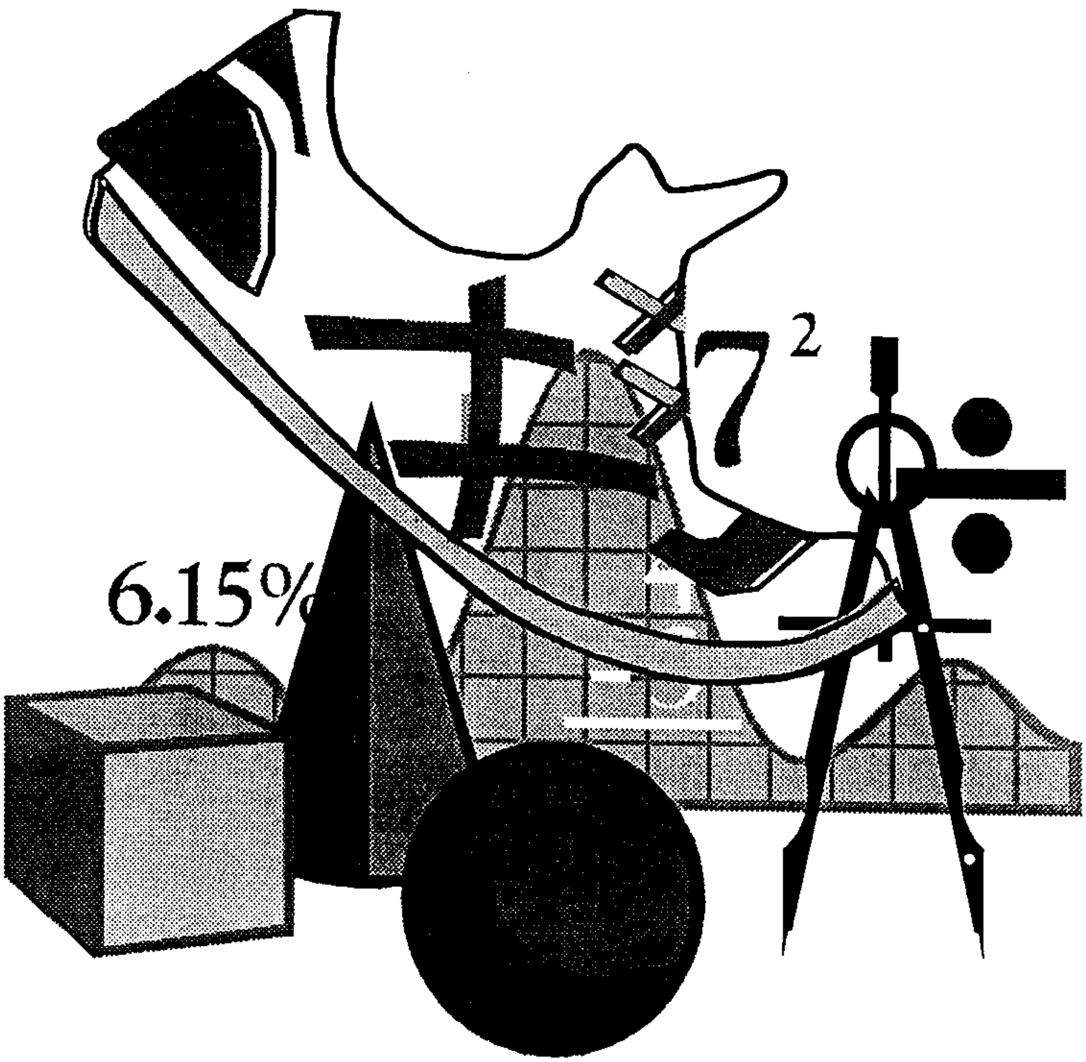
International Society of Biomechanics
WORKING GROUP ON FUNCTIONAL FOOTWEAR

Symposium on

THE BIOMECHANICS of FUNCTIONAL FOOTWEAR



held in conjunction with the
Eighth Biennial Conference of the Canadian Society of Biomechanics

University of Calgary
August 19-20, 1994

INTRODUCTION

The International Society of Biomechanics Working Group on Functional Footwear was formed in July 1993 to provide a forum for those interested in biomechanical aspects of functional footwear, including clinical and athletic footwear. The Working Group currently comprises more than eighty members from around the world. A primary goal is to encourage research and promote discussion of biomechanical issues related to functional footwear by disseminating information and encouraging interaction between group members.

This volume contains abstracts from the Working Group's first Symposium on the Biomechanics of Functional Footwear. The Symposium was held in August of 1994, in conjunction with the Canadian Society of Biomechanics Biennial Conference at the University of Calgary. In addition to abstracts from some of the invited papers at the Symposium, this volume also includes, by kind permission of the editors, abstracts of relevant papers from the Canadian Society meeting.

By linking the Symposium to the Canadian Society of Biomechanics meeting, the Working Group was spared a lot of organisational and administrative work and freed to concentrate on the assembly of a strong scientific program. Our thanks to Dr. Walter Herzog, CSB Conference Chair, who had the insight to recognise that the symbiosis would be good for both groups and the organisational skills to ensure that it worked.

The Symposium was funded entirely by the donations of sponsors from the footwear industry and related businesses. Adidas America Inc., NIKE Inc., Novel gmbh, Schering-Plough, Spenco and Retama Technology Corporation all made generous contributions. The Symposium would not have occurred had these companies had not seen fit to support basic scientific effort without thought for immediate commercial benefit.

Following the success of this Symposium a second one is to be held at the Deutsche Sporthochschule, Cologne, Germany, June 28-30 1995.

For more information on the Working Group and future Symposia, please contact:

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CONTENTS

Mechanical characteristics of the human heel pad at impact loading as deduced from different experimental procedures

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A.Introduction

Running at low to average speeds generally involves a foot strike pattern with heel contact first. In the barefoot condition, the resulting transient force depends distinctly upon the mechanical characteristics of the fatty tissue beneath the heel bone. However, running with athletic footwear implies a mechanical "collaboration" between the shoe and the heel pad during heel strike.

Apart from the evident local protection by covering the heel bone, two dynamic mechanical functions are ascribed to the heel pad: shock reduction and damping. In a one dimensional approach, both properties can be revealed from the relation between force and deformation. On one hand, the slope of the force - deformation diagram, or stiffness, reflects the resistance of the heel pad during its deformation. On the other hand, the area enclosed by the loading and unloading part of the force-deformation curve, or hysteresis, is a measure of the energy absorbed by the heel pad.

In the barefoot condition, similar force - deformation relationships were gathered from "*in vivo*" impact experiments: drop tests (Kinoshita *et al.*, 1993) or pendulum tests (Aerts *et al.*, 1993; Cavanagh *et al.*, 1984; Valiant, 1984). However, in an "*in vitro*" approach (cyclic INSTRON vibration test: Bennett *et al.*, 1990; Ker *et al.*, 1989) isolated heel pad samples were found to behave in a much stiffer and more resilient way. Recently this discrepancy between "*in vivo*" and "*in vitro*" heel pad testing was resolved by exposing the same heel pad specimen to a pendulum impact and to an "adapted" INSTRON impact (Aerts *et al.*, submitted).

Little is known about the mechanical behavior of the fatty heel tissue in the shod condition. Two "*in vivo*" studies demonstrated the deformation characteristics of the fatty heel tissue to be markedly influenced by wearing athletic footwear (pendulum test: Aerts *et al.*, 1993; actual running: De Clercq *et al.*, 1994).

The purpose of this lecture is to summarize and to discuss the findings of our recent work in the field of testing heel impact mechanics.

B.Barefoot human heel pad mechanics by quantitative "*in vitro*" testing: featuring the divergence from the "*in vivo*" approach.

Review

In *in vivo* impact experiments (for instance Aerts *et al.*, 1993) the heel part of the bare foot of the test subjects was deformed by the linear *momentum* of an instrumented mass. In the former "*in vitro*" tests heel pad samples, still attached to part of the *calcaneus*, were subjected to sinusoidal vibrations with a predefined loading range and frequency (for instance Bennett *et al.*, 1990). The results of the two approaches differ in several, functionally important aspects (see figure 1). *In vivo* impact tests reveal a non-linear stiffness, which is at a load equal to body weight about 150 kN m^{-1} and a tendency toward frequency dependency (*i.e.* increasing stiffness with increasing impact velocity). *In vitro* vibrational tests (cyclic INSTRON tests) show a much higher heel pad stiffness at a load equaling body weight of about 1160 kN m^{-1} and no frequency dependency. Moreover a striking difference in energy dissipation exist. The *in vivo* impact tests show energy dissipations between 76% and 95%, being in sharp contrast with 30% of energy loss found for the isolated heel pad samples.

Purpose

Bennett and Ker (1990) explained the distinct results between the two tests by assuming additional compliance and energy loss to occur in the ankle-leg-knee system which is inevitably incorporated in the *in vivo* pendulum or drop tests.

Because it is impossible to perform the INSTRON test procedure with test subjects, the "lower leg hypothesis" was tested in an alternative *in vitro* way: pendulum impact and adapted INSTRON experiments were applied to the same isolated heel pad specimen. If the large differences in stiffness and energy loss are to be only attributed to the presence of the lower leg, both tests should show identical load - deformation curves.

Materials and methods

Four heel pads, still attached to part of the calcaneus, were isolated from lower limbs amputated for clinical reasons (see also Bennett *et al.*, 1990). These specimens were subjected to pendulum impacting (mass of 11.615 kg) with impact velocities ranging from 0.20 m s^{-1} to 0.84 m s^{-1} (see also Aerts *et al.*, 1993). In order to mimic the impact as closely as possible, the "old" full cycle INSTRON procedure (the specimen stays in contact with the load cell throughout the deformation cycle) was altered in such a way that the specimen touched the load cell halfway its sinusoïdal displacement (*i.e.* at maximal velocity of the actuator). In different trials, the deformation of the sample varied from 1.86 mm to 6.3 mm, and frequencies ranged from 1.1 to 11 Hz. For reasons of comparison with the previous INSTRON studies of Bennett and Ker (1990) full cycle tests were also performed.

In both test procedures, force deformation diagrams were obtained. In the pendulum set up only single contacts can be carried out. In the INSTRON set up, first, second,.... n'th force-deformation loops were obtained. The first loop - half cycle INSTRON test mimics a pendulum impact in the best way.

Results and discussion

(a) Mechanical changes due to adapted INSTRON tests

Subjected to full cycle tests, the mechanical behavior after

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many cycles conforms for all pads to the former results obtained by Bennett and Ker in 1990 (mean energy loss of 30.9% ± 2.7% and no frequency dependency). In contrast, first loop - half cycles tests show a significantly larger mean energy loss of 50.4% ± 7.2% and the force-deformation relationship appears to be frequency dependent. Nevertheless, from the second half cycle impact on a rapid conditioning (*i.e.* to become more resilient) occurs. At first glance, such a cyclic "half-cyle" impact regime matches the loading pattern during barefoot running in the best way. Ongoing research on this topic will be published in due time.

(b) Adapted INSTRON "first loop - half cycles" compared to pendulum impact tests (see figure 1)

When properly scaled, the force-deformation curves from pendulum and INSTRON are very similar in shape. For both testing procedures, the loading phases overlap attributing a non-linear, but identical compliance to the bare heel pad specimen. Concerning damping, the fractional energy dissipation is 65.5% ± 5.0% for pendulum impacts *versus* 50.4% ± 7.2% for INSTRON half cycles. This discrepancy is mainly caused by a small difference in the hysteresis loops at the end of the unloading phase. This is most likely due to the pendulum procedure which is, compared to a fine tuned servo hydraulic INSTRON impact, more sensitive to energy losses caused by experimental factors (for instance eccentric rebounds, accelerometry combined with a double integration procedure).

It can be concluded that pendulum and INSTRON tests reveal identical mechanics for isolated heel pads, provided that pendulum tests are performed in a proper way. This finding supports the "lower leg hypothesis". In the case of *in vivo* impact testing the presence of the lower leg most likely influences the force - deformation relationship.

C. Heel pad mechanics in the shod condition

(a) In vivo impact testing

In an earlier study by Aerts and De Clercq (1993) the effect of varying midsole hardness upon the deformation characteristics of the shod foot was investigated. Therefore the heel part of the foot of subjects (*i.e. in vivo*) was impacted by means of an instrumented pendulum (identical to the one used for the comparison between INSTRON and pendulum impacting). This means that the measured force - deformation relationships reflect the mechanical behavior of the fatty heel tissue in combination with the lower leg (see above).

The instrumented pendulum, with a mass of 11.615 kg, touched the heel with a velocity ranging from 0.37 to 1.06 m s^{-1}. This was done in four conditions: two shod conditions (running shoes only differing for midsole hardness; EVA Asker C 65 and 40), a sticked sole condition and a barefoot condition.

By changing the applied impact energy and the mechanical configuration of the shoe-heel system, it could be demonstrated that the shock reduction capacity of the latter (including the lower leg) depends upon a variety of external factors such as maximal load, loading rate, stiffness of the shoe midsole and presence and/or mechanics of the heel counter. Heel counters constructed to allow "adequate" sideways expansion without risk of mechanical overloading of the heel pad, may improve the overall shock reduction of the shoe-heel system.

(b) Cineradiographies from actual heel strike in shod and barefoot running

In this study (De Clercq *et al.*, 1994) lateral cineradiographies were taken at 150 frames/second from an actual running step (running speed equivalent to 4.5 m s^{-1}). Two male subjects ran barefoot and with running shoes. The vertical ground reaction force and the actual deformation of the fatty heel tissue underneath the *tuberculum calcaneum* were measured.

It was found that in barefoot running the heel pad deforms to a maximal percentage deformation of 60.5 ± 5.5% and in shod running only to 35.5 ± 2.5%. In both conditions the amplitude of the vertical component of the ground reaction force is equal. Therefore, it was argued that embedding the foot in a well-fitting shoe increases the effective stiffness of the heel pad.

D. General conclusions

The one dimensional mechanical behaviour of the bare heel pad can be studied in a reliable way by means of simulated impacts on isolated heel pad specimen (*i.e. in vitro*).

For the moment, all the results obtained from impact studies using subjects, incorporate the mechanical behaviour of the lower leg (*i.e. in vivo*).

When the heel is embedded in a well fitting shoe, the heel pad and various shoe components (midsole, insole and heel counter with dynamic changing mechanical characteristics) form an integrated structure with complex mechanical characteristics. The studies discussed clearly reduce the problem to one dimension. A two or even three dimensional stress strain approach might be appropriate.

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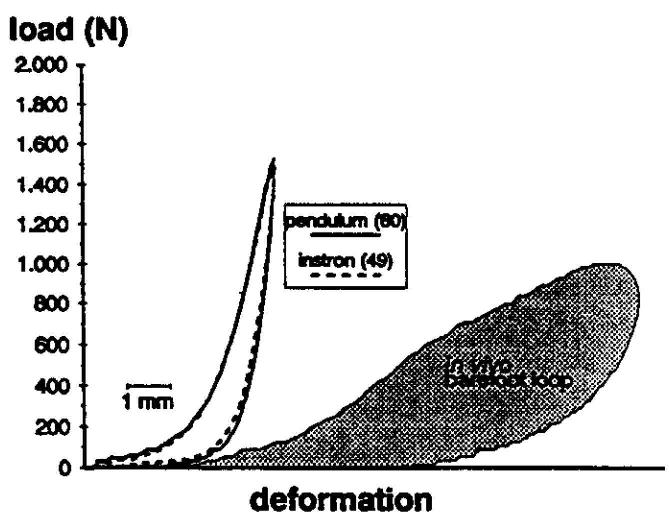

Figure 1: Superposition of a representative INSTRON "first loop - half cycle" and a pendulum load-deformation loop with indication of the percentages of energy loss. The shaded area represent an *in vivo* barefoot loop.

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FATIGUE AND LOWER EXTREMITY FUNCTION

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INTRODUCTION

The development and the research of functional footwear is usually focussed on the concept of "prevention of excessive load" (Nigg and Segesser, 1992). Impact force cushioning at touchdown, rearfoot motion control and guidance of the foot during the pushoff have been treated as major requirements of a technically functional shoe.

Many authors have studied the absolute and relative motion of the rearfoot during running and sideward loading in order to quantify stability requirements of sport shoes (eg. Luethi et al., 1986, Stacoff et al., 1993). Others have described the ground reaction forces at touchdown (eg. Cavanagh and Lafortune, 1980) maximum plantar insole pressure (eg. Hennig et al., 1982) or maximum tibia acceleration (eg. Hennig and Lafortune, 1991). Tests and experiments were executed with new and worn shoes in order to examine material fatigue (eg. Swigard et al., 1993). The interaction of the shoe material and biological structures of foot and leg was studied by Cavanagh et al. (1984) and Shorten (1993). Differences have been shown between results of testing restricted to materials and testing of the same materials conducted with subjects (Kaelin et al., 1985).

We know of no studies dealing with mechanics of the shoe - leg interface taking into account the muscular and neuromuscular control system of the human being. The micro-control of foot motion and its influence on lower extremity function at varying stages of fatigue is not well understood.

PROCEDURES

A series of experiments was designed to determine the influence of muscular and cardiovascular (aerobic) fatigue on lower extremity function during running and related activities. These experiments could be placed into three categories: a) lower extremity function prior to and during aerobic fatigue without fatigue of the local musculature, b) lower extremity function prior to and during local muscular fatigue, c) lower extremity function prior to and during aerobic and local fatigue.

The direct effect of aerobic fatigue on the control mechanisms of lower extremity function was determined by a dynamic leg press test in which the three orthogonal reaction forces at the foot - machine interface were measured. In this specific experiment the subjects were fatigued with a hand ergometer.

The kinematics and kinetics of healthy subjects running on an instrumented treadmill were also studied before, during and after a well defined fatigue treatment. Local muscular fatigue was achieved by a foot musculature training device whereas the combined fatigue effect (aerobic and muscular) was produced by running for 45 minutes at 3.5m/s. EMG patterns (m. triceps surae, m. tibialis anterior and m. peroneus longus) and plantar pressure distribution were additionally quantified.

RESULTS AND DISCUSSION

The kinetic response on lower extremity function under cardiovascular fatigue without local muscular fatigue showed a significant pertubation in relation to non-fatigue activity. No significant ($p < 0.01$) changes in the maximum force and force patterns could be determined in the mechanically dictated direction (leg press) before and after treatment. However, significant increases were measured in the lateral/medial and anterior/posterior directions. These results indicated an influence of the cardiovascular status on the efficiency of the neuromuscular control mechanisms.

The experiment in which subjects were compared while running with and without local fatigue of the lower leg muscles indicated a less controlled touchdown with great variability in kinematic and EMG patterns. In addition the angular velocity of pronation in the early stance phase increased.

The subjects that were fatigued while running showed remarkable lower extremity kinematic and kinetic characteristics. The general trend for one subject is illustrated in figure 1. This shows four major factors quantified during the run:

a) an unexpected decrease in the insole impact peak under the heel,

b) an increase in the angle of pronation (achilles tendon angle) with fatigue,

c) the standard deviations of these parameters increased with running time and

d) a left or right shift of these curves could be observed depending upon the physical fitness of the subjects.

Muscular and aerobic fatigue play an important role in the control of lower extremity function. The efficiency of the lower extremity movement is reduced by fatigue, thus resulting in less controlled touchdown and stance phases. This implies a greater demand on the stability requirements of the shoe for fatigued subjects. The question therefore arises whether functional tests of sports footware should include the special condition of fatigued subjects rather than using only unfatigued people to test specific requirements. This could advance the concepts that have previously been formulated by studies on excessive loads during sport.

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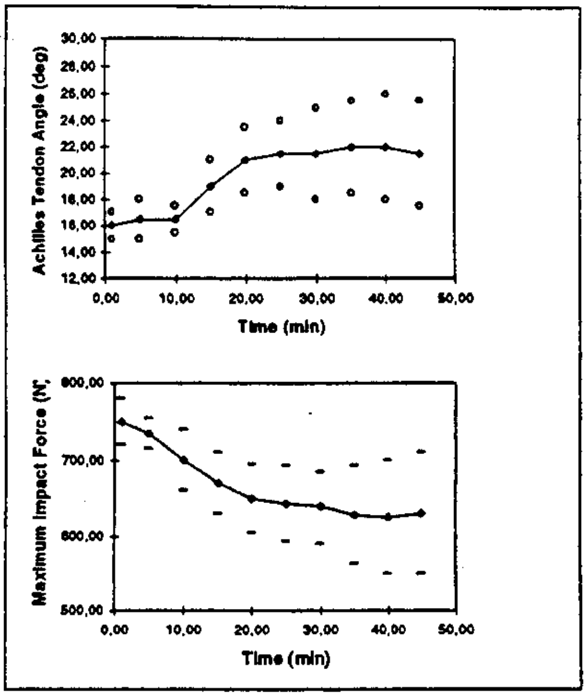

Figure 1: Maximum achilles tendon angle and maximum insole impact peak forces under the heel (mean and standard deviation of 20 steps of one subject measured at five minute intervals).

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PLANTAR PRESSURE MEASUREMENTS AND APPLICATIONS TO FOOTWEAR

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INTRODUCTION

For more than 100 years researchers have been interested in the distribution of loads under the human foot during various activities. Early methods estimated plantar forces from impressions of the foot in plaster of Paris and clay. Later techniques included optical methods with cinematographic recording. During recent years, the availability of inexpensive force transducers and modern data acquisition systems have made the construction of various pressure distribution measuring systems possible.

