggest that individuals with reduced fat pad thickness are at a greater risk of overloading the foot (1). Although it has been inferred that the extent and distribution of plantar soft tissues affect the loading of the foot, the influence of the size and deformational characteristics of the fat pads on the ability to cushion plantar loads observed during locomotion remains to be determined. Little is known about the relationship between volumetric characteristics of the plantar fat pads and dynamic loading, particularly in the forefoot. Magnetic resonance imaging (MRI) is a non-invasive scanning technique which, in combination with stereological methods, can allow efficient quantification of tissue compartment volumes. The aim of this research was to determine the size of the forefoot fat pad under load using MRI and then relate those results to dynamic plantar pressure profiles at corresponding locations recorded during barefoot walking and running.

Methodology
Loads under the foot during locomotion were recorded on eight subjects with no recent history of lower extremity injury. Subjects walked barefoot at a constant speed (4 km/hr) along a 6 m walkway, hitting a pressure mat (EMED-SF, Novel gmbh) mounted flush in the middle. Plantar pressure was recorded at 50 Hz and ten contacts for each foot were collected for averaging purposes (coef. of reliability for peak pressure of >0.95, ref. 3). Similarly, ten contacts were recorded during jogging at a comfortable speed for each individual. Metatarsal head (MTH) location while standing was determined by using palpation and a template for each individual foot of the relative head location obtained from MRI. Then, a footprint with inked MTH locations was obtained for each foot during walking using a Harris mat. This print was superimposed onto a peak pressure plot of each mat transducer (4 per cm2) recorded during walking and jogging. This allowed identification of the transducers under each MTH. The magnitude and timing of loads were assessed for all five MTHs on each foot for the subjects.

A novel apparatus was constructed to allow high resolution MRI of the forefoot while loaded. Each subject was supine with their shoulders supported and knee flexed (45 degrees) so that they could lightly press their barefoot flat against a wooden vertical plate positioned inside the magnet. Their ankle posture was approx. 20 deg. of dorsiflexion. The forefoot of both feet for each subject was coronally scanned inside a wrist coil using a fast 3D gradient echo. This coil (13 cm diam.) was employed to maximize the signal to noise ratio. A series of two-dimensional digitally encoded images related through depth produced the three dimensional framework for volumetric measurements. Each volume element (voxel) was 0.31 mm3. For each MTH, volumes of bone and underlying fat were estimated using the Cavalieri principle (5). Fat compartments were defined with the aid of MR T2 relaxation time mapping (4). The effectiveness of the fat cushion under each MTH was determined by calculating the percentage increase in peak pressure (delta peak) and average load rate (delta rate) after the loading profile for running was compared to that for walking. Fat pad size was expressed relative to the size of the corresponding MTH.

Results
Preliminary results indicated that the highest forefoot pressures recorded during walking were under MTH2 and 3 which is in agreement with the findings of other authors (eg. 6). For jogging the peak pressure under MTH2, 3 and 4 increased an average of 20-30% while there were much greater proportionate increases in the peak loading under MTH1 and 5 (see Table 1).
Table 1. Mean dynamic plantar pressure variables for seven subjects (± SD)

<table>
<thead>
<tr>
<th></th>
<th>MTH1</th>
<th>MTH2</th>
<th>MTH3</th>
<th>MTH4</th>
<th>MTH5</th>
</tr>
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<tr>
<td>Peak Pres. (N/cm²) Walking</td>
<td>33 (15)</td>
<td>46 (12)</td>
<td>43 (9)</td>
<td>35 (8)</td>
<td>23 (11)</td>
</tr>
<tr>
<td>Peak Pres. (N/cm²) Jogging</td>
<td>53 (20)</td>
<td>57 (19)</td>
<td>51 (9)</td>
<td>45 (10)</td>
<td>45 (23)</td>
</tr>
<tr>
<td>Delta Peak (%)</td>
<td>79 (62)</td>
<td>23 (22)</td>
<td>20 (14)</td>
<td>29 (14)</td>
<td>111 (109)</td>
</tr>
<tr>
<td>Delta Rate (%)</td>
<td>593 (289)</td>
<td>299 (97)</td>
<td>288 (76)</td>
<td>296 (67)</td>
<td>371 (221)</td>
</tr>
</tbody>
</table>