SENSORS

Today, the major transducer technologies for pressure distribution devices are based on capacitive, piezoelectric and resistive principles. All methods are based on the effect of changes in the electrical properties of sensors, caused by the mechanical deformation of its material. Therefore, the elastic response of material deformation plays a major role for the quality of a transducer. Reduced accuracy, as it is frequently seen in resistive transducers, is mainly caused by a lack of material elasticity of the resistive layer between the electrodes (e.g., conductive paint). Resistive sensors are often functioning as contact sensors, showing reduced electrical resistance when the surface contact area of the electrodes is increased. Because of an insufficient elastic recoil in a contact sensor, they often demonstrate large hysteresis. Contact sensors often behave differently on the kind of interface they are used on. On rigid surfaces, a reduced contact surface area will cause lower output signals as compared to measurements on a soft interface. Because resistive sensors are inexpensive, pressure distribution sensors with a high spatial resolution can be offered. These systems may not have the required accuracy for research applications but are often sufficient for a clinical assessment of foot loading behavior. Pressure distribution devices on a capacitive basis use the compression of a viscoelastic dielectric to achieve changes in capacitance (Nicol & Hennig, 1976). Large relative deformations of the transducers are necessary to achieve sufficient resolutions for low forces. Depending on the elastic properties of the dielectric, recoil of the material will require time and thus introduces hysteresis effects for rapid loads. Solid rubber pyramids mats as compared to closed cell foam layers offer a faster response. Using this technology, capacitive pressure mats have a good accuracy for dynamic loads under the foot in walking and running. Piezoceramic materials show very small and highly elastic deformations. For rapid loads, these transducers combine good linearity and a very low hysteresis (Hennig et al., 1982). However, piezoelectric sensors have a semistatic response, because volume conduction through the material will discharge the generated electricity. Most piezoceramic materials also exhibit pyroelectric properties and will create electrical charges with changes in temperature. Therefore, for these transducers, thermal insulation or a temperature equilibrium, as it is normally present inside of shoes, is necessary. Piezoceramic transducers have successfully been used for applications ranging from measuring the pressures under diabetic feet (Cavanagh et al., 1985) to analyzing the foot to ground interaction during running and jumping.

SINGLE TRANSDUCER VERSUS SENSOR MATRIX ARRANGEMENTS

Because footwear influences the foot to ground interaction considerably, in-shoe pressure measurements are of special interest. Insoles with a transducer matrix or single transducers, placed under specific anatomical locations, can be used for plantar in-shoe measurements. Pressure insoles with a matrix shaped sensor arrangement can easily be placed inside shoes. Depending on the number of sensors, a more or less detailed picture of the foot to shoe interaction during load bearing can be created. Because most feet show individual differences in geometry, the exact placements of the sensors under the anatomical sites of interest are not known. As shown in figure 1, this can result in a substantial underestimation of the pressures under small bony structures (e.g., metatarsal head V). In situation B (figure 1) the sensors will only measure 25% of the actual peak pressures.

A B

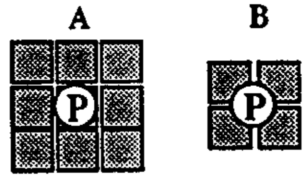

Figure 1: Localized pressure point (P) on top of one (A) and four (B) sensors.

Depending on the construction features of different shoe constructions, the placement of the insole matrix may have a slightly different position under the foot. In the worst case, situation A may occur in one type of shoe, and situation B in another product. Pressure insoles with a large number of small transducers, offering a high spatial resolution, are necessary to overcome this problem. However, pressure mats with a high density of accurate sensors are expensive. Many sensors will also restrict the time resolution of scanning and will generate large amounts of data.

Single transducers are typically fixed with adhesive tape under palpated anatomical locations. Once, the transducers are placed under the known anatomical sites, the influence of footwear can be examined. The major advantage of this technique is an exact placement of the sensors under the foot structures of interest, independent of individual foot geometry. It is also guaranteed that the sensor locations under the foot will be identical in different footwear. The typical locations for single sensors

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under the foot are: medial and lateral heel and midfoot, the metatarsal heads, and the toes. The use of single sensors does not give a complete picture of the foot to shoe interaction. However, figure 2 demonstrates that the vertical ground reaction force curve during running can be reproduced, using the summed pressures of only 8 sensors under the foot. The chosen plantar sensor locations were medial and lateral heel and midfoot, the metatarsal heads I, III, and V, and the hallux. The data from 22 subjects, running in one type of footwear at a speed of 3.3 m/s, were used for the comparison in figure 2.

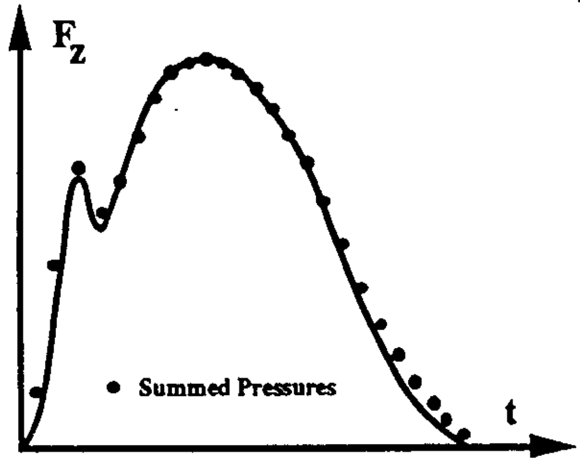

Figure 2: Vertical GRF signal in comparison to the summed pressures of eight single transducers

Mechanical interaction between the measuring insole and the shoe construction will occur with all measuring systems. This will lead to a different viscoelastic behavior of shoe sole properties. Even a thin sheet of plastic foil on top of a soft shoe sole will lead to a distribution of local loads. Single transducers only cover a small portion of the shoe sole and should influence the mechanical behavior of the shoe construction to a lesser degree. However, they will create unevenness of the surface and will overestimate typical pressures. Small thickness and dimensions of single transducers are desirable to reduce this interface problem. There has been a concern that single transducers in a shoe will create discomfort and cause an altered running or jumping style. Using single piezoceramic transducers of small dimensions, subjects reported no discomfort for such high load disciplines as triple jump, long jump, and high jump (Milani & Hennig, 1994). Most subjects did not even feel the sensors under their feet during the activity.

APPLICATIONS

The following example demonstrates the value of in-shoe pressure in comparison to vertical ground reaction force measurements. Twenty-two subjects with normal feet performed rearfoot strike running trials in two different types of commercially available running shoes (A & B) at a speed of 3.3 m/s. For the first vertical ground reaction force maximum (A =1.66 bw; B = 1.67 bw) and the vertical force maximum during mid stance (A = 2.66 bw; B = 2.63 bw) no significant between shoe differences were found. The peak pressures were determined using single piezoceramic transducers under the lateral and medial heel and midfoot (LH, MH, LM, MM), three metatarsal heads (M5, M3, M1), and the hallux (H). With the exception of M5 and H the peak pressures in the two shoes were found to be significantly different at all other anatomical locations ($p<0.05$ for MH; $p<0.01$ for all other sites).

	LH	MH	LM	MM	M5	M3	M1	H
A	873	766	464	228	413	475	955	532
B	1038	825	356	169	396	530	800	535

Table 1: Peak pressures (kPa) during running in two different running shoe constructions A & B.

Along with the determination of the peak pressures relative load analyses have been shown to be particularly useful. A regional impulse is determined by integrating the local forces under the specific anatomical landmarks throughout foot contact. The relative loads are calculated as percentages, dividing the local impulses by the sum of all impulse values. Table 2 shows differences in the foot loads of 22 subjects, running in two different types of footwear. During push-off, the medial load under the first ray (M1 & H) in shoe C is substantially higher, whereas shoe D demonstrates an increased contribution of the third and fifth metatarsal head structures. Considering these large differences in forefoot loading, it should be no surprise that footwear may have an influence on the occurrence of overuse injuries.

	LH	MH	LM	MM	M5	M3	M1	H
C	12.4	11.9	8.8	3.2	9.1	11.6	25.3	17.6
D	13	12.8	7.8	3.2	12.5	14.3	21.6	14.8

Table 2: Relative loads (%) during running in two different running shoe constructions C & D.

In-shoe pressure distribution measurements can also be used to study the loss of midsole cushioning with use. Nineteen running shoes from different manufacturers, worn for a distance of 10 km by 22 runners, were tested after 220 km against 19 corresponding new pairs of shoes. After use, average increases in peak pressures of 8.5% were found in the heel and 8.8% under the forefoot. For one particular shoe product, the loss of cushioning was very high, showing pressure increases of 21% in the heel and 13% in the forefoot. In-shoe pressure distribution measurements have also successfully been used for the determination of foot loads at high impact sport disciplines, for the understanding of foot mechanics during bicycling, and for the evaluation of shoe orthotics.

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FOOT PRESSURE PATTERNS IN CHILDREN AND INFANTS

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INTRODUCTION

Developmental acquisition of various motor skills in children has been of interest to researchers for many years. Developmental gait patterns in children have been studied by a number of researchers. In the most comprehensive study to date, Sutherland et al [1] studied kinematic and kinetic patterns of gait in children learning to walk using modern cinematographic and computer techniques. The dynamic distribution of the forces and pressures under the weight-bearing foot and its importance in both normal and pathological gait has been of great interest to orthopaedists and other clinicians specializing in foot and gait disorders. Reported differences between developing and mature gait, as well as between children and adult foot structure[2], suggest that differences may exist in dynamic foot pressure patterns.

The present study sought to identify and measure dynamic foot pressure patterns in normal infants and children during walking, in order to establish a normative database to provide an accurate basis for comparison to other research populations.

METHODOLOGY

Infants and children were recruited for a data collection session at the gait analysis laboratory in the following age intervals; 1-2 years, 3-4 years and 4-5 years (cross-sectional component). The 1-2 year olds repeated data collection at 6 and 12 month intervals from the date of their original study (longitudinal component). The 4-5 year olds also returned for a second session at a single 6 month interval. Foot pressure data was collected at 50 Hz with an EMED-SF pressure platform.

For each subject at each session, five (5) trials of complete foot pressure data were analyzed. For each trial, 12 plantar regions of interest (Figure 1) were identified and analyzed for; peak pressure magnitude, pressure duration, pressure onset, and pressure offset. Stance phase foot positions were assessed qualitatively by a trained gait lab technician and classified as: 1. pronated, supinated or neutral; 2. toed in, toed out or straight; and 3. heel strike, full foot strike, or forefoot strike. Summary statistics for pressure data were computed using representative trials. Statistics for timing variables were calculated using only trials with pressure in the region. Ratios were also calculated between different regions of interest and appropriate statistics were computed with a significance level of $P<.05$.

RESULTS

Space limitations do not facilitate presentation of all the data; however, a general overview is provided for the purposes of this abstract.

168 infants were studied (93 males, 75 females). Age, age at onset of walking, height, weight and foot length had means of 16.9 months, 11.7 months, 31.0 inches, 24.3 lbs. and 4.9 inches respectively. Significant differences were detected in foot pressure patterns in comparison to adult patterns. Infants displayed a more medial pressure progression with earlier onsets of the 1st MT and 1st Toe versus lateral MTs as is typical of the adult pattern. Highest peak pressure for the metatarsal heads (MT) occurred at the 1st MT which was also atypical from adult loading patterns. At the 6 month follow-up 89 subjects returned to the lab for a data collection session. Slight increases in peak pressure magnitudes were noted, while the pressure pattern remained similar to the initial session. Seventy two (72) subjects returned for a second follow-up session 12 months after the original study. Again slight differences in peak pressure magnitude and timings were observed. Fifty three (53) subjects aged 3-4 years were analyzed. The heel demonstrates the overall highest peak pressure, while the 2nd MT had the highest peak for the metatarsal region. High pressures are still visible at the 1st MT, while pressures at the 5th MT and lateral side of the foot are still relatively low. Sixty two (62) 4-5 year old subjects were analyzed and demonstrated foot pressure patterns similar to the adult pattern. Forty three (43) 4-5 year old subjects returned for a second visit at a 6 month interval with only minimal differences detected.

DISCUSSION

In summary, significant differences existed in infant foot pressure patterns from adult patterns. These differences have been quantified in the present study. A steady progression seems to occur in the foot pressure regions, timings and

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magnitudes, until approximately age 4 (Figure 2), when the pattern appears similar to the foot pressure patterns of normal adults, but with smaller magnitudes of peak pressure in the regions. Some foot positions were correlated with specific foot pressure patterns, until the foot assumed a more neutral position as observed in the 4 year olds. Therefore, foot pressure patterns in children appear to progress through a developmental pattern similar to other neuromuscular skills. However, a mature pattern seems to occur at approximately 4 years old, slightly ahead of the onset of the mature gait pattern. This data was used by Stride Rite to develop a new line of infant shoes. In-shoe pressure patterns were measured and compared with barefoot pressure patterns to assist in the development process. Brief examples of these pressure measurements and the biomechanical principles which were incorporated into the final footwear designs will also be presented.

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The authors are indebted to the staff of the Gait Analysis Laboratory . This research was funded by the Stride Rite Corporation, Cambridge, Ma.

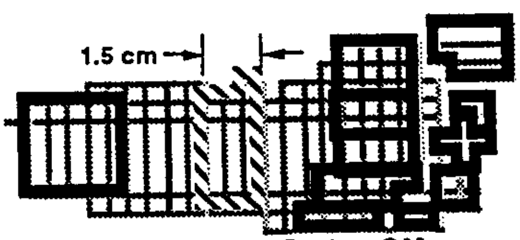

Figure 1. Twelve Plantar Regions Of Interest

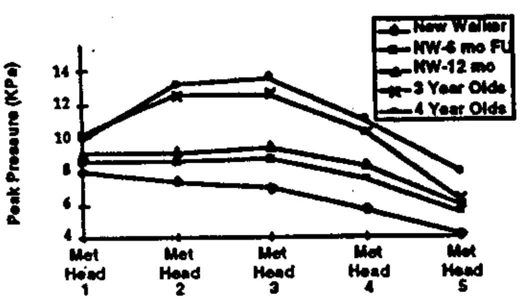

Figure 2. Metatarsal Head Peak Pressures By Age

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An Instrumented Linkage to Measure the Level of Support Provided to the Ankle Joint by High-top Athletic Footwear

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INTRODUCTION

High-top footwear which cross the ankle joint include basketball shoes, hiking shoes, military boots, etc. One major goal of these shoes, as well as of other external ankle support systems such as ankle braces and tapes, is to support the ankle as the foot moves over uneven ground or is involved in jumps and other fast lateral movements. (Bauer, 1970; Robinson, 1986). Optimizing the support characteristics of a high-top shoe is a complex task since often increasing the stiffness of the shoe results in decreased ankle mobility which in turn interferes with athletic performance. Therefore, the ability to quantify the amount of support provided to the ankle by a high-top shoe can be of significant value in assisting in the design of improved high-top shoes.

We developed a six-degrees-of-freedom instrumented linkage capable of measuring the rotatory and translatory stiffness characteristics of the ankle-shoe complex and which can be used to quantify the support (i.e. - increased stiffness) provided to the ankle by the high-top shoe. We conducted preliminary tests with this device on a number of high-top shoes including basketball shoes and military flight boots and found the device capable of identifying the support characteristics provided by the different shoes.

REVIEW AND THEORY

In previous studies (Siegler et al., 1988), an analytical and experimental technique was described to determine the three dimensional, six-degrees-of-freedom stiffness characteristics of the ankle joint complex. This technique was based on calculating, from experimental data, the stiffness coefficients of the ankle joint using the Grood and Suntay parameters (Grood and Suntay, 1983). The technique was applied to study the effect of ligament injuries (Siegler et. al, 1990) as well as to study the support characteristics of some athletic footwear (Siegler and Black, 1991). The experimental technique consisted of applying loads across the ankle joint through a set of pneumatic actuators and measuring the resulting joint motion using an Opto-electric kinematic data acquisition system. This system proved to be too time consuming and cumbersome for extensive in-vivo testing and furthermore it required complex and time consuming data processing. To overcome this problem, a six-degrees-of -freedom instrumented linkage, shown schematically in Figure 2, was developed. This system allowed application of forces and moments along and about the Grood and Suntay parameters axes of the ankle, shown in Figure 1, while measuring the corresponding translations and rotations along and about these axes. The simplicity of this device, both in terms of its alignment to the ankle as well as in its operation, and the lack of complex data processing make this device particularly suitable for in-vivo studies such as the one which was required in the present work.

EXPERIMENTAL PROCEDURE

Subjects: Young male volunteers ranging in age between 20 and 40 years of age with size 9(USA) feet were tested. None of these subjects had a history of severe ankle injuries such as fractures or severe sprains. All were involved in recreational sports.

Materials: Two types of high-top shoes were tested in this study. The first was a high-top basketball shoe equipped with inflatable air pockets. These pockets, located in the lining of the shoe surrounding the ankle, could be inflated to various pressures up to 9 psi. A pressure gage was used to set the pressure within these pockets. The second type of shoe was a high-top standard military flight boot.

Testing procedure:

The first group of subjects, were tested with the military flight boot and the second group of five subjects were tested with the high-top basketball shoe. The first group were tested on both limbs both barefoot and with the military boot on. Each test was repeated twice. The second group was tested with the high-top basketball shoe on under two different conditions. First, with no pressure in the air chambers and the second time with the air chambers inflated to 6 psi.

The test was conducted as follows. First, the tips of the medial and lateral maleolii were marked on the skin and then on the outside of the corresponding high-top shoe. The subject was then seated in a semi-reclining position and his tibia was secure to the device. The foot or the shoe was then attached to a foot plate in the device. In this position, the foot was fixed at neutral (i.e. - 90 degrees). In this position, several loading unloading cycles were performed in the direction of inversion/eversion and in the direction of internal rotation/external rotation. The foot was then set at 30 degrees of plantarflexion and the test repeated in this position. During the test, the subject was instructed to relax his leg muscles.

RESULTS:

Typical data obtained with the military flight boot is shown in Figure 3. This figure shows a comparison between the inversion/eversion and internal rotation/external rotation stiffness characteristics of the bare ankle and the stiffness characteristics of the ankle when wearing the military boot. In Figure 3a, the inversion/eversion angle is plotted as a function of the applied inversion/eversion torque for the two cases and in Figure 3b the internal rotation/ external rotation angle is plotted against the applied internal/external torque.

Typical data showing the effect of inflating the air chambers in the high-top basketball shoe is shown in Figure 4. This figure compares the inversion/eversion (Figure 4a) and internal rotation/external rotation (Figure 4b) stiffness of the ankle joint complex before and after inflating the air chamber to a pressure of 6 psi.

DISCUSSION:

The results obtained with the high-top basketball shoe suggest that the presence of inflated air chambers in the lining of the shoe surrounding the ankle increase the support to the ankle primarily in inversion, eversion and internal rotation. External rotation appeared to be less effected by inflating the air chambers. The military flight boot was found to increase the support to the ankle joint complex in inversion/eversion and internal rotation/external rotation.

The six-degrees-of-freedom instrumented linkage presented in this study may be a valuable tool in quantifying the support characteristics of various high-top shoes and may be used to assist in designing improved high-top shoes.

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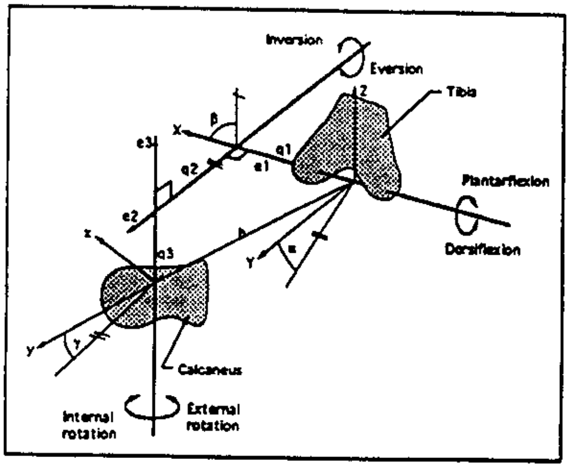

Figure 1 - Definition of the Grood & Suntay parameters for the ankle joint complex. e1 - axis fixed in the tibia. Rotation about this axis - a corresponds to dorsiflexion/plantarflexion and translation - q1 correspond to lateral translation. e2 - floating axis mutually perpendicular to e1 and e3. Rotation - b correspond to inversion/eversion and q2 - correspond to anterior/posterior drawer. e3 - axis fixed in the calcaneus. rotation - g correspond to intenal/external rotation and q3 correspond to distraction/compression.

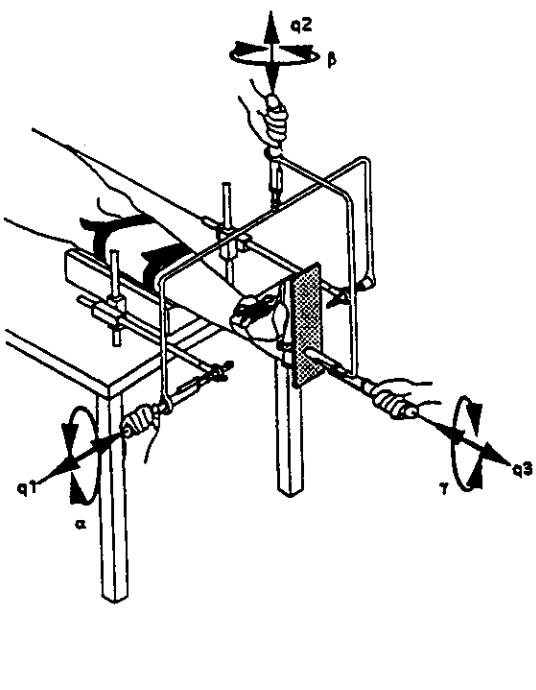

Figure 2 - Schematic view of the six-degrees-of-freedom instrumented linkage used to measure the load-dispalcement characteristics of the ankle joint complex along and about the Grood & Suntay parameter axes.

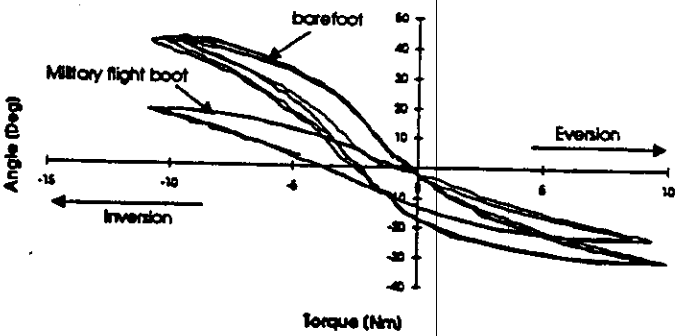

Figure 3a

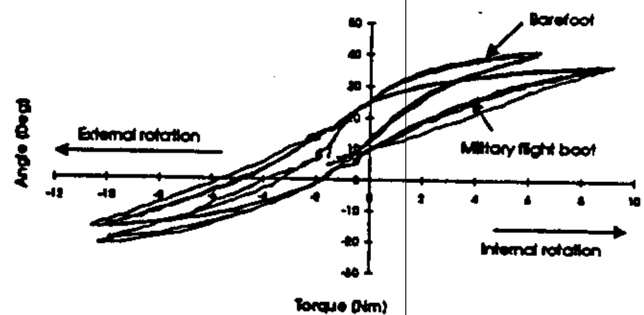

Figure 3b

Figure 3 - Data obtained from subject JDM comparing the load-displacement characteristics of the ankle barefoot to those when wearing a military flight boot. Each curve was obtained from five loading - unloading cycles.

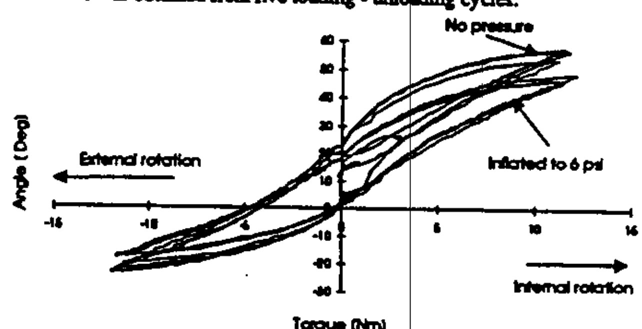

Figure 4a

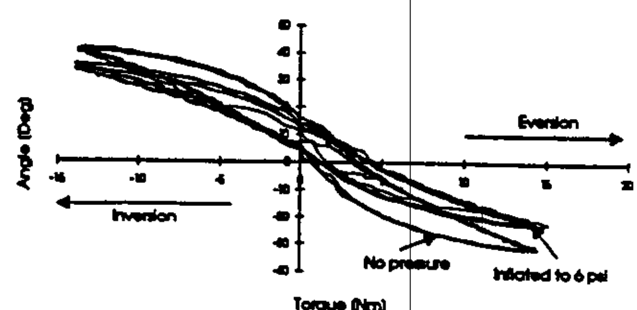

Figure 4b

Figure 4 - Data obtained from subject SS comparing the load-displacement characteristics of the ankle joint before and after inflating the air chambers in the basketball shoe to a pressure of 6 psi.

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STABILITY AND REARFOOT MOTION TESTING: A PROPOSED STANDARD

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INTRODUCTION

The factors associated with athletic shoe rearfoot stability are extremely complex. To evaluate the stability of athletic shoes, rearfoot motion measures have been used. Rearfoot motion is equated with the pronation/supination actions of the sub-talar joint and refers to the relative motion of the foot and leg that occurs most rapidly during the first 100 ms of the support phase of the running stride.

The pronation motion of the sub-talar joint is part of the body's natural shock attenuation mechanism. Therefore, a certain amount of pronation is desirable. However, excessive pronation of the sub-talar joint may lead to the development of lower extremity injuries. The construction of athletic shoes to provide control of excessive pronation can help to reduce the risk of injury.

The necessity of standards for testing rearfoot stability in the footwear industry is obvious. In 1992, the ASTM sub-committee F08.54 established a working committee to recommend standards for the testing of stability of athletic footwear. The remainder of this paper deals with the recommendations of this sub-committee for the establishment of such standards. It should be noted that these proposed standards pertain only to two-dimensional rearfoot motion and do not address three-dimensional rearfoot motion.

SUBJECT RECRUITMENT

When subjects are recruited for a study pertaining to rearfoot stability, it is suggested that the subject undergo a complete static lower extremity evaluation. The evaluation should be conducted by clinically trained personnel. This evaluation will provide a profile of potential subjects and should include a history of lower extremity injury, foot type, forefoot alignment, rearfoot alignment, tibial rotation, and the range of motion of the sub-talar joint. The preferred running footfall pattern should also be determined. In addition, a training history of the subject should be recorded. This history should include particular training habits, miles per week, training pace, and race pace. The lower extremity evaluation and training history could be used for inclusion or exclusion of subjects from a study.

If the stability test is to be carried out on a treadmill, the subjects should be experienced treadmill runners. If the subjects are not experienced treadmill runners, a minimum of one 20-minute period should be held prior to the day of data collection. During this initial treadmill run, the subject should be started at a slow pace with the speed gradually increasing until it is at the test speed. The duration and number of practice sessions will depend on the comfort of the subject with treadmill running.

DATA COLLECTION

The initial concern for collecting rearfoot motion data is the frame rate of the video/cine camera. The sampling frequency for any data collection is based on the highest frequency in the signal. In papers by Williams et al. (1991) and Hamill et al. (1992) the highest frequency in the signal was 15 and 16 Hz respectively. Further, in unpublished work completed in our laboratory, the highest frequency in rearfoot motion data was found to be 18 Hz. The Nyquist theorem suggests that the sampling frequency (ω) be at least two times the highest frequency (Winter, 1991). However, by sampling at exactly 2ω, signal aliasing becomes a severe problem. It is therefore suggested that the sampling frequency for rearfoot data collection be at least 5ω (Oppenheim, Willsky, and Young, 1983). For rearfoot motion, therefore, a sampling frequency of at least 90 Hz is recommended.

Figure 1 illustrates the differences in rearfoot data curves of the same footfall digitized at sampling frequencies of 200, 100, and 50 Hz. There were no differences in the curves digitized at 100 and 200 Hz. These two curves are in fact overlayed in Figure 1. However, the rearfoot angle varied from -15° at a 200 Hz sampling frequency to -13° at 50 Hz. The rearfoot angular velocity also varied from -585 °/s at 200 Hz to -368 °/s at 50 Hz. This example illustrates the differences that can be obtained simply by virtue of varying the sampling frequency.

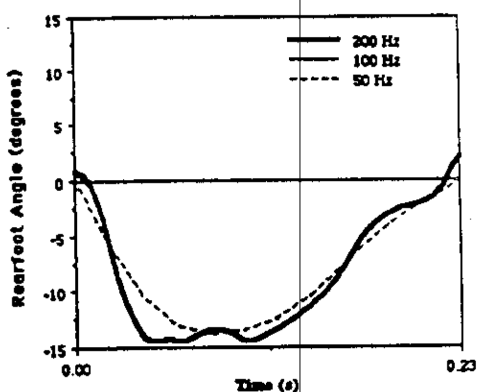

Figure 1. A single rearfoot angle curve generated from data digitized at sampling frequencies of 200, 100, and 50 Hz.

In preparing the subjects for the data collection session, it is necessary to accurately and objectively place the markers on the leg and the shoe. The protocol for marker placement established by Clarke et al. (1984) is recommended. The use of a "jig" as suggested by Clarke and associates is important to remove the experimenter bias in the placement of markers.

Clarke et al. (1984) also suggested a standing position to determine the calibration angle once the markers have been placed. A "jig" in which the subject stands with their feet abducted to 7° with their heels 5 cm apart should be used to standardize the calibration position. From this static posture, the heel angle is calculated. The heel calibration angle can then be subtracted from all heel angles obtained dynamically. This technique scales the heel angle to the vertical during static weight bearing.

In the data collection session, all subjects should have a warm-up run of approximately 2 to 5 minutes prior to the actual collection of the data. This run should be consistent throughout the study. The warm-up time should be given in each shoe condition to allow the subject to accommodate to the shoe. During the data collection run, the running pace for all subjects should be held constant throughout all footwear conditions.

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Close attention should be paid to the experimental design of the study. There are three considerations in this regard. First, the number of subjects should be determined using a sample size estimation procedure (Levy and Lemeshow, 1980). Second, the shoes (conditions) tested should be presented in a balanced order to minimize an order effect. This also requires that the number of subject be at least two times the number of conditions. Lastly, an equal number of footfalls (i.e. trials) for each subject should be analyzed. If these three items are taken into consideration, the study should have sufficient statistical power from which to make valid conclusions.

DATA ANALYSIS

Upon digitizing the video/cine data, decisions concerning the cut-off frequency for digital filtering must be made. As mentioned previously, both Williams et al. and Hamill et al. suggested cut-off frequencies of 15 and 16 Hz respectively. From unpublished data collected in our laboratory, the cut-off frequency for rearfoot data was found to be 18 Hz. A cut-off frequency below 15 Hz will attenuate the maxima of the data set and could cause the experimenter to draw incorrect conclusions. In Figure 2, a single trial, sampled at 100 Hz, is presented with cut-off frequencies at 12 Hz and 18 Hz. At 12 Hz, the high frequency components of the signal are severely attenuated. Therefore, it is our recommendation that cut-off frequencies between 15 and 18 Hz be used.

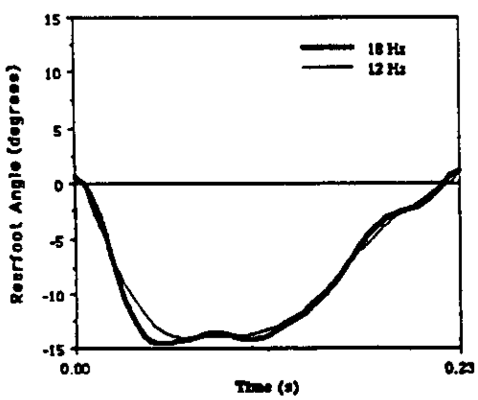

Figure 2. A single rearfoot angle trial that was digitized at a sampling frequency of 100 Hz and filtered with cut-off frequencies of 18 Hz and 12 Hz.

There are several angle conventions that are evident in the literature concerning rearfoot motion (Bates et al., 1979; Nigg et al., 1980; Clarke et al., 1983). While all of these conventions ultimately provide the same information, the committee decided that the most prevalently used convention was that of Clarke et al. (1983). Thus, the recommendation is that the rearfoot angle should oscillate about 0^{o} (i.e. neutral) with positive angles designating supination and negative angles designating pronation.

There are a number of variables that can be generated from a single rearfoot-time curve. However, many of these parameters occur when the leg is out of plane and thus the calculated angles may not be an accurate representation of the actual event. Areblad et al. (1990) illustrated the difference between two-dimensional and three-dimensional rearfoot angles. At maximum pronation, the 2-D rearfoot angle is approximately equal to the 3-D angle. For this reason, this committee urges reporting only events concerned with maximum pronation angle. These parameters include: 1) maximum pronation angle; 2) maximum leg angle; 3) maximum heel angle; and 4) time to maximum pronation angle. Other parameters may be calculated but it should be understood that these parameters may be influenced by the 2-D nature of the analysis.

The suggested statistical protocol for rearfoot motion studies involves a repeated measures ANOVA with at least two repeated factors, conditions (shoes) and trials. This analysis provides information on the generalizability of the results. In addition, analysis of individual subject trends is recommended to determine if "out-lyers" have disproportionately affected the data.

CONCLUSIONS

The standardization procedure for the collection and analysis of rearfoot data that is presented here only concerns two-dimensional analyses. Issues regarding subject selection, sampling frequency, data collection, and data analysis are presented. The protocols used by several laboratories have been examined in the development of a proposed standard procedure for rearfoot stability testing. Standardization of testing within the industry will result in consistent methods being used and comparisons of results to be made where appropriate.

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EVALUATING OUTSOLE TRACTION OF FOOTWEAR

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Introduction

Sufficient traction is a required characteristic of athletic footwear for the development of high horizontal forces against the playing surface for fast running, quick starts and stops, rapid changes in running direction, etc. High traction may be associated with a higher risk of injury, however, due to excessive foot fixation. In the workplace, or when worn for casual wear, footwear outsoles require sufficient traction for the prevention of slips and falls. Since outsole traction provides so many useful functions, including the basic provision for the horizontal force development necessary for most forms of locomotion, and since outsole traction is an empirical quantity which does not follow the classical relations of Coulomb friction (van Gheluwe et al., 1983; Schlaepfer et al., 1983; Valiant, 1990), there is a need for its accurate evaluation.

Identification of Evaluation Methods

Subject Tests

Nigg et al. (1987) imply that physical traction tests are inappropriate for assessing frictional forces developed during common sports activities. They emphasize that tests for traction, particularly rotational traction, should involve subjects and the measurement of loading on the locomotor system during actual human movement. Stucke et al. (1984) also employ multiple trials of human subjects performing actual movements on playing surfaces mounted on a force platform to quantify traction characteristics. Observing kinematic adjustments as a function of the frictional characteristics of playing surfaces demonstrates the need including human movement in the quantification of footwear outsole traction.

However, there are descriptions of many different physical traction testing methodologies and their applications to different sports or to different types of footwear. Both approaches have their merits and limitations.

Mechanical Tests

Bonstingl et al. (1975) and Rheinstein et al. (1978) described a physical traction testing device, instrumented with a strain guaged dynamometer, which measured rotational traction characteristics as a weighted pendulum struck a loaded foot form to which was fitted a shoe. A hydraulic piston was used to drag the shoe outsole linearly on the test surface to measure friction coefficients. Bonstingl et al. (1975) reported that frictional resistance to physically rotated cleated shoe outsoles was highly related to effective cleat surface area. Rheinstein et al. (1978) showed that resistance to rotation about a vertical axis decreased as the hardness of the outsole increased.

van Gheluwe et al. (1983) described a similar traction testing device in which a shoe was pushed or rotated along a force platform mounted playing surface at relatively fast speeds by a striking pendulum. Friction was greater when a larger portion of the shoe outsole contacted the playing surface. A dependence of coefficient of friction on normal load was also reported.

Schlaepfer et al. (1983) also used force platform signals to calculate static and kinetic coefficients of friction. Shoe outsoles weighted with Plaster of Paris and attached to an independent cart were pulled at various velocities along a playing surface mounted on the force platform. Coefficients of friction of tennis shoe outsoles on an artificial playing surface were dependent on normal force. Chapman et al. (1991) used a similar approach to drag court shoes filled with plaster across surfaces mounted on a force platform to measure coefficients of limiting friction on different dry, wet, and dusty surfaces.

Andréasson et al. (1986) measured the friction forces and torque developed by shoe outsoles on specific playing surfaces with a rotating platform and a strain guaged loading frame. The apparatus can accommodate many different orientations of the shoe with respect to the playing surface as well as different normal loads and relative sliding velocities. They reported a slight relationship between the coefficients of friction of different shoe outsoles on artificial turf and torque developed during rotation.

Winkelmolen et al. (1991) used a custom mechanical apparatus to measure translational and rotational friction coefficients of shoes.

Considerable investigations on the characteristic of slip resistance of walkway surfaces and shoe outsole materials have been carried out under the jurisdiction of organizations such as the American Society of Testing and Materials, Underwriters Laboratories, National Bureau of Standards, and Occupational Safety and Health Agency. The James Machine, an articulated strut which measures the critical angle at which loaded outsole specimens slip on a test surface (James, 1944) has historically been used for determining safe walkway surface materials, floor polishes, and to defend such products from liability. Several portable and non-portable testers of walkway surfaces have since been developed, primarily for identifying slip resistance in workplaces (Irvine, 1967; Brungraber et al., 1978; Meserlian, 1993).

Description of a Mechanical Traction Testing Device

Yet another mechanical traction testing device that evaluates both translational and rotational traction during closely simulated in-use loading conditions is sketched in Figure 1. It can also be used as a research tool to investigate the unique physics of friction of footwear polymers. A shoe with the outsole to be tested is fitted over a foot form. Pinned to the foot form in a hole normal to its bottom surface is a shaft supporting a stack of weights so that tests can be made at normal loads specific to the sport for which the shoe is intended. Interchangeable sections of applicable playing surfaces can be firmly bolted to a force measuring platform.

In the rotation test, an actuator rotates a shoe on the playing surface 2.2 radians about a vertical axis passing between the first two metatarsal heads. Angular velocity is adjustable and calculated from a sampled potentiometer signal. The frictional resistance to rotation of the outsole is quantified by calculating the torque about the axis normal to the outsole and the playing surface from the recorded force signals. In the translation test, a shoe is pulled or pushed 0.35 m by an air cylinder piston linked to the shaft. The foot form can be pinned in many different positions to simulate different foot orientations. Sliding velocity is adjusted to speeds approaching foot strike velocities and measured with an LVT. The normal and frictional force components are calculated from the sampled forces. The ratios of these components equals a coefficient of friction.

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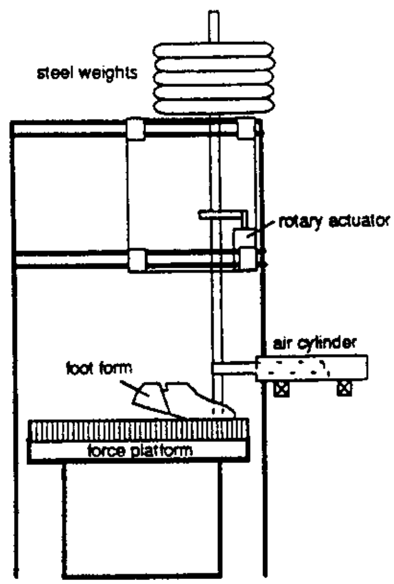

Figure 1. Sketch of a device for mechanically evaluating translational and rotational traction characteristics of shoe outsoles under sport-specific loading conditions.

Example Applications of the Mechanical Device

The material out of which outsoles or playing surfaces are composed has the greatest effect on traction. Valiant et al. (1985) showed that harder Nitrile rubber cleats develop 26% lower frictional forces on artificial turf than softer Styrene-Butadiene rubber cleats moulded in identical geometries. Physical test results reported by Nigg et al. (1987) demonstrate that friction coefficients of different shoe outsoles are about 50% greater on artificial grass than on a non-porous synthetic tennis surface.

Traction is also affected by the conditions of the playing surface or the outsole material. Kinetic coefficients of friction of walking shoe outsoles are reduced an average of 65% when linoleum is wetted (Valiant, 1993). Translational and rotational traction of shoe outsoles are decreased 55% when a basketball surface becomes dusty (Figure 2).

Outsole design also affects traction. The coefficient of friction of rubber or urethane basketball shoe outsole material increases with increases in surface area (Valiant, 1987a). A first order relationship between kinetic coefficient of friction and normal pressure within the range 35 - 207 kPa accounts for 90% of the variance of the combined non-Coulomb dependence of outsole traction on surface area and on normal load (Valiant, 1987b).

Factors affecting translational traction also affect rotational traction, including material effects, playing surface condition effects, and design effects, as demonstrated by the data in Figure 2.

Concluding Remarks

Physical traction measurements and traction tests involving human subjects performing movements both have merits and limitations. Physical tests are very repeatable, and subject variability or trial - trial variability is removed. Thus, the dependent variable associates a characteristic to an outsole / playing surface combination for a specific sports application as long as loading conditions are accurately simulated. The main limitation of physical traction tests is the imperfect reproduction of in-use loading conditions. It becomes especially important to deal with this limitation knowing that the classical Coulomb relations do not apply to footwear outsole rubber and polymers. The primary advantage of human subject traction tests is the application of the unique subject variability in specific human movements that cannot be incorporated in mechanical tests, particularly if traction characteristics are responsible for kinematic alterations. However, the experimental movements performed in the laboratory are not guaranteed to be valid reproductions of in-use performance.

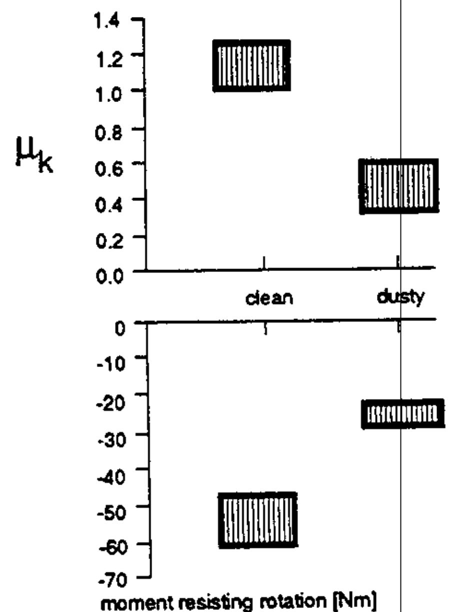

Figure 2. Ranges of kinetic coefficients of friction and ranges of peak moments resisting medial rotation of a urethane basketball shoe outsole on clean and dusty finished wooden basketball surfaces (from Valiant, 1993).