From walking to running the percentage increases in peak pressure and average loading rate (delta peak and rate) were much greater under MTH1 and 5. These locations under the foot also demonstrated the largest intersubject variability in loading profile and load increases with jogging. These findings indicated that, in general, cushioning was reduced in these areas and individual structural factors might explain subject differences in cushioning. For these reasons, initial data analysis focused specifically on the size of the fat cushions under MTH1 and 5. For four subjects, the mean (±SD) fat : bone volume ratios for MTH1 and 5 were 0.53 (0.07) and 0.13 (0.06), respectively. There was relatively little fat present under the MTH5 (only 13% of the MTH bone volume) which could potentially have explained the large delta peak values found under this MTH. However, poor correlations were obtained when fat pad size for both MTH1 and MTH5 for these subjects was related to their corresponding delta peak data. The delta rate variable also demonstrated a non-significant correlation with the size of the fat pad under MTH1 and MTH5.

Discussion

The presented methodology allowed specific forefoot structures to be associated with localized plantar pressure measures during locomotion. The effectiveness of the fat pads under the forefoot to cushion the plantar loads experienced was examined by calculating the changes in forefoot loading from walking to jogging. Initial results indicated that fat pad size under each MTH did not influence the changes in forefoot loading. Although reduced plantar tissue thickness has been associated with the risk of tissue injury in diabetics (1), for healthy subjects the highest plantar pressures have been found under those MTHs with the thickest layers of underlying plantar tissue (2). The latter author speculates that increased plantar tissue thickness is a functional adaptation to elevated pressures experienced during locomotion. As such thickness measures include fat and other ligamentous tissue structures, it is not clear which tissues might be involved in the proposed adaptation process. The findings of the present study suggests that other factors, in addition to the fat pads, likely contribute to cushion the increased forefoot loads experienced during jogging. Factors such as joint mobility, sesamoid deformation and other possible cushioning tissues, should be investigated alongside fat pad characteristics in future studies of the mechanics of forefoot cushioning. This future work would be geared towards the identification of individuals at risk of forefoot overloading and the design of appropriate protective footwear.

References.

(1) Gooding et al. (1986) Investigative Radiology, 21: 45-49.
EXTERNAL ANKLE TORQUE - A COMFORT PARAMETER FOR FOOTWEAR
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Gait Analysis Laboratory, Central Leather Research Institute, Adyar, Chennai 600 020, India

INTRODUCTION
Comfort of footwear is a complex phenomenon. It could be considered that absence of discomfort is comfort. There are several factors that would contribute to footwear comfort. The fitting properties, thermophysiological properties, physical properties, weight, stiffness, footwear construction, shock absorption capability are some of the parameters. The normal range for these parameters and not just normal values have to be identified. Some of these parameters can effectively be measured by employing a movement analysis system using a video digitization technique with attached force platforms. These type of dynamic studies and tests would serve a better purpose to analyze comfort when compared to static laboratory tests. The footwear for day long wear must absorb shock and should be resilient to avoid leg and back strain. The dorsiflexion and plantarflexion are the two ranges of movements that occurs at the ankle joint during human gait. The torque during these flexion arises due to the external reaction forces that are generated during heel strike and push off periods. Peak dorsiflexion and plantar flexion torque values are considerably less during normal level walking. However, these peak moment values and their timing are dependent on the footwear bottom components. Viz. the outsole and the midsole. A study to look at the flexion torque values have been conducted and the results are reported and discussed.