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THE INFLUENCE OF RUNNING SPEED ON REARFOOT MOTION, TIBIAL ACCELERATION AND IN-SHOE PRESSURE DISTRIBUTION

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INTRODUCTION

The influence of running speed on various kinematic and kinetic variables has been reported by several researchers. Increased running speed (Nigg, 1986) results in a higher rearfoot range of motion. For treadmill running, Clarke et al. (1983) found increased shank acceleration of 72% for a group of 10 subjects when running speed increased from 3.35 m/s to 5.36 m/s. Milani & Hennig (1988) investigated the influence of running speed on in-shoe pressures. Comparing the two running speeds 3.0 m/s and 4.5 m/s, peak pressures increased approximately 13% in the rear and 8% in the forefoot at the higher velocity. Most of the above studies have only looked at isolated biomechanical variables of foot ground interaction. Because these studies were independently conducted with different kinds of footwear at various speeds, a general picture of the effect of running speed on kinematic and kinetic variables is not available. The purpose of the present study was to simultaneously record rearfoot motion, tibial acceleration and in-shoe pressures for running speeds between 3.0 m/s and 5.0 m/s.

METHODS

Twelve male subjects with an age of 27.7 years (SD 3.1) and a body mass of 78.7 kg (SD 9.9) were instructed to run in a randomized sequence of speeds at 3.0, 3.5, 4.0, 4.5, and 5.0 m/s on a "Woodway" treadmill (type Ergo ES2). The subjects were well-trained runners with treadmill running experience. Eight repetitive trials were performed by all subjects for each of the speeds. The same type of running shoe (Adidas Torsion ZX 8000) was chosen for all subjects. Heel strike running style was identified by analyzing the pressures under the heel, midfoot and forefoot structures. If the heel pressures during the first 20 ms after contact were more than 5 times higher than the sum of all other pressure values, the trials were accepted as rearfoot strike running. Eight anatomical locations under the foot (medial & lateral heel and midfoot; 1st, 3rd and 5th metatarsal heads; hallux) were palpated and piezoceramic transducers (4 mm x 4 mm) were fastened under the foot with an adhesive tape. An "Entran EGAX-F-25" miniature accelerometer was attached to the medial aspect of the tibia at a mid location between medial malleolus and the tibial plateau. An electrogoniometer (Megatron MP 10) was fixed at the heel counter with its axis of rotation at the approximate height of the subtalar joint. The movable part of the goniometer was fixed at the lower leg in parallel to the achilles tendon orientation. Angular motion between rearfoot and achilles tendon orientations is an indirect estimation of subtalar joint pronation. The pronation values, as they are presented in this study, refer to the angular motion from a bipedal standing situation to the maximum achilles tendon angle during ground contact. Maximum pronation velocity was calculated from the angular values. The data of tibial acceleration, rearfoot motion and pressure distribution were collected simultaneously by a computer in a pretrigger mode at a rate of 1 kHz per channel and a resolution of 12 bit. Maximum pronation (PRON) and pronation velocity (PVEL), peak tibial acceleration (PACC), and the mean peak pressures and impulses were averaged across 8 repetitive trials in all conditions. Relative loads were determined as the percentage of each single impulse to the sum of all impulses. Tibial acceleration data were only available from 9 of the 12 subjects. A repeated measures ANOVA was used in this study.

RESULTS & DISCUSSION

Table 1 summarizes the mean values for maximum pronation and pronation velocity as well as the maximum tibial acceleration at the chosen running speeds. The ANOVA revealed significant ($p<0.01$) speed effects for the three variables. An almost linear increase of all three parameter values with running speed could be observed. The maximum achilles tendon angle increased by 1.6 degrees for a change in running speed from 3 to 5 m/s. At identical running speeds, Nigg (1986) reported a similar change in achilles tendon angle of 1.5 degrees for 16 subjects, running in a shoe with a soft midsole material. Eddington et al. (1990) reported large deviations in the values of rearfoot motion variables, when comparing studies of different researchers. Pronation velocity at 4.5 m/s, as found in this study compares well with the relatively high values (789 °/s) from Cavanagh (1978) at the same running velocity. It is interesting to note that pronation only increased by 1.6 degrees whereas pronation velocity changed by more than 50% from the lowest to the fastest running speed. From 3.0 m/s to 5.0 m/s the tibial acceleration increased about 78.5 %. This value is similar to the increased tibial acceleration of 72%, reported by Clarke et al. (1983) for the higher running velocity of 5.36 m/s, when compared to a low speed of 3.35 m/s.

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The peak pressures under the foot are summarized in table 2. Under most anatomical locations a steady increase of the peak pressures with higher running velocities can be seen. Peak averaged heel (PAV-H) and averaged forefoot pressures (PAV-F) were calculated from the two heel (LH &MH) and four forefoot (M-5, M-3, M-1, H) anatomical locations and plotted against the running speed (figure 1).

speed (m/s)	PRON (°)	PVEL (°/s)	PACC(g)
3.0	3.86	564	6.5
3.5	3.90	598	7.5
4.0	4.52	633	8.8
4.5	4.80	737	10.7
5.0	5.48	857	11.7

Table 1: Pronation (PRON), pronation velocity (PVEL) and tibial acceleration (PACC) for 5 different running speeds

	3.0	3.5	4.0	4.5	5.0
LH	815	921	985	1064	1102
MH	991	1078	1167	1242	1326
LM	446	482	477	491	531
MM	242	254	272	277	272
M-5	740	756	745	757	741
M-3	758	826	829	879	907
M-1	1230	1268	1361	1417	1572
H	658	706	734	780	836

Table 2: Planatar peak pressures (kPa) at 5 running speeds (m/s). Lateral and medial heel & midfoot (LH, MH, LM, MM), 5th, 3rd, 1st met. heads (M-5, M-3, M-1) and hallux (H)

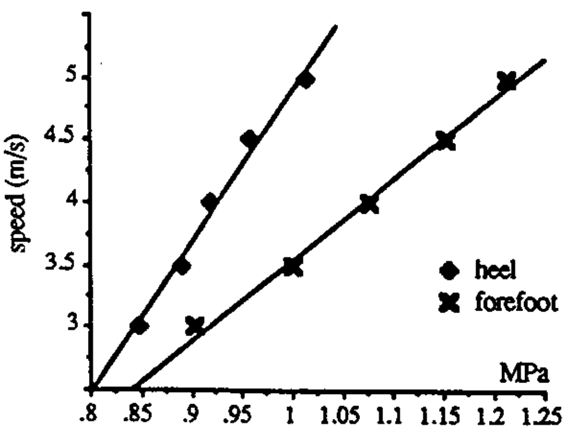

Figure 1: Peak pressures under the heel and forefoot with increasing running velocities

The graph demonstrates that higher running velocities result in a substantially steeper increase of the heel versus the forefoot pressures. However, this imbalance of rear to forefoot loading with speed is not reflected by the relative load analysis. Although heel pressures increase by approximately 15% more with speed than the forefoot pressures, the relative loads remain almost the same in both foot regions. At 3m/s heel and forefoot loads of 17.1% and 71.6% are similar to those found at 5 m/s (heel=18%; forefoot=71.6 %). This indicates that at higher running speeds rearfoot strikes occur harder but for a shorter time duration. The trend towards harder but shorter heel impacts with higher running velocities may also be responsible for the surprisingly large increases of peak tibial acceleration with speed.

	VEL	PRON	PVEL	PACC	PAV-H	PAV-F
VEL	1					
PRON	.97	1				
PVEL	.96	.972	1			
PACC	.993	.965	.964	1		
PAV-H	.997	.95	.938	.988	1	
PAV-F	.994	.973	.979	.984	.985	1

Table 3: Correlation matrix between the different kinematic and kinetic variables (VEL = running speed)

Because this study combined the informations of pronation, tibial acceleration and plantar pressures the relationship between these variables with increases of running speed was of special interest. A correlation matrix (table 3) shows that high correlation coefficients were found for all variables with speed. The high correlations between the biomechanical parameters also demonstrate that the speed effect influences the different variables to a similar degree.

CONCLUSION

Maximum rearfoot pronation and pronation velocity, peak tibial acceleration and the plantar in-shoe foot pressures were found to increase in an approximate linear fashion with running speed. At higher running speeds the subjects run with increased rearfoot impacts of shorter duration. Lafortune et al. (1994) found in a comparison of treadmill and overground running substantial differences in plantar pressures and tibial shock variables. A future study should, therefore, investigate if the results from the present study also apply to overground running.

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IN-SHOE HEEL PRESSURE DISTRIBUTION DURING TREADMILL RUNNING

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INTRODUCTION

Previous work has demonstrated that in-shoe pressure measurements during overground running are able to differentiate the *in-vivo* cushioning abilities of footwear with different midsole densities (3). Research into the biomechanical aspects of locomotion is often facilitated by using a treadmill in a controlled laboratory setting. The degree to which treadmill experimental results can be used to evaluate footwear properties is still open to question. Heel in-shoe pressure measurements were recorded during treadmill running in nine subjects wearing shoes with different midsole densities. The pressure results were able to discriminate between the midsole densities of footwear but the distribution of pressure was different from overground data.

REVIEW AND THEORY

Several studies have indicated that there are small differences in the kinematics of treadmill running compared to normal overground running (1,7). Little research has focused on whether treadmills adequately simulate the loads experienced during overground locomotion. Milani et al. (1988) found considerable increases in calcaneal impact shock experienced during barefoot treadmill running compared to overground running. More recently, shank acceleration and discrete pressure sensor measurements during shod treadmill and overground running indicated that impact loading was reduced on the treadmill (4). In-shoe pressure measurements provide information relevant to reduction of symptoms associated with overloading of the foot (2) and to the design of footwear to cushion the forces associated with foot-ground impact. This study compared the magnitude and distribution of in-shoe heel pressures during treadmill running in experimental shoes with differing midsole densities.

PROCEDURES

Nine male recreational runners (mean: 71.0 kg and 175.1 cm) were asked to run in three experimental shoes differing only in midsole density (hard, 0.4 g/cm^3; medium, 0.3; and soft, 0.25) at 3.83 m/s on a motorized treadmill (Collins, model 07203). All subjects were rearfoot strikers and familiar with treadmill running. Heel pressure distribution was monitored with 25 h.a.l.m. discrete pressure sensors located in specific areas. To overcome point loading (Lake et al., 1991), the pressure sensors (size: 5 mm by 4 mm by 2.5 mm) were encapsulated in a thin layer of silicone rubber material to form an insole which fitted tightly inside the shoes. Subjects performed a five minute treadmill run in each shoe condition on two visits to the laboratory. The order of shoe conditions was randomized between subjects. Heel pressure signals were sampled at 500 Hz by a MicroVax/GPX II computer and digitally filtered at 50 Hz. Pressure signals from a total of forty contacts for each shoe condition were recorded for averaging purposes. The distribution of pressure was determined by calculating the relative impulse of the various heel regions as a percentage of the total impulse in all regions.

RESULTS

For the entire heel area, as the midsole density increased so did rate of loading (ROL) and peak pressure (PK). The results of these two variables for specific regions of the heel can be seen in figure 1. Generally, the ROL variable was better able to differentiate between shoes. Independently of the shoe used, area B results (nearer the arch) were significantly lower than under any other regions of the heel. Pressure measurements along the lateral border (area A) and the extreme rear part of the heel (areas C and D) were better discriminators of midsole density. The relative impulse distribution over the heel areas was similar between shoe conditions, with approximately 29 % in area A, 26 % in E, and 7-9 % in areas B, C and D.

DISCUSSION

In comparison to overground running heel pressure data on the same subjects (3), treadmill ROL and PK were 30-40 % lower. In addition, the relative impulse for the overground data in areas C and D was larger (12 %), while in area A it was lower (23 %). This may suggest a slight change in strike index on the treadmill which was consistent between shoes. This comparison indicates that for a group of heel-toe runners, pressure related impact loading is reduced during treadmill running. This agrees with the recent treadmill results of Lafortune et al. (1994) who found that acceleration of the shank and pressure under the

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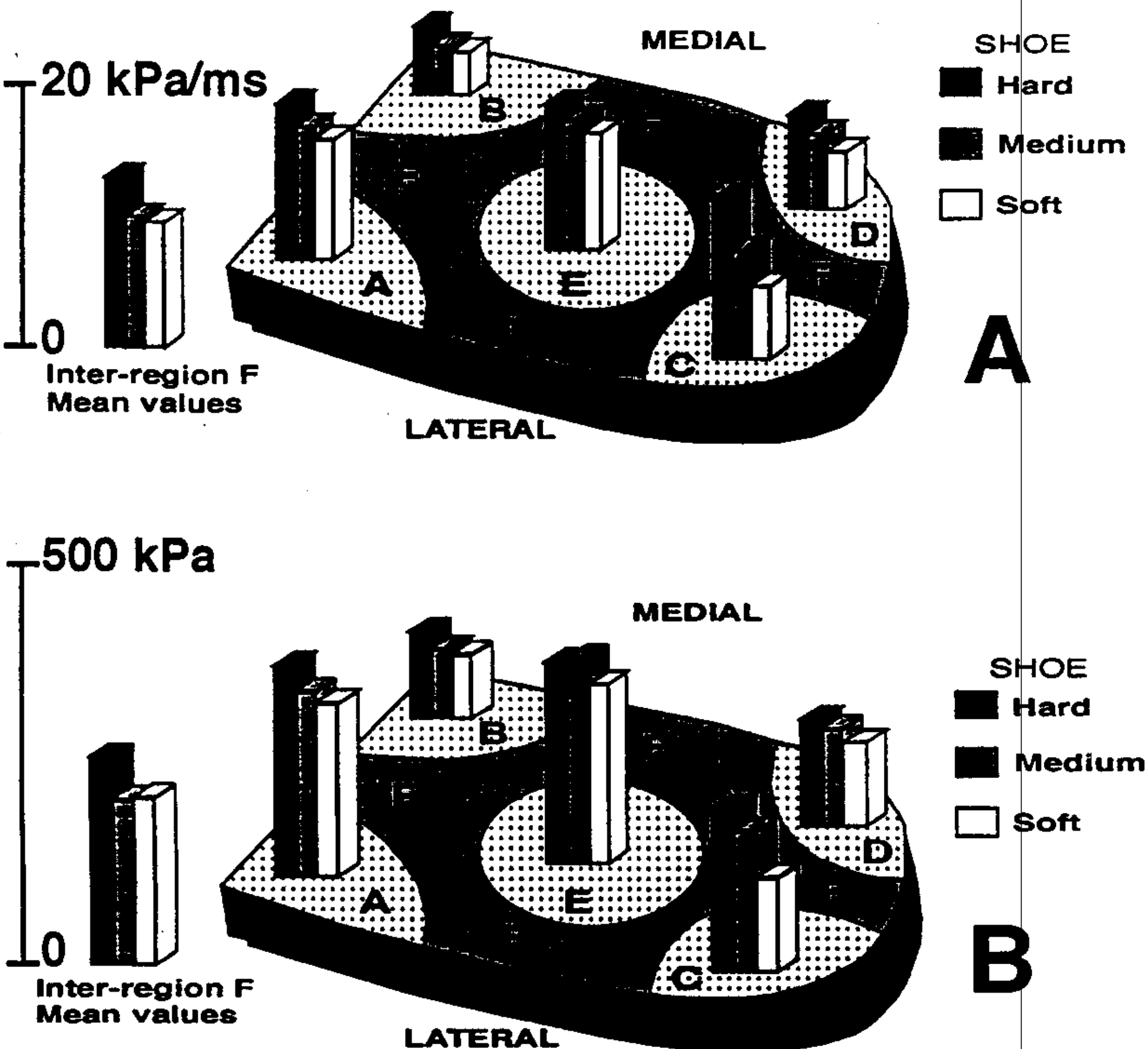

Figure 1. Rate of loading (A) and peak pressure (B) distribution.

central portion of the heel was significantly lower compared to overground running. The more compliant properties of the treadmill may be the main reason why pressures were reduced over the entire heel area in this study. Nevertheless, the treadmill modality still allowed footwear properties to be discerned. The rear and lateral border of the heel were shown to differentiate better between shoe conditions than other heel areas. This was also the case for overground running (3), and has implications for sensor placement when the number of sensors are limited.

This research was sponsored by Nike.

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SURFACE EFFECTS ON IMPACT FORCE ATTENUATION

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INTRODUCTION

Treadmill (TM) manufacturers have recently begun to design shock absorbing systems to reduce impact forces (IF). The effects of these mechanical systems on running IF magnitudes has not been thoroughly investigated.

Two instrumentation systems, high speed video and skin-mounted accelerometry, were incorporated to evaluate the cushioning effect of two TM beds. Results identified IF differences between surface hardness (33.6%) for all group and 88.9% of the within-subject contrasts. Methodologically, accelerometry produced greater g values than comparable videography measures.

REVIEW AND THEORY

IF is a well documented cause of running injuries (James et al., 1978; Nigg, 1985; Perry, 1983). Traditionally, researchers have primarily focused their efforts on the effects of shoes for reducing impact forces (Bates et al., 1983; Clarke et al., 1983; Nigg et al., 1988; Nigg et al., 1987) with little emphasis directed toward the effects of surfaces on IF (Bates et al., 1992). Shock absorbing characteristics are currently a design feature being offered by several TM manufacturers, with the intent of reducing IF. The purpose of this study was to investigate the effectiveness of a shock absorbing treadmill bed system compared to a non-cushioned bed in reducing lower extremity impact forces. A secondary purpose was to compare the results from two measurement systems.

PROCEDURES

Data were obtained from six male subjects running at a self-selected pace. A Kistler uniaxial accelerometer was securely attached to the right tibial tuberosity. Ten footfalls of accelerometer (1000 Hz) and kinematic video data (200 Hz) were simultaneously collected for two treadmill conditions: non-cushioned bed (NC) and shock absorbing bed (SA). Kinematic data were digitized using a Motion Analysis VP320 video processor while accelerometer data were processed via an Ariel APAS System. The kinematic data were smoothed using a second-order Butterworth filter with an optimal cut-off algorithm based on Jackson (1979). Vertical ankle and knee acceleration values were calculated using a cubic spline function. The first maximum and minimum values were identified along with corresponding absolute times from contact. Similar values were obtained for the accelerometer data. The analysis consisted of within-subject (Model Statistics: Bates et al., 1992) and group (repeated measures ANOVA) evaluations.

RESULTS

Table 1 contains the maximum IF values for all subject-conditions along with the group mean values. NC produced significantly greater IF values for 88.9% of the individual subject comparisons which is also reflected in the significant group results. Within conditions, ankle values were greater than knee values for 91.7% of the within-subject comparisons and the NC group comparison.

The times to maximum acceleration for the ankle and knee joints are presented in Table 2 (temporal data unavailable for the accelerometer). As would be expected, the times on NC were on the average shorter (not significant) than the SA values, but only two subjects (S1, S3) actually exhibited statistically significant differences. Differences between joints within conditions were more clearly defined with shorter times for the ankle maxima compared to the knee maxima for all NC and 68.0% of the SA comparisons. Both group comparisons were significant.

Temporal differences between the maximum and following minimum acceleration values are presented in Table 3. Although 66.7% of the subject comparisons were significant, the only significant group mean was for the accelerometer data. Nine of 10 between joint within condition comparisons resulted in significantly longer ankle values. However, only the NC group condition value was significant.

Comparisons were also made between the acceleration values for the ankle and knee compared to the accelerometer data. Individual results were not consistent for SA ankle, producing a nonsignificant group result. All other comparisons except for S5 ankle NC resulted in significantly greater accelerometer values. The same trend was observed for the corresponding time differences (Table 3) with 21 individual (17 significant) and three group values being shorter. The SA knee values were less consistent for these data.

DISCUSSION

The range of values observed for the accelerometer data in the present study was considerably greater than that observed by Hennig et al. (1993). Two probable reasons for the difference are subject body mass and running speed. Subjects in the present study were purposely selected to cover a wide range of sizes (80.1 ± 10.1 kg compared to 72.0 ± 7.2 kg) and allowed to run at a self-selected pace. Although mean velocity was the same (3.3 m/s) variability for this study was considerably greater (SD = 0.85 m/s). A third possible reason for the differences could be due to the accelerometer attachment. We believe this is a less likely reason since considerable care was taken when mounting the accelerometer.

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Results

Maximum rise distance of the foot off the ground was significantly different at 0.88 ($p<0.01$)and 1.32 m/sec ($p<0.05$). The mean maximum rise values at a walking speed of 0.88m.sec were 0.06 meters on the ground versus 0.08 meters on the treadmill. At a walking speed of 1.32 m/sec maximum rise of the foot was 0.07 meters on the ground versus .09 meters (Fig. 1). Larger knee and ankle flexion values were also observed for treadmill versus overground walking although they were not significantly different. However, significantly larger hip flexion angles were found at both 0.88 and 1.32 m/sec ($p < .001$) on the treadmill. Hip flexion at a walking speed of 0.88 m/sec was 20.62 degrees on the ground and 23.61 degrees on the treadmill; at a walking speed of 1.32 m/sec hip flexion was 20.18 degrees on the ground and 23.67 degrees on the treadmill. (Fig. 2) These results indicated that the increased rise distance of the foot on the treadmill versus the ground was probably due to an increase in hip flexion. Stridelength was also significantly different between the two conditions at 0.44 m/sec (0.87 on the ground versus 1.19 meters on the treadmill) and 0.88 m/sec (1.26 meters on the ground, 1.31 meters on the treadmill) but not at 1.32 m/sec (Fig 3).

Discussion

These results indicated that walking strategies differed at slow versus fast walking speeds. At the two slowest walking speeds, subject's steps were significantly shorter on the ground than on the treadmill. While at the two fastest walking speeds the strategy was towards greater hip flexion and higher rise distance of the foot from the ground during treadmill walking versus overground walking. The results lead to the conclusion that the treadmill influences walking kinematics compared to overground walking and influences kinematics differently throughout the range of speeds.

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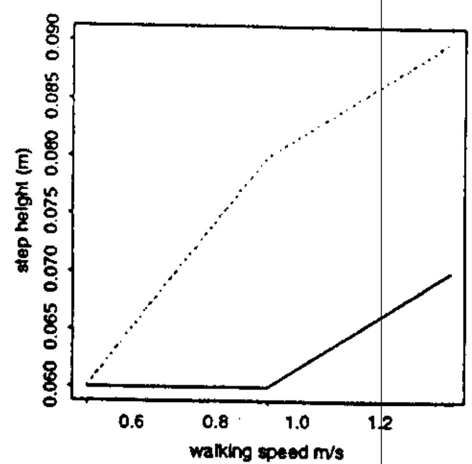

Figure 1. Maximum rise of the foot from the ground (step height) at .44, .88 and 1.32 m/sec. Solid lines indicate overground walking and dotted lines are treadmill walking

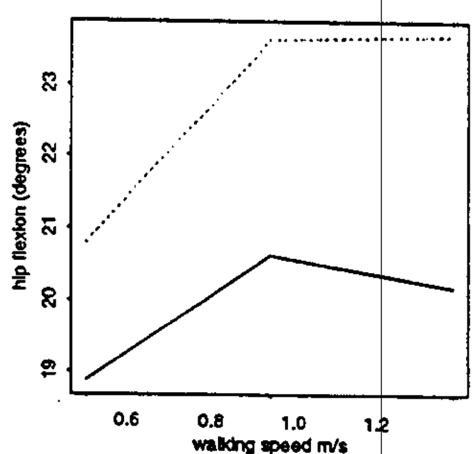

Figure 2. Hip flexion during swing phase at all walking speeds for overground (solid line), and treadmill walking (dotted line).

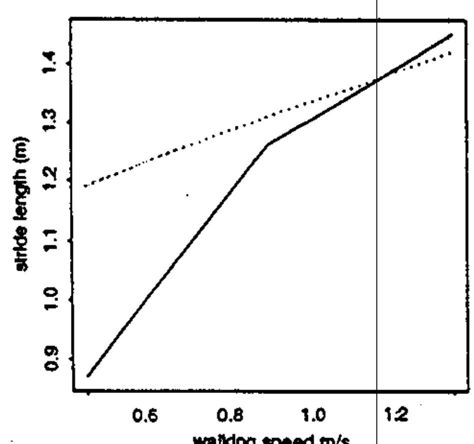

Figure 3. Stridelength for overground walking (solid line) and treadmill walking (dotted line).

COMPARISON OF TREADMILL AND OVERGROUND RUNNING

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INTRODUCTION

Treadmills are commonly used as rehabilitative devices, to evaluate physiological performance and footwear properties. Several biomechanical studies have assessed the ability of treadmills to mimick overground running. Based upon kinematic measurements and associated parameters, researchers have reported minimal differences for treadmills with powerful drive systems.

In the present study, kinetic and rearfoot measurements were used to compare treadmill to overground shod running. Kinetic measurements included foot plantar pressure and shank axial acceleration. Treadmills reduced peak pressure under the foot except under the arch and the big toe, they also reduced shank impact shock but had minimal effect upon pronation. Some differences were observed between treadmill types.

REVIEW

Scheib (1986) reported that habituation to treadmill is necessary to establish a normal running gait. While Dal Monte et al. (1973) suggested that differences existed between treadmill and overground running, Van Ingen Schenau (1980) stated that no biomechanical difference existed for devices with strong driving mechanisms. This last study relied upon a theoretical model of the energetic aspects of running. In 1989, Milani et al. presented results of barefoot calcaneal acceleration that implied higher impact during treadmill running. Their distal accelerometer positionning excluded the foot/ankle complex which likely plays an important shock attenuating role. Thus, impact measurement at the calcaneus may not be representative of the shock wave travelling up the locomotor system. Barefoot locomotion was also shown to produce impact loading that is different from shod locomotion (Lafortune, 1992). The purpose of this study was to examine the treadmill-overground comparison for shod running.

METHODOLOGY

Fifteen healthy male subjects experienced at treadmill running participated in the study (73.3±5.5 kg). They ran at 3.3 m/s overground (OVGR) and onto two different treadmills (TREAD1, Woodway; TREAD2, Jaeger Laufergotest) while wearing running footwear. The overground range of accepted velocities was ± 0.1 m/s. Plantar pressure was monitored with piezoceramic transducers (4mm x 4mm). Their physical properties were described by Hennig et al. (1982). The sensors were fastened under the foot with adhesive tape at the following palpated anatomical locations:beneath the medial and lateral heel and midfoot, the head of the 1st, 3rd and 5th met., and the hallux. An electrogoniometer was used to determine the rearfoot angle which provides us with an estimate of foot pronation. Shank acceleration was measured with an "Entran EGAX-F-25" miniature accelerometer attached to the medial aspect of the tibia at mid distance between medial malleolus and the tibial plateau. To improve its mechanical coupling to the tibia, the accelerometer was forcefully pressed against the bone by an elastic band. Analog signals were sampled at 1 kHz.

The two treadmills were presented on different days while overgroung running was repeated on both days to account for potential differences in transducer's mounting and/or location. Eight trials were performed in each condition and mean values were calculated for pressure, acceleration and pronation (rearfoot) variables. Peak (PKP) values characterized plantar pressure; peak (PKA) and median power frequency (MPF) for shank acceleration; maximal rearfoot angle (MRA) and angular velocity (MAV) for pronation. The load bearing description at individual locations was further complemented by integrating the local pressures throughout the stance phase. Relative loading (RLP) was calculated by dividing each local

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impulse by the sum of all eight local impulses. ANOVA (α=0.05) was used to compare experimental conditions.

RESULTS

Compared to OVGR both treadmills were found to significantly reduce PKP under most foot structures (Fig. 1). The differences were not significant under the arch and big toe and, the 5th met head with TREAD2. The only difference in relative loading (RLP) was at the lateral heel, TREAD2 decreased RLP. PKA was significantly reduced by both treadmills but only TREAD2 increased MPF (Table 1). Treadmills had no effect upon peak pronation but TREAD2 increased MAV. Finally, it was found that TREAD2 reduction in PKA was significantly larger than the reduction with TREAD1.

Table 1. Means and standard deviations.

	OVGR1	TREAD1	OVGR2	TREAD2
PKA g	8.8	7.5 *	8.8	5.5 *+
s.d.	2	2.9	1.8	1.9
MPF Hz	15.6	17.6 *	15.4	16.6
s.d.	0.9	2.6	1.2	2.7
MRA d	11	11.2	10.5	10.7
s.d.	2.7	3	3.9	4.4
MVA d/s	482	587 *	480	525
s.d.	98	150	106	133

* OVGR ≠ TREAD + TREAD1 ≠ TREAD2

DISCUSSION AND CONCLUSION

The large discrepancies in peak pressure and peak shank acceleration indicate that the loading characteristics of shod treadmill running were substantially different from overground running. The lower values obtained for the treadmills were likely indicative of a greater compliance in these devices rather than kinematic modification of the foot-ankle complex. The shock attenuating pronation was found to be similar between treadmill and overground conditions. The greater peak pressure reduction under the heel as compared to the forefoot could have implied a change toward midfoot striking during treadmill running. Yet, the relative loading data showed that load sharing among foot structures remained unchanged from overground running. These findings also suggest a greater compliance in the treadmills.

From a footwear evaluation perspective, the shock and pressure reduction associated with treadmill use may lead to lower discriminating power. Yet, in rehabilitation these same cushioning characteristics would be benefitial. Pronation velocity, median power frequency and peak shank acceleration indicate that treadmills differ and as such the choice of a specific model should depend upon its intended usage.

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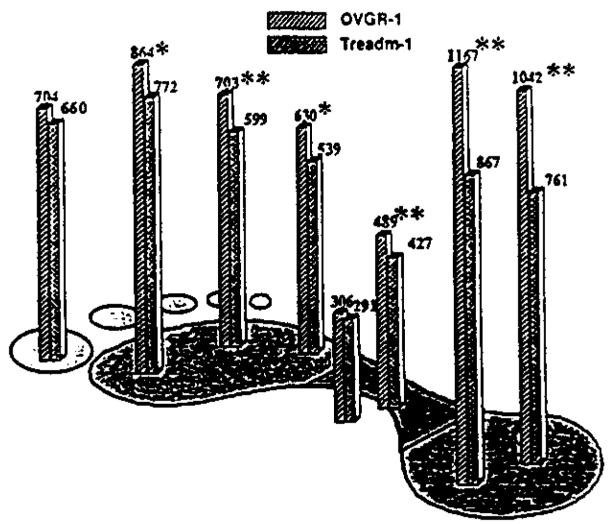

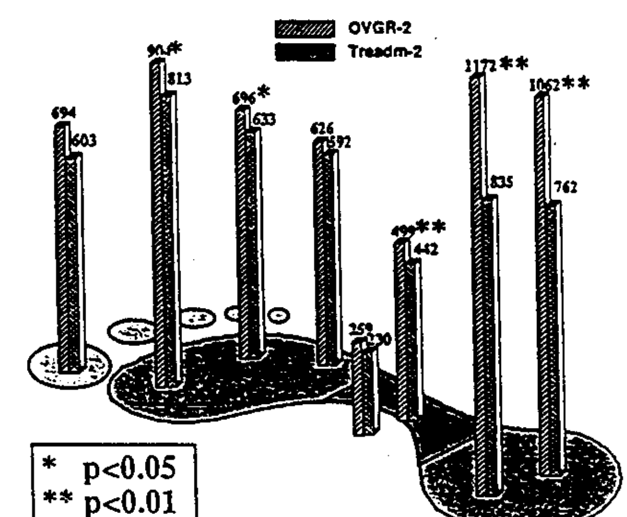

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RESTRICTION OF FOOT INVERSION BY TREKKING SHOES IN UNEXPECTED ANKLE TURNS

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INTRODUCTION AND REVIEW

Ankle sprains are the most frequent injuries in sports (Mack 1982). In various sport disciplines (e.g. basketball, tennis & mountain climbing) 30% to 34% of the athletes are suffering ankle sprains (Southmaid & Hofman, 1981). According to Taylor (1988), ankle sprains are mostly acute athletic injuries, resulting from sudden blows or twists. Therefore, an important criterion for the protective function of trekking shoes is the effectiveness of the shoes to protect the body from foot inversion injuries, which generally occur unexpectedly. According to Melville-Jones & Watt (1971) and Greenwood & Hopkins (1976) an unexpected fall situation is forced when the time to react on a stretch is less than at least 200 ms to initiate a voluntary contraction. Unexpected turning of the ankle normally occurs too fast for protective voluntary muscular contraction. Up to a time of 100 ms only simple (at around 30 ms) or vestibular reflexes (at approximately 100 ms) can be expected. Using a similar protocol, Kimura et al. (1987) studied the effectiveness of different ankle braces on rearfoot motion. This study was performed to investigate the mechanical properties of 13 different trekking shoe products in preventing excessive foot inversion in a simulated unexpected ankle turn situation.

METHODS

Twenty-three male subjects volunteered to participate in this study. The average age was 24.9 (SD 4.3) years, the mean height 178 (SD 4.8) cm and the mean mass 70.1 (SD 6.4) kg.From the 13 trekking shoe constructions, one had a low, nine a medium and three a high cut heel shaft. A metal platform with the dimensions of 35 cm x 36 cm was constructed. The functional portion of the platform is a tiltable section to invert the right foot to a preset angle of 20° (Fig. 1).
For protection the tiltable part of the platform (17 cm x 35 cm) was built with a 7 cm high boundary. Triggering an electromagnetic release mechanism the movable platform part dropped to an angle of 20°. The subjects on the platform were not aware of the time of release to create an unexpected fall situation. The subjects were instructed to position their right foot with the lateral edge parallel to the boundary. The subjects stood with almost full body weight on their right foot on the movable platform section. The left foot on the fixed portion of the platform served only to achieve a stable standing balance. The falling distance of the heel center was determined as 4.3 cm, which corresponds to a falling time of approximately 94 ms. According to Melville-Jones & Watt (1971), during this period only muscular activity by a simple reflex can be expected. Therefore, the restriction of foot inversion can be expected to be mainly a consequence of the type of the shoe construction.

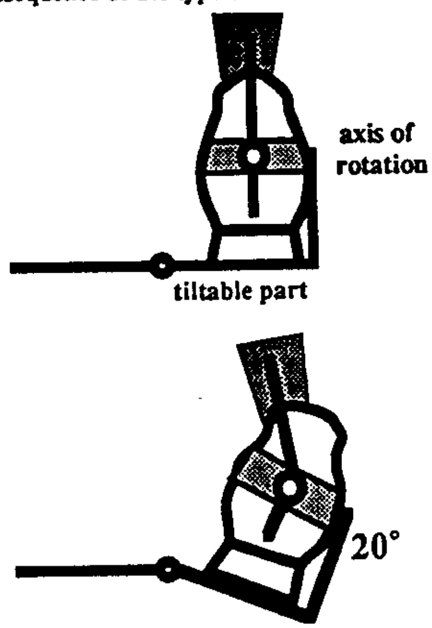

Figure 1: Platform and goniometer arrangement for the simulation of unexpected foot inversion

The angle of inversion was measured by an electrogoniometer to determine the achilles tendon angle for the estimation of rearfoot motion. A light weight half circular metal construction with a potentiometer was fastened at the heel counter of the trekking shoes. The axis of rotation of the goniometer was adjusted to the approximate height of the subtalar joint (figure 1). A precision conductive plastic potentiometer (Megatron MP10) exhibited a high angular resolution and good linearity (1%). The movable part of the goniometer was fixed at the lower leg in alignment to the achilles tendon orientation. The inversion platform was mounted onto a Kistler force platform (Model 9261A) and data collection was initiated with a pretrigger mode, using the ground reaction force signal. With all subjects 5 trials of unexpected falls were carried out in each of 13 commercially

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available trekking shoes. Testing for one person was done in a single session using a randomized shoe sequence. The data were sampled with 12 bit resolution and at a sampling rate of 1 kHz. The maximum inversion angle was defined as the range of angular displacement from standing to maximum inversion. This range of angular motion was easier to detect and its use seemed adequate, because the amount of pronation during standing was very similar between the different footwear constructions. Additionnally the maximum inversion velocity were determined for each trekking shoe and each subject. After averaging the parameter values across the repetitive trials in each of the footwear conditions, the mean values were used in a repeated measures ANOVA.

RESULTS AND DISCUSSION

The statistical evaluation shows large differences between the different shoe constructions. For the maximum inversion angle as well as for the maximum inversion velocity the ANOVA revealed significant ($p<.0001$) between shoe differences. The inversion angles for the 13 trekking shoes ranged between 34.3° to 43.7° (table 1).

Parameter	Mean	SD	Range
Angle of inversion [°]	37.0	5.04	34.3-43.7
Inversion velocity [°/s]	658.4	164.9	576-807

Table 1:Mean, standard deviation (SD), range of inversion and inversion velocities

Figures 2 and 3 show the inversion angles and angular velocities for the 13 pairs of shoes. Whereas the highest angle of inversion was observed for the shoe with the low heel shaft (#1), low angles could be found for the three trekking shoes with high shafts (#11, #12, #13). However, trekking shoe #3 with a medium shaft height limited angle of inversion as effectively as the high cut models.

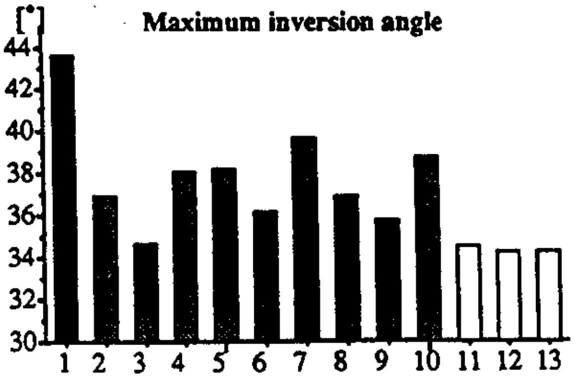

Figure 2: Maximum inversion angle

The angular velocity measurements (figure 3) show a similar trend as the maximum inversion findings. Again, shoe #3 with a medium shaft height limits the angular velocity as effectively or even better than the high cut shoes. A simple regression analysis revealed a significant correlation ($p<.01$) of $r = +0.84$ between maximum inversion and maximum inversion velocity. However, shoes #2, #4, #12 and #13 show that inversion range and inversion velocity behave differently. Therefore, both quantities are important to judge the protective function of footwear. The maximum inversion of the foot may be used to estimate the amount of loads at the ligaments. A low inversion velocity of a trekking shoe may give the amount of extra time to use voluntary muscle contraction in avoiding an excessive inversion. Overall comfort of the shoes was evaluated using a 5-point perception scale (1=very comfortable, 5=very uncomfortable). A correlation analysis revealed a weak, but significant relationships ($p<.05$) between perception of comfort and maximum inversion ($r = +0.60$) as well as inversion velocity ($r = +0.60$). Opposite to what one might have expected, the more comfortable shoes had a tendency to restrict inversion to a higher degree.

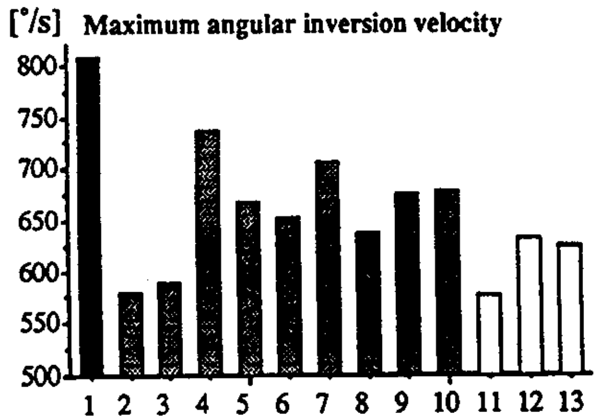

Figure 3: Maximum angular inversion velocity

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MECHANICAL PROPERTIES OF RUNNING SHOES - MEASUREMENT AND MODELLING

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1. INTRODUCTION

In the last twenty years the running shoe has often been subject of mechanical (1,3) and biomechanical (4) studies. The shoe can be characterized by mechanical parameters such as "stiffness" and "damping", often being more or less complex functions of deformation and deformation velocity respectively.

The purpose of this paper is to present a new measuring device, that offers the possibility to measure under "realistic conditions" and to demonstrate different possibilities in quantifying the mechanical properties of running shoes as viscoelastic materials. With the collected data of different running shoes in a new and worn condition, the following thesis are provided with the help of an appropriate statistical design (multifactorial analysis of variance).

1. There are differences in the mechanical properties of the tested high price running shoes.
2. The properties of the shoes change after a sportspezific usage.

2. MATERIALS AND METHODS

The tests were done with each of 10 pairs of five different high price running shoes of important manufacturers in a new and worn (200 km) condition. The mechanical properties of the shoes in the heel and midfoot region were measured using a dynamic test device. The equipment offers the possibility to exert force on sportshoes, sportshoe-materials or sportfloors. The force acts over a variable piston onto selected parts of the shoe. The maximum of effective force, the time of contact and the velocity of the piston is variabel in certain limits.

For this study we choose the parameters in the way that the time-curve of the resultant force is similar to the Vertikal Ground Reaction Force in running. During the contact phase force, deflection of the material and derivated parameters can be measured with high frequence or calculated and recorded to a computer.

With the help of this measurement the following parameters describing mechanical properties of the shoes are calculated for the heel and the midfoot region of the shoe:

- stiffness between 200 N and 400 N (ST 1 [N/mm]) and between 1000 N and 1400 N (ST 2 [N/mm]).
- maximum eniergie input (E_{IN} [Nm]) and the loss of energie durning the contactphase (E_L [%]).

Moreover we calculated the constants of a linear Kelvin-Voigt-Model and some nonlinear models describing the viscoelastic properties of the shoes. The three discussed models are given with the following differential equations.

$$F(t) = c_1 \cdot u(t) + \eta_1 \cdot \dot{u}(t) \tag{1}$$

$$F(t) = c_1 \cdot \frac{u^2(t)}{u_0 - u(t)} + \eta_1 \cdot \dot{u}(t) \tag{2}$$

$$F(t) = \sum_{p=1}^{6} c_p \cdot u^p(t) + \frac{\dot{u}(t)}{|\dot{u}(t)|} \cdot \sum_{q=1}^{3} r_q \cdot |\dot{u}(t)|^q \tag{3}$$

3. RESULTS AND DISCUSSION

As one can see in table 1, 2 and figure 2 (because of a better illustration only four of the five shoes are shown in figure 2) there are differences in the calculated properties of the tested running shoes in the heel as well as in the midfoot region. The most important parameter is stiffness 2, because it describes the properties of the tested shoe under a realistic high force input. Especially the difference in both parameters (ST 1 and ST 2) between shoe 3 and shoe 5 is very interesting though the maximum deflection of both shoes is nearly the same (see figure 2). From this point of view the relationship between the ST 1 and ST 2 could be an aditional parameter of importance.