METHODOLOGY
The experiments were conducted in a Gait Laboratory equipped with a VICON movement analysis system running AMASS software. 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 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 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 trials. 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 calculations and comparisons. The net moment that caused the flexion at the ankle is calculated as follows:

\[ \text{Max} = -F_z \times (P_x - R_x) \]

where Max is the ankle moment in the X Direction in Nm, Fz is the vertical ground reaction force in Nm, Px is the x coordinate of the centre of pressure in mm, Rx is the x coordinate of the ankle joint in mm.
RESULTS

Ankle dorsiflexion and plantarflexion moments for barefoot walking were 8.45 and 78.33 Nm on an average, whereas the corresponding values for different combinations of outsole are given in the table below:

COMPARATIVE MOMENT VALUES FOR DIFFERENT FOOTWEAR

<table>
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<tr>
<th>WALKING WITH</th>
<th>MOMENT VALUE (Nm)</th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Outsole</td>
<td>Midsole</td>
<td>DORSIFLEXION</td>
<td>PLANTARFLEXION</td>
</tr>
<tr>
<td>Leather</td>
<td></td>
<td>13.48</td>
<td>74.69</td>
</tr>
<tr>
<td>Hard MCR</td>
<td></td>
<td>15.36</td>
<td>79.68</td>
</tr>
<tr>
<td>Leather</td>
<td>Soft MCR</td>
<td>10.89</td>
<td>75.94</td>
</tr>
<tr>
<td>Hard MCR</td>
<td>Soft MCR</td>
<td>10.51</td>
<td>77.80</td>
</tr>
</tbody>
</table>

Results show that the peak dorsiflexion moment is lowest for barefoot walking (7.28 Nm), and it increases when walking with footwear. However midsoled footwear results in a increase of only 3 Nm, Whereas footwear with leather and hard MCR outsole resulted in moments twice that of barefoot walking (15.02). However the bottom components did not change ankle plantarflexion moment appreciably. During heel strike in barefoot walking, the line of action of the vertical impact force passes closely to the ankle joint axis. Hence the moment produced with respect to this axis is small and the forces on the long extensor muscles of the foot are small. The load situation on the muscles changes while wearing a footwear. This effect increases with increasing sole stiffness and flexibility. During impact the moments produced by the ground reaction force and increased lever is too big to be equalized by the anterior tibial muscle group and there fore a forefoot slap is inevitable with hard soled footwear. Hence the muscles are loaded eccentrically at a higher rate, that might lead to certain insertion problems at various tendon levels. The wear of the outsole is caused by the relative motion between the outsole and ground, hence there is a relationship between the regions of maximum wear and regions of maximum forces. Due to this the outsole would scuff upon initial contact. The moment about the vertical axis could be correlated with the amount of scuffing using the biomechanical force platform.

REFERENCES

INTRODUCTION

To date, athletic shoes have been principally classified by gender and sport. While age is an established classification in childhood, only minimal attention has been given to adapting footwear to structural and functional changes in older individuals. In order to address the question of whether age should be a consideration in shoe design, this study was conducted to identify changes in foot structure and function of healthy individuals across the adult life span.

METHODS

Fifty-five subjects ranging from 20-69 years of age (11 subjects per decade) volunteered to participate in the study. Foot structure was obtained from standardized lateral and dorsi-plantar weight bearing plain radiographs, passive range of motion, and an ultrasound based system for characterizing soft tissue properties (1). Foot function was obtained during steady speed barefoot walking at 0.78 statures per second from stride parameters, 3D rearfoot joint kinematics, EMG of selected foot muscles, and plantar pressure under five foot regions. Analysis of variance was used to identify structural and functional differences between the five age groups.

RESULTS

The results (Table 1) indicated significantly reduced passive range of motion at the subtalar joint, and a trend towards reduced talocrural range of motion with increasing age. No differences in dynamic range of motion during walking at either of the two joints were found.

We found a significantly reduced heel peak pressure and a non-significant trend towards stiffer soft tissue under the heel (estimated by the force required to compress the tissue to 66% of its total compression) in the older groups. No significant differences or trends were found in the height of the medial longitudinal arch during standing (determined from radiographic measurements) or the unloaded thickness of the heel pad (determined from ultrasound) between the age groups. Similarly, no significant differences in peak pressures were found under the midfoot, forefoot and hallux.

The older subjects walked with a shorter stride length and significantly increased activation of the tibialis anterior, expressed as a % of maximum voluntary action (MVA), during the first quarter of stance.

DISCUSSION

It has been well documented that walking speed affects various gait parameters (2). In this study we controlled for walking speed within a range of ±10%. In addition, there were no differences in body dimensions (height and weight) between the age groups.

It has been speculated that increased joint stiffness may cause reduced foot motion in the elderly (3). The results of this study indicate that passive ROM decreased with age, however, dynamic ROM at both the talocrural and subtalar joints did not change. The correlation

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coefficients between static and dynamic ROM at both joints were low. Only approximately 25% of the available maximum range of motion was used during walking. The mean ratio of dynamic/passive ROM at the subtalar and talocrural joints for all subjects were 0.25 and 0.26 respectively.

It has been speculated that lower heel pressures found in the older individuals, compared to younger ones, are due to lowering of the medial longitudinal arch which results in increased midfoot load (4). The results of this study suggest that the main reason for this difference may be a stiffer heel pad in the older subjects. It is likely that longitudinal arch structure played only a minor role in the process since midfoot peak pressures were not significantly increased in the older subjects.

Increased activation of tibialis anterior (TA) at the first quarter of stance supports the results from Nigg and Skleryk (3) who reported a smaller change in shoe sole angle (the angle between the sole of the shoe and the ground from lateral view) over the first 3 frames of stance (0.06 s). The reason for a smaller change in this angle may be a choice for a more controlled approach, perhaps to reduce forefoot pressure in late stance, since tibialis anterior acts eccentrically to control foot motion between heel strike and foot flat.

This study found only a small number of significant changes in foot structure and function as a function of age. Among the limitations of the present study are the cross sectional design, the relatively small number of subjects in each decade group, the absence of individuals over 69 years old, and the fact that slow walking was used as the functional task. However, the several trends that have been identified provide potential directions for future studies.

**Table 1.** Mean values and ANOVA results for selected structural and functional measures

<table>
<thead>
<tr>
<th>Variable</th>
<th>Units</th>
<th>20-29</th>
<th>30-39</th>
<th>40-49</th>
<th>50-59</th>
<th>60-69</th>
<th>p</th>
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<tbody>
<tr>
<td>subtalar passive ROM</td>
<td>deg</td>
<td>49.3</td>
<td>45.1</td>
<td>37.7</td>
<td>45.7</td>
<td>37.6</td>
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<tr>
<td>subtalar dynamic ROM</td>
<td>deg</td>
<td>9.7</td>
<td>10.3</td>
<td>10.9</td>
<td>9.2</td>
<td>9.8</td>
<td>0.47 ns</td>
</tr>
<tr>
<td>talocrural passive ROM</td>
<td>deg</td>
<td>70.6</td>
<td>66.5</td>
<td>61.3</td>
<td>63.0</td>
<td>59.8</td>
<td>0.12 t</td>
</tr>
<tr>
<td>talocrural dynamic ROM</td>
<td>deg</td>
<td>16.8</td>
<td>16.0</td>
<td>19.5</td>
<td>17.2</td>
<td>18.1</td>
<td>0.35 ns</td>
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<tr>
<td>heel pressure</td>
<td>kPa</td>
<td>463.6</td>
<td>392.5</td>
<td>456.1</td>
<td>345.3</td>
<td>390.3</td>
<td>0.008 s</td>
</tr>
<tr>
<td>Force at 66% compression</td>
<td>N</td>
<td>22.7</td>
<td>26.9</td>
<td>25.1</td>
<td>32.7</td>
<td>31.9</td>
<td>0.13 t</td>
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<tr>
<td>Calcaneal inclination</td>
<td>deg</td>
<td>22.7</td>
<td>21.5</td>
<td>21.4</td>
<td>19.9</td>
<td>21.0</td>
<td>0.89 ns</td>
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<tr>
<td>Heel pad thickness</td>
<td>mm</td>
<td>14.5</td>
<td>15.0</td>
<td>13.6</td>
<td>13.8</td>
<td>14.3</td>
<td>0.49 ns</td>
</tr>
<tr>
<td>Step length</td>
<td>cm</td>
<td>69.4</td>
<td>71.1</td>
<td>66.0</td>
<td>67.9</td>
<td>65.0</td>
<td>0.07 t</td>
</tr>
<tr>
<td>Tibialis anterior activity</td>
<td>%MV A</td>
<td>21.7</td>
<td>28.5</td>
<td>25.0</td>
<td>40.3</td>
<td>36.0</td>
<td>0.002 s</td>
</tr>
</tbody>
</table>