The change of the properties after a specific use of 200 km running is dependent on the tested shoe. The relative as well as the absolute differences, especially of the parameter ST 2, correspond with the stiffness of the shoes in the way that the differences decrease with an increase of the stiffness of the shoes. The increase of ST 2 is more distinct in the midfoot than in the heel region.

The modelling of viscoelastic properties is connected with two problems of different nature. The usage of a linear Kelvin-Voigt-Model results in physically meaningful calculated spring and damper constants. The disadvantage of this model is the unacceptable agreement of the calculated with the measured

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forces. The application of model 3 shows an opposing character (good agreement of calculated and measured forces (see figure 3) but no possibility to a meaningful physical interpretation of the constants). Model 2 is a compromise of the two models because it is realtiv simple as well as it avoided the disadvantages listed above.

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Table 1: Mechanical parameters of the heel region

param	cond.	shoe 1	shoe 2	shoe 3	shoe 4	shoe 5
ST 1	new	122	104	77	96	61
ST 1	worn	130	104	78	94	62
ST 2	new	201	136	119	147	147
ST 2	worn	195	147	124	157	163
E_{IN}	new	8,5	11,6	13,9	11,1	12,3
E_{IN}	worn	8,6	11,1	13,4	10,7	11,5
E_L	new	66	59	62	62	51
E_L	worn	65	63	67	62	54

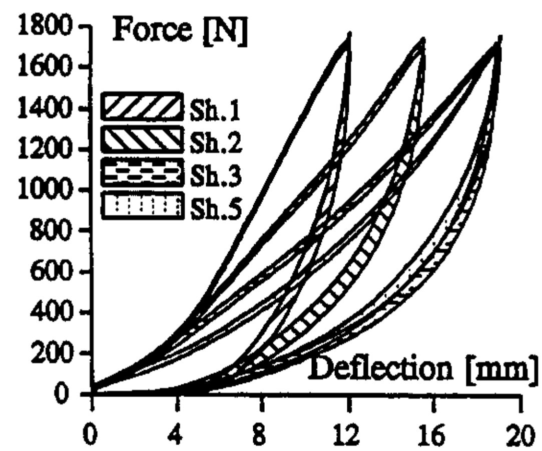

Figure 2: Force-deflection-curve - heel region, new condition.

Table 2: Mechanical parameters of the midfoot region

param	cond.	shoe 1	shoe 2	shoe 3	shoe 4	shoe 5
ST 1	new	117	85	95	88	74
ST 1	worn	117	78	91	91	72
ST 2	new	287	220	205	205	284
ST 2	worn	296	249	224	263	307
E_{IN}	new	6,8	8,7	9,0	9,1	7,8
E_{IN}	worn	6,5	8,1	8,4	7,8	7,5
E_L	new	70	65	70	71	61
E_L	worn	72	69	73	72	65

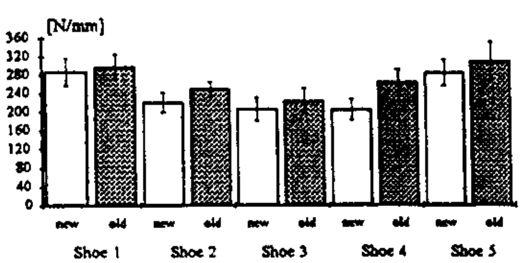

Figure 1: Stiffness2 (ST 2) in new and worn condition in the midfoot region

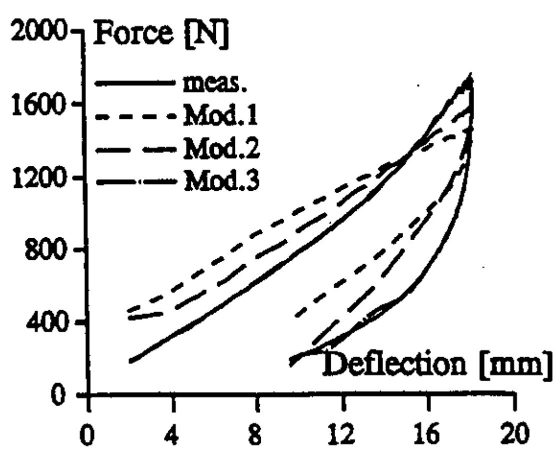

Figure 3: Measured and calculated force-deflection-curves - heel region, new condition.

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SHOCK TRANSMISSIBILITY OF THE HUMAN BODY

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INTRODUCTION

Acceleration transients result from the impact of the feet with the ground during locomotion. The human body controls the transmission of the shock to the skull through active and passive attenuation. Active attenuation refers to the kinematic responses of the locomotor system during impact while passive attenuation involves structures and visco-elastic properties of tissues.

The shock transmissibility of the human body was evaluated under 3 different impact conditions. A human pendulum was used as impact modality while knee kinematic response was minimized. The results indicated that the locomotor system was effective at attenuating the impact shock transmitted from the shank to the skull. Its performance appeared to be independent from the severity of the impact force.

REVIEW

Shorten and Winslow (1992) examined the frequency gain/attenuation of the human locomotor system during running. In comparing head to shank acceleration signals, they found that the spectral components of the head signal were attenuated above 10 Hz. The authors attributed this attenuation to kinematic adaptations that focused upon knee flexion. McMahon et al. (1987) reported that body shock transmission during running, ratio of head to shank peak acceleration, passed from ≈0.5 to 0.1 as thigh flexion increased from 20 to 30°. The purpose of this study was to assess the shock transmissibility of the human locomotor system under different impact conditions while knee kinematic response was minimized.

METHODOLOGY

Nine male subjects participated in the present study. They were free from lower limb injuries at the time of the experiment. Their mean age, mass and height were 24.7 yr, 78 kg and 178 cm respectively. A human pendulum served as impact modality (Lafortune and Lake, 1993). Entran accelerometers recorded the shock experienced by the shank and head of the subjects following impact of their foot with the force platform.

Three experimental interfaces were randomly assigned to each subject:

1)barefoot and force platform (BARE),
2)barefoot and force platform covered with an EVA mat (MAT),
3)shod and force platform (SHOE).

The EVA mat and shoe sole had identical hardness (55 Asker C) and thickness (24 mm). Fifteen impacts were recorded for each interface. The subjects were informed to keep their knee from flexing. Trials that demonstrated knee flexion were repeated. The 0.9 ms^{-1} impact velocity was achieved by pulling the pendulum 0.54 m away from the force platform. Exact impact velocities were monitored with a velocity transducer. (Celesco DV-301-0075). Signals were sampled at 1500 Hz and subjected to a fourth order low pass (200 Hz) digital filter. Foot contact was identified from the wall reaction force signal.

Transmissibility was assessed in the time (TD) and frequency (FD) domains. TD variables consisted of peak acceleration ratio, transient rate ratio and peak time delay ($DTIME=tpk_h-tpk_t$). The (FD) variables included mean power frequency shift (DMPF) and gain/attenuation profile (5.9-100 Hz) of the shank-to head shock transmission. The initial 256 points of the acceleration signals were used in the spectral analysis leading to a resolution of 5.86 Hz.

RESULTS

The human pendulum allowed for a tight control of the impacting velocity

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(Intra-and intersubject coeff. of var: 1.73 % and 0.99%). Between interfaces, the mean impact velocities deviated by less than 0.003 ms^{-1}. BARE interface induced significantly higher peak impact force (PKF) and rate of loading (ROL) than the MAT and SHOE interfaces (Table 1). The rate of loading was also higher in MAT than in SHOE interface. The results showed that the experimental interfaces had no significant influence upon the TD and FD transmissibility variables. On average, DTIME was 6.5 ms while peak acceleration and transient rate ratios were almost identical at 0.34 and 0.33 respectively. MPF was 4.4 Hz lower at the head than at the shank and, the gain/attenuation profiles for each interface indicated attenuation over the entire frequency range (Fig.1).

DISCUSSION AND CONCLUSION

The time and frequency domain results revealed that the human locomotor system was effective at attenuating the shock wave resulting from impact to the foot. In the time domain, the substantial shock reduction (66%) in absence of observable knee flexion implies that the locomotor system possesses other valuable shock attenuators. Within the range of impact severity simulated in this study, the results also indicate that these shock attenuators performed independently from impact magnitude. The frequency domain results further suggest that the abilities of the locomotor system to attenuate the shock wave and to filter out its higher frequencies were unaffected by the spectral components of the shock wave travelling up from the shank segment. Conversely, Shorten and Winslow (1992) reported that < 6 Hz frequencies were amplified and, signal attenuation above 10 Hz increased with increasing impact shock. The absence of knee kinematic adaptations in this study provides a potential explanation for these discrepancies.

The present results support the idea of a shock attenuating mechanism that maintains the shock to the head below a set threshold (Shorten and Winslow, 1992). The significant differences in head transient rate and mean power frequency suggest that the mechanism might be restricted to shock magnitude. However, our approach prevented kinematic adaptations which are integral components of the shock attenuation function of the locomotor system. The effects of kinematic adaptations upon shock attenuation will be examined in future studies.

TABLE 1. MEANS AND STANDARD DEVIATIONS

	BARE	MAT	SHOE
PKF (BW) a	1.87 (0.20)	1.64 (0.18)	1.48 (0.20)
ROL (BW/s) b	317 (57)	199 (26)	117 (44)
PKS (g) c	10.8 (2.1)	9.9 (1.5)	8.4 (2.5)
PKH (g)	3.3 (0.4)	3.2 (0.5)	2.9 (0.6)
RATIO	0.32 (.08)	0.33 (.06)	0.37 (.11)
DTIME (ms)	5.4 (3.8)	5.8 (5.0)	8.4 (3.9)
TRS (g/s) a	1326 (498)	925 (337)	742 (305)
TRH (g/s) a	387 (112)	296 (89)	218 (81)
RATIO	0.32 (.11)	0.34 (.11)	0.31 (.11)
MPFS (Hz) a	24.7 (2.9)	22.6 (1.9)	22.0 (2.4)
MPFH (Hz) a	20.6 (5.9)	17.7 (3.3)	17.9 (3.6)
DMPF (Hz)	4.1 (6.1)	4.9 (4.2)	4.1 (3.7)

a BARE > MAT and SHOE (α=0.05)
b BARE > MAT > SHOE (α=0.05)
c BARE > SHOE (α=0.05)

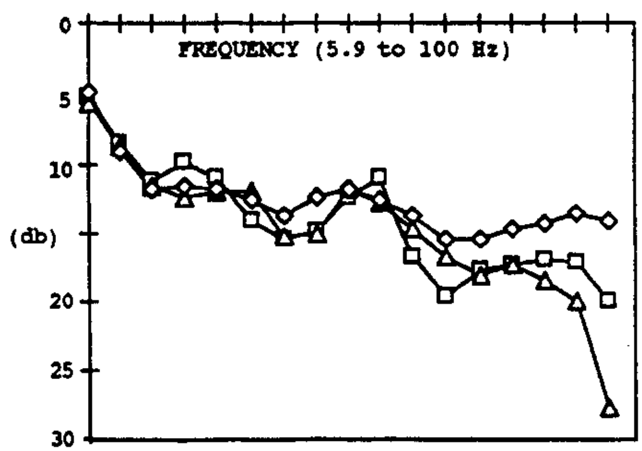

Fig 1. Attenuation vs frequency, data at 5.86 Hz intervals.

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RELIABILITY OF FORCE PLATFORM DATA IN THE ESTIMATION OF INSOLE SHOCK ATTENUATION

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INTRODUCTION

The "heelstrike transient" (fig. 1) is a short-lived force (typically 20ms) which occurs each time the heel of the foot contacts the ground during normal walking (Collins & Whittle, 1989). It probably results from the deceleration of the moving leg, with a resulting transfer of momentum from the leg to the ground (Whittle, 1993). It is most accurately measured using bone-mounted accelerometers (Light et al., 1980), but such an invasive procedure is clearly not suitable for routine use. Whittle (1993) demonstrated that the heelstrike transient could be studied using a force platform, providing the force platform had a high enough frequency response, data were sampled at a high enough rate, and smoothing was not employed.

The ability of athletic shoes to attenuate the transient force at impact during running has been studied extensively (Nigg et al., 1987). However, less attention has been paid to the heelstrike transient during walking, despite the fact that it has been implicated as a possible cause of degenerative joint disease in the lower limb (Radin, 1987). A number of insole materials have been developed which claim to reduce the magnitude of the heelstrike transient, and thereby possibly to offer some protection against the development of osteoarthrosis. Johnson (1986, 1988, 1990) developed a portable instrument for measuring the magnitude of the heelstrike transient, using an accelerometer strapped firmly to the lateral malleolus, and showed that it could be used to study the effects of different shoe and insole materials. The present study offers an alternative technique for the same purpose, based on the use of equipment found in most gait laboratories.

METHODS

Tests were carried out using two Bertec force platforms, data being sampled (without filtration) at a frequency of 1000Hz, using a Vicon gait analysis system. In this pilot study, measurements were made on a single subject, a 52 year old male weighing 680N, wearing "Oxford" shoes with a hard rubber heel. Two walks were made under each of eight conditions: without an insole, and with seven different types of insole or heel cup. Details of these insoles are unimportant - they were simply an assortment of different devices covering a range of shock attenuation properties. No formal effort was made to control the subject's cadence or stride length, although he attempted to walk in the same manner for each test.

The Vicon files of three-dimensional data (*.C3D) were transferred to a personal computer, where a specially-written program extracted the force platform data channels of interest and copied them into a file which could be accessed by the Microsoft Excel spreadsheet program, in which all further processing was performed.

The magnitude of the ground reaction force (GRF) was calculated from its three orthogonal components. Four variables were derived from the data, and examined for reliability. The GRF consists of the vector sum of the heelstrike transient and the "weight acceptance", which consists of the transfer of the body weight to the leg. The latter element was subtracted using the procedure suggested by Whittle (1993). A straight line was fitted between the point of initial contact and the "trough" in the GRF following the heelstrike. The value given by this straight line was then subtracted from the overall GRF at that instant

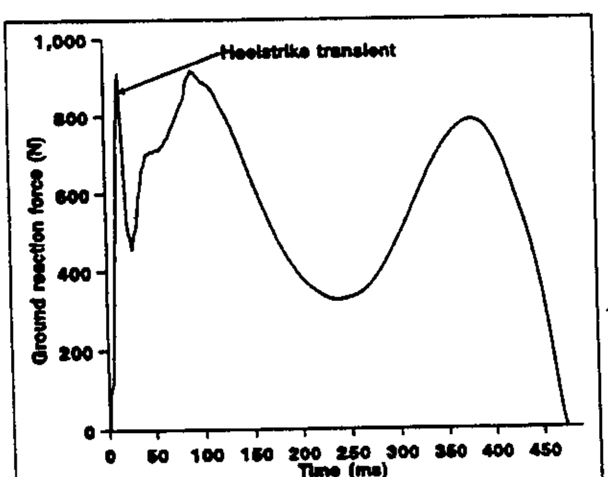

Figure 1: Magnitude of ground reaction force vector during stance phase of gait, showing heelstrike transient.

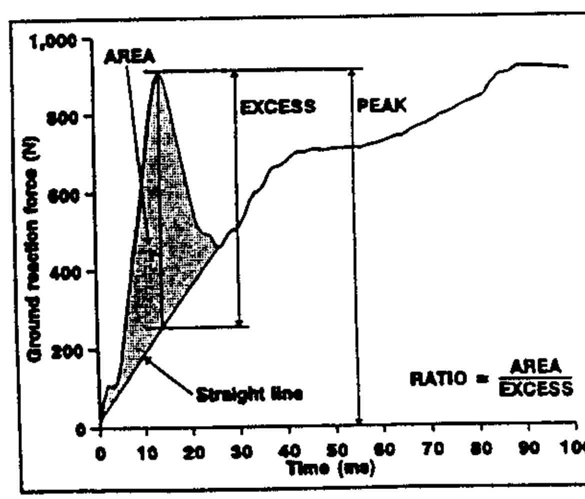

Figure 2: Detail of first 100ms of ground reaction force, showing definition of variables PEAK, EXCESS, AREA and RATIO.

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Table 1: Values of variables for each test condition with mean, standard deviation (S.D.), coefficient of variation (C.V.), and correlation coefficients CORR 1v2 and CORR LvR (see text).

Variable:	PEAK	EXCESS	AREA	RATIO
Units:	N	N	Nms	ms
No insole	908	639	6793	10.6
Insole A	782	547	6692	12.2
Insole B	818	566	6826	12.1
Insole C	688	464	6406	13.9
Insole D	745	506	6059	12.0
Insole E	707	487	5834	12.0
Insole F	606	359	5459	15.1
Insole G	643	363	6040	16.5
MEAN	737	491	6264	13.0
S.D.	98.1	96.6	496	2.0
C.V. (%)	13.3	19.7	7.9	15.0
CORR 1v2	0.69	0.66	0.16	0.93
CORR LvR	0.71	0.71	0.05	0.95

(fig. 2), the force in excess of the straight line being taken as the transient. The variables examined were 1) PEAK (absolute force magnitude at peak of transient); 2) EXCESS (force magnitude at peak of transient, minus value given by straight line); 3) AREA (product of time and force magnitude minus value given by straight line); 4) RATIO (AREA divided by EXCESS). It was decided that a realistic test for reliability would be firstly to compare replicate measurements made under the same test conditions, and secondly to compare the measurements made on one foot with the measurements made simultaneously on the other foot.

RESULTS

Table 1 gives the mean values from two walks, from both feet, of PEAK, EXCESS, AREA and RATIO for each of the eight test conditions (without an insole, and with seven different insoles). It also gives, for each variable, the Pearson product-moment correlation coefficient (r) between the data from the first and second measurements (CORR 1v2) and between the data from the left and right sides (CORR LvR).

DISCUSSION

There are a number of laboratory tests which measure impact attenuation in vitro, and the introduction of an American Society of Testing Materials (ASTM) standard on this subject is currently under discussion. However, it is unclear how such measurements relate to conditions in vivo. In the absence of a "gold standard" against which to compare the results, the present study therefore addressed only the reliability of force platform measurements of impact attenuation, and not their validity.

The only variable with a high reliability was RATIO, which had a correlation coefficient between the first and second tests of 0.93, and between the two sides of 0.95. Both PEAK and EXCESS had much lower correlation coefficients, in the region of 0.7 for both comparisons. AREA showed negligible correlations, with values of less that 0.2.

Whittle (1993) suggested that the weight acceptance phase of the GRF is limited to low frequency components, and that over the duration of the heelstrike transient it can be approximated by a straight line. Any force in excess this line constitutes the "transient", which is thought to be due to the loss of momentum from the decelerating leg. Since force can be defined as rate of change of momentum, the area beneath the force/time curve should equal the change in momentum of the leg. Assuming there is no difference in the gait pattern, the properties of the insole material should not alter the absolute magnitude of the momentum lost from the leg at heelstrike. This loss of momentum is represented by the variable AREA, which should thus be constant from walk to walk, with a coefficient of variation (C.V.) of zero. This was clearly not the case in practice, although the C.V. for AREA was less than that of the other variables. In contrast, the impact attenuation properties should affect both the height of the peak force (related to EXCESS) and the time course of the event (related to RATIO). In other words, a greater level of impact attenuation should result in a lower value for EXCESS, and a correspondingly higher value for RATIO.

The current data support the hypothesis that insole properties change the shape of the curve of the GRF more than they change the area beneath the heelstrike transient, and suggest that the GRF may thus be used to provide a reliable measure of transient attenuation. In the present study, the variable most sensitive to impact attenuation was RATIO, which relates to the time over which deceleration occurs. However, other methods of analyzing the data might prove even more effective, for example some form of frequency analysis, such as that used by Johnson (1986) for accelerometer data.

The more difficult problem of validity requires heelstrike attenuation measured from force platform data to be compared to some other measure. The "shock meter" (Johnson, 1990) is probably not accurate enough to provide a standard against which other methods can be judged, and it may require invasive studies, using bone-mounted accelerometers, to validate the method described here.

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THE EFFECT OF REARFOOT MOTION ON THE ATTENUATION OF THE IMPULSE WAVE AT IMPACT DURING RUNNING

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INTRODUCTION

With each foot contact, runners experience an impact force which is 2-3 times their body weight. Injuries such as stress fractures, shin splints, cartilage breakdown, low back pain, and osteoarthritis have been associated with these increased forces (Shorten & Winslow, 1992).

Energy absorption mechanisms, both internal and external to the musculoskeletal system, can affect the force at impact and the resulting impulse wave. Soft tissue, bone, articular cartilage of the joints and the motion of the joints may contribute to the attenuation of the impulse wave. External factors affecting the magnitude of the force at impact include running speed, footwear and the gradient of the running surface.

Few studies have focused on the attenuation properties of the foot. In theory, pronation of the subtalar joint should extend the duration of the impact phase thereby reducing the peak impact force at heel contact. In the present study, subtalar joint motion (rearfoot motion) was restricted to investigate the affects of reduced rearfoot motion on the impulse wave associated with impact.

METHODS

Fifteen subjects with informed consent and who were free of any injuries at the time of data collection participated in this study.

EGAXT accelerometers were used to record the magnitude and time duration of the impulse wave experienced by the lower extremity with and without a medial wedge in the running shoe. They were interfaced to a light weight portable amplifier/ telemetry system (2 kg.), which was carried in a backpack securely attached to the subject. One accelerometer was attached to the skin overlying the proximal, medial-anterior aspect of the tibia, a second to the calcaneus and a third on the heel of the shoe.

Four reflective markers were attached to the subject's posterior leg and foot to define the relative angle between the calcaneus and the leg during the running trials. This angle represents the angle of the subtalar joint (Clarke, Frederick & Cooper, 1983). The subjects were videotaped from the rear during each trial to record the position of four markers placed on the subjects. The markers were used to calculate the rearfoot motion for subjects running with and without the wedge. A synchronizing system (Kristal Instrument Corp.) was used to align the heel accelerometer signal with the video data in the time-domain.

The subjects ran on a treadmill at a cadence of 160 steps/min at a comfortable speed under two conditions. One running condition included running with a medial wedge inserted into the running shoe (NIKE Air Huarache) to limit rearfoot motion. The other condition included running with a flat insert (7 in. long, 0.5 cm. thick) placed in the shoe, made of the same material as the wedge. The subjects were familiarized with the treadmill, the wedge and the cadence during a 5 minute warm-up period. Each running trial was approximately 67 seconds with data collection occurring during the final 7 seconds.

Rearfoot motion was calculated for both conditions for the initial 15% of the stance phase. Accelerometry signals were A/D converted on-line at a rate of 3787.88 Hz. using Bioware (Kistler Instrument Corp., Amherst, NY 14228). Accelerometry data were analyzed in both the time and frequency domains. In the time-domain, the magnitude of the acceleration peaks of the heel, calcaneus and tibia accelerometry records and the time-to-peak value of the tibia were determined (Figure 1). The accelerometer signals were then transformed from the time-domain to the frequency-domain using a Fast Fourier Transform (FFT). The predominant frequency of the impulse peak occurs between 12-25 Hz. (Shorten & Winslow, 1992). The frequency of the impulse peak and it's magnitude were compared between the two running conditions (Figure 2).

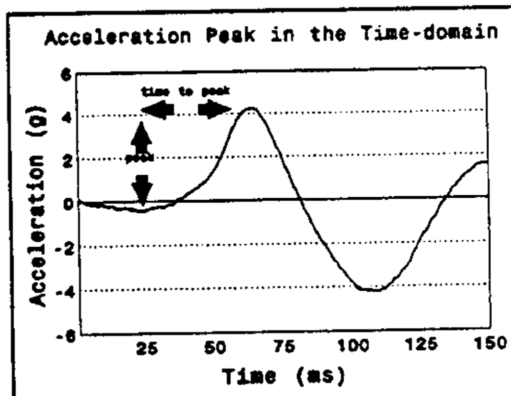

Figure 1: Schematic diagram representative of the peak accelerations and the time-to-peak values for one subject.

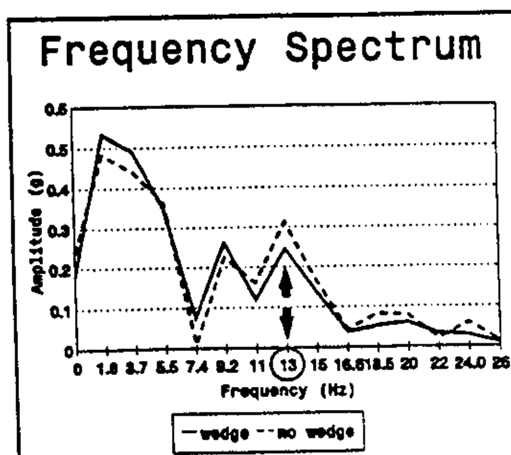

Figure 2: Frequency spectrum of the tibia acceleration signal shown in Figure 1.

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A paired t-test ($p \leq 0.05$) was used to establish the difference in the rearfoot motion between the two conditions (wedge -v- non-wedge). The time-domain peak accelerations of the heel, calcaneus and tibia accelerometers, the time-to-peak values of the tibia accelerometer, predominant frequency of the impulse wave at the tibia and the amplitude at that frequency were also compared between conditions.

RESULTS

Restricting rearfoot motion did not result in an increase in the magnitude and time-to-peak value of the impulse wave at the tibia. Both the time-domain and the frequency-domain results yield no significant differences between the two running conditions, (no-wedge and wedge), examined in this study (Table 1).

DISCUSSION

The collision between the foot and ground at heel contact results in a large force acting over a short period of time as the momentum of the body is arrested. Rearfoot motion is thought to attenuate the magnitude of the impact force at heel contact by increasing the time interval over which the impact force acts on the body. Three explanations may help interpret these results. First, the results might suggest that the rearfoot movement, although limited during the wedged condition, was still sufficient for shock attenuation. The motion immediately after heel contact may be enough to increase the duration of the impulse and thereby decrease the magnitude of the impact force and thus the accelerometers at the tibia would detect no changes in the impulse wave.

Secondly, sagittal plane kinematics may have been altered to compensate for a reduction in rearfoot motion. Joint motion can reduce the effective vertical stiffness of the body and lessen the transmission of mechanical shock up the body. Joint motion has been found to decrease the impulse wave measured between the tibia and the head (Paul et al., 1978) and the motion of the ankle, knee and hip joints may have been altered to some degree by the subjects of this study to compensate for the loss of rearfoot motion.

A third possibility is that there may have been a change in the amount of co-contraction of muscles at the joints at impact. A change in position of the foot when the wedge is introduced may alter the feedback of the muscle receptors and change the stiffness of the joint.

Excessive pronation has been associated with injury during running. Shoe and commercial orthotic design have focused on controlling subtalar joint motion, primarily pronation. Since pronation is a naturally occurring movement and is thought to aid in shock absorption, controlling pronation may have implications on the absorption of shock. Studies involving shoe design in the control of rearfoot motion found that midsole hardness and heel flare affect maximal rearfoot motion. These studies restricted rearfoot motion by 11.7% a much smaller amount than the current study. Previous studies involving commercial orthotics have limited maximum pronation in the range of, 10 - 36.4%. The present study showed no significant increases in the impulse wave at the tibia using a medial insert which restricted rearfoot motion in the initial 15% of stance phase by an average of 62%. Shoes and orthotics designed to control rearfoot motion therefore should have limited affect on the magnitude and the timing of the impulse wave resulting from heel contact.

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ACKNOWLEDGEMENTS

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Special thanks to NIKE Inc. for donating the running shoes used in the study and to Kristal Systems Inc. for the use of the synchronization system.

Table 1: Mean time and frequency domain values for the no-wedge and wedge running conditions.

	No wedge	Wedge
Rearfoot angle motion (degrees)*	12.60 (8.46)	4.79 (4.17)
Peak heel acceleration (g)	2.891 (1.045)	2.783 (1.031)
Peak calcaneus acceleration (g)	3.936 (1.572)	3.818 (2.244)
Peak tibia acceleration (g)	3.764 (0.891)	3.806 (1.069)
Time-to-peak tibia acceleration (ms)	.044 (0.009)	.047 (0.010)
Predominant freq. of tibia impulse wave (Hz)	15.7 (1.5)	15.9 (1.7)
Amplitude of predominant tibia freq. (g)	.360 (.160)	.376 (.160)

* Significant Difference ($p \leq 0.05$)

IMPACT SHOCK ATTENUATION AND STRIDE FREQUENCY RELATIONSHIPS

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INTRODUCTION

The loads resulting from repeated impacts during the support phase of a running cycle have been implicated in overuse injuries and degenerative diseases (Simon et al., 1972; Radin et al., 1972; Voloshin and Wosk, 1982). These loads are transmitted throughout the skeletal system and thus affect all segments of the body. Increases in impact shock can result from an increase in running speed (Hamill et al., 1983), from increasing stride length (McMahon et al., 1987) or from running downhill (Hamill et al., 1984) The cost of shock attenuation may be seen in increased oxygen consumption (McMahon et al., 1987). It has also been shown that a deviation from the preferred stride frequency (PSF) increases oxygen cost. (Holt et al., 1991). It would appear, therefore, that impact shock attenuation is an important factor on which individuals optimize.

While the importance of shock attenuation may be a critical factor in the susceptibility of an individual to injury, there may be other reasons for optimizing on this factor. For example, since shock is transmitted throughout the skeletal system to the head, stability of the head during impact may be important to maintain consistent information from the vestibular and visual systems.

REVIEW AND THEORY

The notion that humans are self-optimizing machines is supported by research findings in biomechanics (Kugler and Turvey, 1987; Holt et al., 1990). A model that has proved effective in describing this self-optimization during locomotion is a pendular model, the force-driven harmonic oscillator (FDHO). The FDHO requires a periodic forcing function to maintain its oscillations with a minimum force at the resonant frequency to maintain its oscillations. Holt et al. (1991) illustrated that oxygen cost is optimized at the PSF during locomotion. Another optimizing factor may be the transmission of the impact shock to the head. The purpose of this study was, therefore, to determine if impact shock at the head was minimal at the PSF, and thus the resonant frequency of the FDHO during running.

PROCEDURES

Five young, healthy adult males served as subjects in this experiment after signing informed consent forms in accordance with University policy. All subjects were free of lower extremity injury that may have affected their performance in this study and all were experienced treadmill runners. They ranged in body mass from 55.0 kg to 83.9 kg (mean = 74.4 ± 12.1 kg) and in stature from 1.75 m to 1.96 m (mean = 1.84 ± 0.75 m).

Shock attenuation was determined using two 1.7 gram Kistler accelerometers interfaced to a microcomputer via an A/D converter. Data sampling was accomplished at 1000 Hz. One of the accelerometers was mounted on the anteromedial distal aspect of the left tibia with an elastic strap (Valiant et al., 1987). The other accelerometer was mounted on the frontal bone of the head of the subject also using an elastic strap (Wosk and Voloshin, 1981). To record the instant of impact, a force transducer was placed under the treadmill and interfaced to the computer via the A/D converter. The sensitivity was adjusted such that any contact with the treadmill was detected by the computer.

A Gould 9000IV Computerized Metabolic Cart was used to collect the metabolic data. The metabolic cart was calibrated before each test session. Heart rate was monitored with a Vantage Performance Monitor telemetry system. Oxygen consumption and heart rate values were sampled every 20 s.

Each subject attended one experimental session. The session began with the subjects warming up by running on a heavy-duty motorized treadmill. After the warm-up was completed, each subject chose their preferred running speed (PRS) following a protocol used by Holt et al. (1991). The PRS ranged from 5.0 to 5.6 mph. When the PRS was established, the preferred stride frequency (PSF) at that speed was determined. The PSF ranged from 76 to 83 strides/minute. The conditions of PSF, +20%, +10%, -10%, and -20% were then calculated resulting in five experimental conditions including the PSF. Each subject ran at their PRS in each of the five conditions until steady state was achieved. The five conditions were balanced to reduce order effects. At steady state, five 2-second trials of accelerometry data were collected. Two footfalls were collected and subsequently analyzed in each trial.

The accelerations data were evaluated to determine the peak acceleration during the contact phase of the left footfall. for both the head (peakH) and tibial (peakL) accelerometers. Since two footfalls were present in each trial, a 10 footfall mean for each subject/condition was calculated. A transmission ratio, indicating the shock transmitted from the tibia to the head, was calculated. During the steady state period of each condition, the average value of oxygen consumption and heart rate during the final two minutes were calculated.

The mean peakL and peakH acceleration values for all subjects/conditions were evaluated using a repeated measures ANOVA and a trend analysis. The criterion alpha level was 0.05.

RESULTS

Mean values for oxygen consumption and heart rate are presented in Figure 1. ANOVA on the oxygen consumption and heart rate data revealed significant results among the locomotor frequencies ($p < 0.05$). Post hoc analyses on both parameters revealed that the -20% condition was significantly different from the PSF and the +20% condition ($p < 0.05$). Trend analyses indicated a significant quadratic trend for VO_2 ($R^2 = 0.99$) and for heart rate ($R^2 = 0.97$).

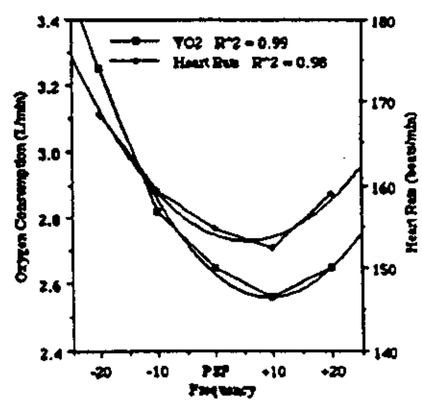

Figure 1. Metabolic parameters as a function of stride frequency.

Mean values for peakH and peakL are presented in Figure 2. ANOVA revealed statistically significant differences among the frequency conditions ($p < 0.05$). Post hoc analysis indicated that the two low frequency conditions were significantly different than the PSF and the two high frequency conditions ($p < 0.05$). Trend analysis indicated a significant linear trend ($R^2 = 0.83$) with a negative slope. There were no statistically significant differences among the frequency conditions for peakH.

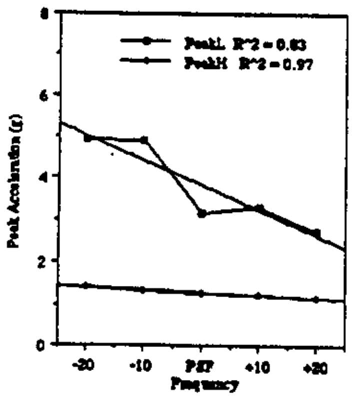

Figure 2. Peak accelerations for the leg and the head as a function of stride frequency.

The ratio of the peakH/peakL produced results similar to the peakL parameter. That is, significant differences among the conditions and a significant linear trend ($R^2 = 0.75$) with a positive slope (Figure 3).

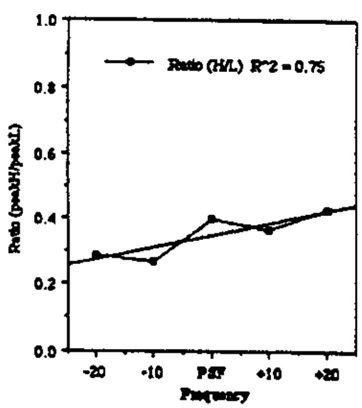

Figure 3. PeakH/peakL ratios as a function of stride frequency.

DISCUSSION

The oxygen consumption and heart rate data are similar to those presented by Holt et al. (1991). These data reinforce the notion that there is a frequency that can be associated with a minimal metabolic cost. In the FDHO model, this frequency is the resonant frequency that can be equated to the PSF (Holt et al., 1990). In this study, the minimal cost is not at the PSF but at a frequency slightly greater than the PSF (Figure 1). The indication here is that the observed PSF was not the FDHO resonant frequency

Shorten et al. (1992) reported an increase in peak leg acceleration as running speed increased. In this study, running speed remained constant for each subject but stride frequency and hence stride length changed. As stride length increased, impact shock measured at the leg increased indicating that the input force increased.

Regardless of the input force, the peak acceleration at the head remained constant. The transmission of the input shock to the head, as indicated by the peakH/peakL ratio, thus varied in order to keep the head stable. The transmission ratio, however, did not follow the typical quadratic function that is associated with the FDHO. These data suggest that stability of the head during locomotion may be a critical optimizing factor for the system regardless of the frequency.

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MODELING LANDING IMPACTS

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INTRODUCTION

Macauley (1987) has stated that the human body is a complex machine and that its functional responses to impact are not well understood. A plethora of current landing literature documents the magnitude of forces associated with impact (eg., DeVita and Skelly, 1992; McNitt-Gray et al., 1993; Schot and Dufek, 1993). The impact adaptation or modification process related to performance is yet unclear.

A series of rigorous statistical procedures were invoked to predict impact forces associated with a range of landing performances varied by knee joint flexion. Results of the two group models averaged 74.7 and 98.6% explained variance to which two and zero individual subjects subscribed, respectively. These results suggest the potential error in generalizing such an overall (group) performance model to all members of an activity group.

REVIEW AND THEORY

Using correlation techniques and a large sample size (150 trials per subject x 10 subjects), Dufek and Schot (1993) suggested that discrete variables (kinematics and joint kinetics) best represented the landing impact phenomena. Furthermore, landing technique characterized by joint kinematics has been shown to influence impact (DeVita and Skelly, 1992; Dufek and Bates, 1990). However, the concept of heterogeneity of performance (expanding the performance outcome range) has yet to be incorporated. By eliciting a greater range of performances (impact forces and associated kinematic patterns) within a given skill, one can maximize the ability to identify performance relationships. The purpose of the study was to predict landing impact forces from a heterogeneous sample of performance trials.

PROCEDURES

Six male volunteer subjects granted written consent and then each performed three conditions of 0.6 m step-off landings onto a dual AMTI force platform system (one foot per platform). The first condition (NOR) consisted of 30 landing trials using a self-selected technique. For the second and third conditions, consisting of 10 trials each, subjects were instructed to land as softly (SO) or stiffly (ST) as possible using only the knee joint to manipulate performance outcome. Right side vertical ground reaction force (vGRF) values were obtained (1000 Hz) and evaluated during the impact phase of landing (100 ms post-contact). Two vGRF variables were identified, normalized to body mass and used for subsequent analysis: first (F1) and second (F2) maximum vGRF values. Kinematic information from the right side was obtained using a high speed real time video acquisition system (Motion Analysis, 200 Hz). Kinematic data were processed (digital filter, 10 Hz cut-off), synchronized to the force platform data, and 30 kinematic independent variables (IVs) were computed using laboratory software (Table 1). The kinematic variable set was then systematically reduced to produce an IV set representative of landing performance across all subjects and used as input to predict F1 and F2, on a group and individual-subject basis.

RESULTS

Within-subject F1 and F2 condition comparisons were first evaluated using the Model Statistic single subject technique (Bates et al., 1992). Results identified 97.2% significant (α = 0.05) condition differences overall indicating that all subject-conditions except S2-F1 (NOR vs SO) elicited performance results that were different. These results suggest the landing conditions (SO, NOR, ST) may represent unique skills. This question was examined by computing the effective curve correlation coefficients using a Shapiro-Wilks' normality test. The percent explained variance associated with these combined condition correlations across subjects averaged 88.8 (F1) and 87.4 (F2) and appeared to represent expansion along a continuum (ie, produce heterogeneity within a skill). This interpretation is graphically illustrated with randomly selected trials from one experimental subject (Figure 1).

Within-subject correlation matrices were computed among F1, F2 and the IV set. All correlation coefficients not significantly different from zero (α = 0.01) were first eliminated from the analysis. Of the remaining IVs, those identifying >= 25% common variance across at least four subjects for F1 and F2, respectively, were then retained for the prediction models and are identified in Table 1.

Table 1. Kinematic variables examined and prediction model compositions.

Ankle	Knee	Hip
θ contact *	θ contact *#	θ contact
θ event	θ event *#	θ event #
ω contact	ω contact #	ω contact
ω event	ω event	ω event
Vv contact *	Vv contact *	Vv contact
Vv event	Vv event	Vv event #
Hv contact	Hv contact	Hv contact
Hv event *	Hv event	Hv event #
Va contact	Va contact	Va contact
Va event #	Va event #	Va event *#

Note: contact = landing contact; event = time of DV occurrence; Vv = vertical velocity; Hv = horizontal velocity; Va = vertical acceleration; * = used in F1 prediction model; # = used in F2 prediction model.

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A stepwise multiple regression procedure was incorporated with a limit for overall model significance set at $\alpha = 0.05$. A group model using subject-condition mean values was first computed for F1 and F2. The resulting regression equations were: [F1 = (Hip Va@F1 * 0.002) + 13.032]; [F2 = (Ankle Va@F2 * 0.003) - 20.56] and accounted for 74.7 and 98.6% of the respective variance. Similarly constructed within-subject models were then computed using each set of 50-trial values. Results of the subject models are comparatively summarized in Figures 2 and 3. For F1, two subjects (S2, S3) generally performed like the group model. A second pair of subjects (S4, S5) were characterized as ankle joint dominant. The primary predictor for S1 was knee joint vertical velocity at contact. No significant model was produced for S6. For F2, no subject displayed the dominant characteristics of the ankle joint vertical acceleration at F2, as per the group model. F2 was best predicted for four subjects with hip joint vertical velocity and acceleration values, accounting for an average of 84.8% explained variance across these subjects. The remaining two subjects (S1, S4) exhibited models that identified kneeθ, kneeω and knee vertical acceleration accounting for an average of 76.0% explained variance.

DISCUSSION

The process of IV identification for inclusion into the model was stringent. The methods employed produced a trial:IV ratio of 7:1 and 9:1 for F1 and F2, respectively. These ratios are much more conservative than the often recommended 5:1 ratio. It is acknowledged that the "pre-selection" of IVs using a procedure such as a correlation matrix has the potential of applying only to the study sample from which it was generated. In this study, however, a more critical outcome was observed--that of the inaccuracy of the subset (subject) models with respect to the overall (group) models. This finding, perhaps attributable to the multiple degrees of freedom within the human system, suggests that biomechanically, one should be concerned not only about generalizing empirical results to an unmeasured "population", but also with generalizing all subject performances to the group or average performance outcome.

SUMMARY

Group and individual multiple regression models were computed to predict F1 and F2. The average explained variance for significant group and single-subject models was 74.7 and 77.5% (F1) and 98.6 and 90.3% (F2). Unique IV entry to the models should caution researchers as to any functional interpretation of the task and suggests a possible technique to categorize subjects based upon "strategies".