s = significant finding, t = trend, ns = non significant, (p=0.05)

**REFERENCES**

**ACKNOWLEDGMENTS**
The authors would like to thank Jan Ulbrecht, M.D., Vladimir Zatsiorsky, Ph.D., Janice Derr, Ph.D., and the students and staff of the Center for Locomotion Studies at Penn State for their contribution to this study.

FOREFOOT ABDUCTION AND ITS RELATION TO CHANGES IN FOOT LENGTH
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†University of Calgary, Calgary, Canada and ‡Decathlon Footwear, France

The human foot can be approximated as separate forefoot and rearfoot rigid bodies. However, no 3-D data has been published on dynamic relative motion between these two segments and the associated changes in foot shape. The purpose of this study was to quantify forefoot ab-adduction relative to the rearfoot in vitro, and to determine the relationship between relative forefoot ab-adduction and foot length. Video data was collected from reflective marker triads affixed to the ends of Steinmann pins drilled into the tibia, calcaneus, cuboid, and the first and fifth metatarsal bones. Forefoot ab-adduction relative to the rearfoot and foot length were calculated under two axial tibial loads (200N, 600N) and two input motions (dorsi-plantarflexion, internal-external tibial rotation). Ex-internal tibial rotation always produced relative forefoot ad-abduction, with increasing axial tibial load reducing the range of movement. An increase in medial foot length was associated with increasing relative forefoot abduction. However, the forefoot abducted relative to the rearfoot during plantarflexion for flexible feet, but adducted during plantarflexion for rigid and normal feet. These results suggest that the relative movement between the forefoot and rearfoot in the horizontal plane is dependent on the degree of rigidity of the foot.
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The ISB Technical Section on Footwear Biomechanics has a web site at:

http://www.teleport.com/~biomech/sneakers.html

The site includes Symposium abstracts, downloadable files and other information about the Technical Group.

Other sites of interest:

- ISB Home Page
- Biomechanics World Wide
- BIOMCH-L
- Foot & Ankle Special Interest Group

http://www.kin.ucalgary.ca/isb/index.html
http://dragon.acadiau.ca/~pbaudin/biomch.html
http://www.kin.ucalgary.ca/isb/biomch-l.html
http://jan.ucc.nau.edu/~cornwall/fasig/fasig.html

THE INTERNATIONAL SOCIETY OF BIOMECHANICS

The International Society of Biomechanics is an organisation dedicated to the promotion and support of scientific research in the field of biomechanics. Areas of interest include Sports Biomechanics, Clinical and Rehabilitation Biomechanics, Locomotion, Neuromuscular and Tissue Biomechanics. Technical sections of the ISB cater for members with special interests in Computer simulation, 3D analysis of human movement and footwear biomechanics.

Every two years, the ISB hosts an International Congress of Biomechanics. The most recent Congress, the 16th, was held in Tokyo, Japan during August 1997. Over 400 papers were presented in many different fields of biomechanics. The next meeting will be in Calgary, Canada, during August of 1999.

Contacts for more information:

- The ISB Home Page (http://www.kin.ucalgary.ca/isb/index.html)
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