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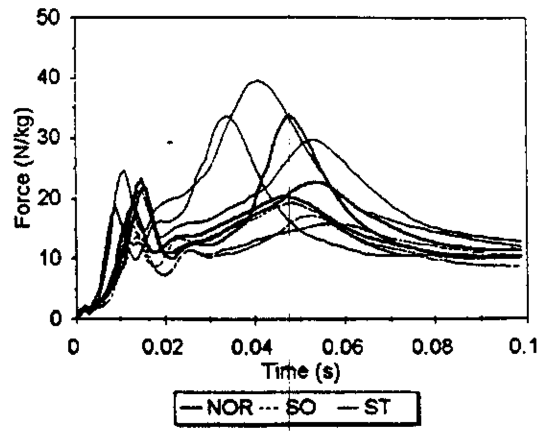

Figure 1. Randomly selected vGRF-time histories for S1 illustrating the same "skill" across conditions.

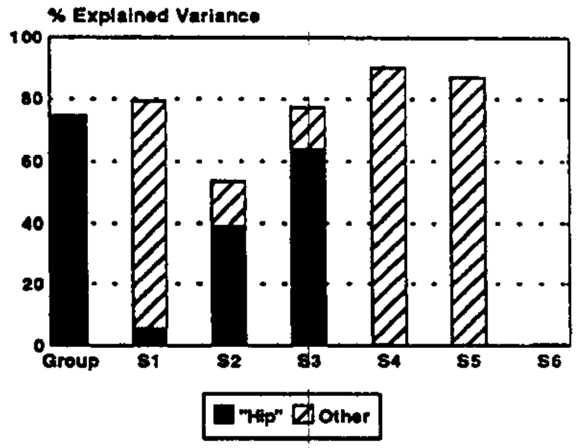

Figure 2. F1 Prediction Models illustrating comparison between group (hip joint vertical acceleration at F1) and single-subject IV model composition.

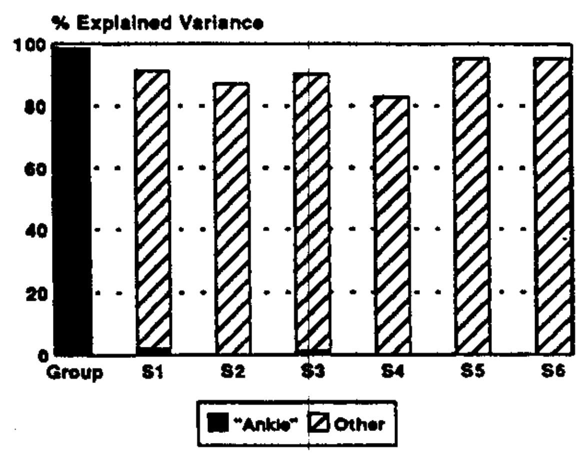

Figure 3. F2 Prediction Models illustrating comparison between group (ankle joint vertical acceleration at F2) and single-subject IV model composition.

RUNNING IMPACT FORCE MODELING

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INTRODUCTION

Impact forces (IF) have been implicated as a major cause of running injuries (James, et al., 1978; Nigg, 1985; Perry, 1983). Individual adaptations to factors influencing IF are not well understood.

Impact characteristics were evaluated via group and individual-subject regression models. Results identified a single predictor for the group models while individually, an average of 3.8 unique variables were identified. Three categories of impact accommodating strategies were observed across all subjects.

REVIEW OF LITERATURE

It has been suggested (Clarke, et al, 1983; Nigg, et al., 1987) that variations in the motor program can be used to accommodate to environmental changes introduced during running. Changes including footwear, surfaces and speed as factors have been documented to influence IF values (Bates, et al., 1983; Clarke, et al., 1983; Bowers and Martin, 1974; Bates and Stergiou, 1993; Hamill, et al., 1983)
It has also been suggested that response patterns to these different environmental factors can range along a continuum from completely ignoring (Newtonian or mechanical response) to total accommodation (neuro-muscular response). The former response will produce varying IF values while the latter will result in consistent IFs across conditions. One approach to gaining a better understanding of these phenomena is to determine the relationship between IF and other descriptive parameters. The purpose of this study was to predict and evaluate group and individual subject IF values during heel-toe running from selected kinematic parameters.

PROCEDURES

The research protocol used attempted to create a more heterogeneous IF sample for each subject by imposing moderate perturbations on normal stride (NS) length by requiring specified trials to be slightly shortened (understride, US) or elongated (overstride, OS). This procedure obviously has the potential to create three independent conditions, each a separate skill. If successful, however, creating a more heterogeneous sample enhances modeling capabilities for the techniques used (multiple regression).

Six college-age males volunteered as subjects. All subjects ran with a heel-toe pattern at their preferred running pace, while being monitored by a timing light system (+/- 5% target pace). They performed 30 trials of normal running an 20 trials for each of the OS and US conditions. Right side vertical GRF data were collected using a force platform system (1000Hz), while both rear and sagittal view kinematic data were acquired using an automated tracking system (200Hz). The spatial coordinates obtained were scaled and smoothed using a low-pass digital filter with a selective cut-off algorithm based on that of Jackson (1979).

The first step in the analysis was to evaluate the IF condition mean values for the group and individual subjects using ANOVA and an individual subject technique (Model Statistics; Bates, et al., 1992) respectively. Next, the combined IF data sets for each subject were evaluated using a curve correlation technique to determine whether or not the three conditions could be combined. The final step was to generate the IF regression models for each subject (individual trial data) and for the group (subject mean data). Twelve independent variables (Table 1) were identified based upon the literature and pilot work for the modeling process. The variable set was limited to 12 in order to maintain a minimum 5 to 1 ratio between the number of trials and independent variables. The results of these analyses were compared to identify performance strategies.

RESULTS

The group and individual subject mean IF values are given in Table 2. The group results indicate no difference between the US and NS condition while the OS value was significantly greater. Single subject comparisons resulted in 88.9% significant differences. Three subject strategies were observed none of which were the same as the group result (S1, S2, S4; S3, S5; S6).

These results initially suggest that the conditions represent different skills and should not be grouped for further analysis. To verify or reject this conclusion the grouped data sets were evaluated using a curve correlation technique that compared the grouped distribution with a normal distribution. The results of this analysis are given in Table 3 (Normal). The mean shared variance for all subjects was 88.0% (r=0.938) suggesting a high degree of similarity between the combined individual subject data sets and a normal distribution. Based upon a common variance criterion of 80% (r=0.895) all sets of subject data were retained for further analysis.

DISCUSSION

The results from the regression analyses (Table 3) illustrate that the group regression identified a single variable ($K\Theta_r$) for best predicting IF, accounting for 66.6% (r=0.816) of the variance. This value was slightly greater than the mean subject value of 65.0% (r=0.806) but less than three of the subject models. The subject models included an average of 3.8 variables with explained variance ranging from only 9.4%

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configuration, until the end of the range of motion is encountered. The program then produces an animated graphical output, showing the positions of the ligaments, contact points and femoral and tibial surfaces as the model moves through its range of motion.

Results

A preliminary set of anatomical parameters was obtained from Zavatsky's study of joint ligament mechanics [7]. The motion of the model was determined by the computer program. Figure 2 shows a sagittal plane view of the model in (a) full extension, and (b) 70 degrees of flexion. This clearly shows that both contact points have moved posteriorly on the tibial plateau with flexion. This figure also shows that the lateral contact point is posterior to the medial contact point, which demonstrates that the tibia has rotated internally about its long axis relative to the femur. Rotation of the tibia is plotted against flexion angle in Figure 3.

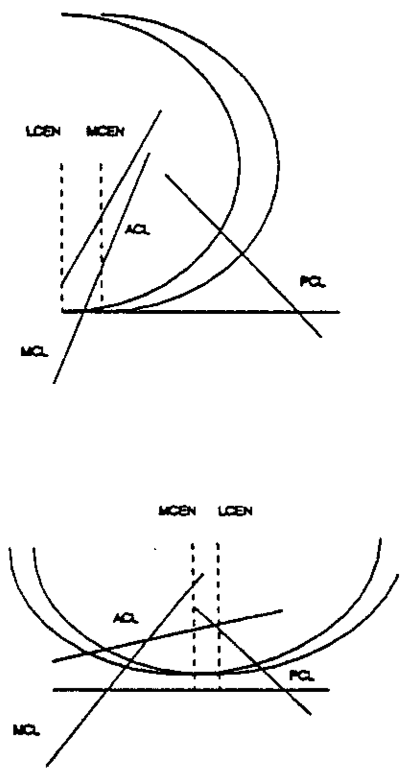

Figure 2: Sagittal plane view of the model in a) full extension and b) 70 degrees of flexion.

Discussion

The two significant motions predicted by the geometric model have been observed in studies of passive knee flexion. Goodfellow and O'Connor [2] studied the translation of the contact points in the knee during passive flexion by observing the motion of meniscal bearings in a knee prosthesis. They found consistent posterior translation of both the medial and lateral contact points on the tibia as the knee was flexed. O'Connor and Biden *et al* [4] studied the motions of the knee in passive flexion using cadaver specimens in a 6 degree-of-freedom test rig. They observed significant internal rotation of the tibia with knee flexion. The average results for eight knees from O'Connor and Biden's work are compared to the results of the mechanism model in Figure 3.

The preliminary set of anatomical parameters used in this study has produced a model whose kinematics are qualitatively similar to the knee joint. To validate the model quantitatively, the motion of a knee specimen should be compared to motion predicted by a geometric model using anatomical parameters obtained from that specimen. It is anticipated that these experiments will be undertaken soon.

This model also demonstrates the important role the ligaments play in guiding the motion of the knee joint. Some authors contend that the knee ligaments have no mechanical function. It is clear from Equation (1) that the ligaments are required to produce the one DOF motion observed in studies of knee joint motion. It is important to note that a change in the position of the orientation or attachment of a ligament will affect the kinematics (and by extension, the mechanics) of the entire joint. This should be a consideration when planning surgery that disrupts or replaces the ligaments.

Conclusions

The three-dimensional geometric model in this study qualitatively describes the significant features of knee motion. Because it predicts motions that cannot be accounted for by planar geometric models, it will provide more accurate values for the position and orientation of the anatomical structures of the knee at a given flexion angle. The model clearly demonstrates that the ligaments and contact points work together to control joint motion, and that, in passive flexion, the knee is a one DOF mechanism.

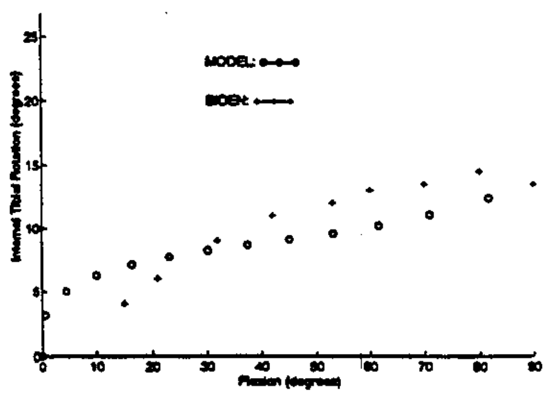

Figure 3: Tibial rotation with knee flexion.

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MEASUREMENT OF FOOT STIFFNESS DISTRIBUTION AND COMPUTATIONAL SIMULATION OF PLANTAR PRESSURE DISTRIBUTION

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INTRODUCTION

It is widely accepted, that increased plantar pressure represents a risk factor for lesions of the diabetic foot. Although the complexity of the interaction of high plantar pressure and ulceration is still not fully understood, it is obvious that reducing the pressure peaks is beneficial in diabetic foot care. Elastic, individually shaped insoles are used to achieve this reduction during walking. However, no technique is available to find an optimized design of these insoles. In our work, a procedure to numerically predict the evolution of the plantar pressure distribution during walking is presented. As basis for all mathematical treatment an adequate biomechanical model has to be established.

Most of the models published so far are structure-mechanical models, consisting of structural components like bones, ligaments and joints and are able to calculate internal forces or deformations. Among these, different mechanical approaches are found: static models, some of which also can treat dynamic motion by a quasi-static consideration (MORLOCK, PROCTER); elasto-static models including the compliances of joints, ligaments, etc. (SALATHE, SIMKIN); elasto-static models in a quasi-static imitation of dynamic motion (SCOTT 1993); models based on inverse dynamics which employ gait-analysis (APKARIAN, SCOTT 1990) and direct dynamic models (PANDY).

After all, it cannot be expected that structure-mechanical models will lead to a continuous pressure distribution at the sole of the foot. This requires a continuum-mechanical approach. Few attempts have been made to apply the method of finite elements (NAKAMURA, CAVANAGH). However, the complexity of the foot and the lack of reliable material data limit the potential of this approach. In our project a different type of continuum-mechanical model was developed, specially designed to predict the plantar pressure distribution. It should be mentioned that recently SHORTEN published a multiple-element model of a running shoe that essentially incorporates a similar mechanical description.

BIOMECHANICAL MODEL

A multiple-element, quasi-3dimensional, non-linear elastic model is established as sketched in Fig.1. The bony rear foot consisting of calcaneus and naviculare is assumed a rigid body, to which a coordinate system is attached. Presuming the foot resting unloaded on a horizontal plane, the x-y-plane is defined to be horizontal, with the y-axis in posterior-anterior direction. The sole of the foot is divided into a 2dimensional arrangement of prismatic elements. Each of these multiple elements represents a non-linear spring with 1 DOF, deformable in the vertical (i.e. z-) direction. The upper end of every spring is supported in the x-y-plane of the rear foot coordinate system, whereas the lower end matches the plantar surface and allows for its elastic deformations.

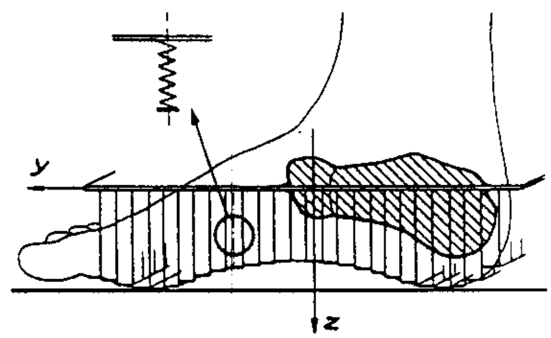

Fig.1: Multiple-element model

In order to properly reflect the elastic behavior of the actual foot, the properties of every spring element have to be equivalent to the actual compliance at the corresponding location which of course is a superposition of all contributing joint-, ligament- and soft tissue compliances. Thus, by the lengths of the multiple spring elements the shape of the plantar surface is represented, and by the stiffness characteristics of the spring elements the elastic behavior of the foot along the plantar surface is represented.

With no additional mechansism included, the model is valid only for the description of motion without significant pronation/supination. It is clear that no information about actual internal forces can be gained by this model. All required model data can be measured individually and in vivo.

MEASUREMENT SETUP

A special measurement setup is designed to enable an acquisition of the required stiffness characteristics. A stiffness characteristic of a non-linear elastic element is nothing else than a diagram of the load-displacement relationship. Thus, the concept is to simultaneously measure both the load and the displacement histories of all of the previously introduced spring elements during a representative quasi-static loading of the foot.

An EMED-SF pedography system with 2 sensors per cm^2 and a memory of 150 complete pressure pictures is used for measuring the load histories. Favorably, the 2dimensional arrangement of elements in the model will coincide with the 2dimensional array of load sensors of the pedography plate. A special, PC-driven device is developed and named "pedomech" system, which is connected to and synchronized with the pedography system to measure the displacement histories. It essentially employs 3 electro-mechanic displacement transducers which act in vertical direction –

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sufficient to monitor the rigid body motion of the rear foot (i.e. the current location of the rear foot coordinate system). Knowing the undeformed shape of the sole of the foot from a prerequisite geometrical measuring and knowing that the lower boundary of the foot in any load situation is given by the horizontal pedography plate, the deformation-displacement at any location at the sole of the foot can be deduced from the rigid body motion of the rear foot. A specific software is written to perform the necessary data manipulations and calculations.

Special concern deserves the linkage from the displacement transducers to the bony rear foot. A shaped foot sleeve which conjugates with particular bone projections is closely fit to the rear foot. By means of thin, vertical rods which compensate for horizontal perturbations the transducers are connected to the foot sleeve, and by 3 three-axial guideways they can be fixed in that position. With this method the crucial problem of skin-displacement errors (MASLEN) can be overcome.

At the beginning of the measurement the bare foot, wearing the foot sleeve that is connected to the transducers, rests unloaded on the pedography plate. Then, within some seconds, the test person shifts its weight entirely on that foot while both, the evolution of the pressure distribution, and the motion of the rear foot due to elastic deformations of the foot are measured and stored to a PC. By moving the body forward and backward during the measurement the full load range that will appear in dynamic motion can be covered.

MEASUREMENT RESULTS

A single measurement yields about 200 stiffness characteristics. These data define the properties of the multiple spring elements introduced in the biomechanical model and will serve as input data for the computational simulation.

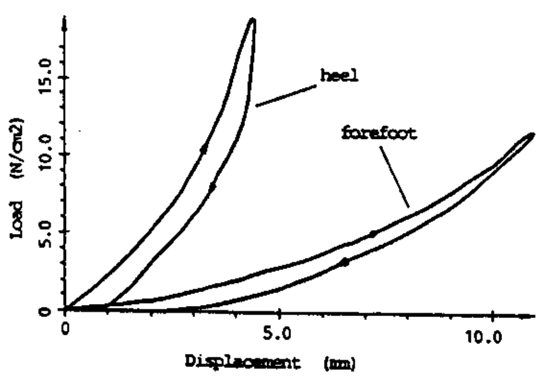

Fig.2: Measured stiffness characteristics

Fig.2 shows two selected curves, one for a location at the heel and one from the ball of the foot. The progressive load-deformation relationships reveal the nonlinear stiffness which is expected of biological tissues. It is significant that the heel behaves much stiffer than the forefoot. However, this does not imply that the heel was harder then the ball of the foot. Rather the equivalent stiffness of a forefoot spring element also accounts for the compliances of the tarsal joints.

SIMULATION

From a mechanical point of view applying a given load to the multiple-element model represents a non-linear elasto-static problem including a contact problem. The latter one corresponds to the activation of indiviual springs only if the eventual gap between the plantar and the supporting surfaces is closed. Using the principle of virtual work the stiffness matrix is derived, in which both the values of the elements and the matrix-rank are deformation-dependent. By employing the Pure-Newton-Raphson method a numerical algorithm is established to find equilibrium, and implemented in a computer program. The required input data are: the undeformed shape of the plantar surface, the stiffness characteristics of the spring elements and the load. In a quasi-static description also time-dependent loading like in walking can be considered. By an additional measurement the time histories of magnitude and location of the ground reaction force are acquired from the test person walking across the pedography plate. These data are input into the simulation program as load history and the successive equilibrium states are calculated by the incremental treatment of the Newton-Raphson method.

At the present stage of this study the contact boundary in the simulation is assumed a rigid, horizontal plane. Comparisons of the predicted pressure distributions to pedographic measurements are presented.

DISCUSSION

The presented simulation algorithm based on a specific modelling of the foot proves suited to predict the plantar pressure distribution for simplified boundary conditions. Future work will be dedicated to adding a model of the shoe and elastic insole, thus creating a tool to design orthopedic footwear with regard to the anticipated in-shoe pressure distribution. It is worth mentioning that the presented measurement technique itself has a potential in the diagnosis of the diabetic foot. Diabetic neuropathy is indicated by a loss of foot flexibility which could be detected by means of stiffness measurements.

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THE INFLUENCE OF PARAMETER VARIABILITY ON SHOE TESTS

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Purpose of the study

The recent study wants to investigate the influence of parameter variability as reported earlier by different authors (see NIGG 1986 for details and references) on tests of different shoe models. Four kinetic and two kinematic parameters generally regarded as loading factors in running have been investigated concerning their abilities to distinguish between the shoe models.

Methods and procedures

Eight injury free subjects with sufficient running experience performed twenty runs with each of eleven pairs of running shoes over a *Kistler* force platform mounted level to the surface of an indoor track at the German Sports University Cologne. Running speed ranged from 3,6m/s to 3,8m/s. A *NAC HSV-400* Highspeed video camera provided a rear view of the runners' left lower limb from the heel to the knees during foot platform contact. The camera's position was adjusted to the average foot eversion angle at midstance of each runner. Kinetic data were gathered at 1000Hz; the frame rate of the camera was adjusted to 200fps. According to *Nigg/Denoth (1980, p.44)* the resulting force time history was assumed to consist of an initial passive part and an active part due to muscular control of the accompanying movements. The initial peak force value as well as the initial force rate were assumed to characterize the passive part. The total impulse together with the second maximum force value were taken as descriptives of the active part. Range of rearfoot motion between first heel-ground contact and takeoff of the heel as well as the maximum rearfoot angular velocity after touchdown of the heel were taken as kinematic parameters of load. After visual inspection of each footfall 1715 and 1708 trials respectively could be included into further analysis.

Results

The values for the parameter patterns are within the ranges which were reported earlier by different authors (see NIGG (1986) for details and references).

It is but evident that the variabilities (see table 1) for the parameters related to the active part of the force-time-history are considerably lower than those related to the initial moment of footstrike.

As far as the total impulse and the "active" maximum force are concerned their comparably low variabilities of 7,9% and 13,0% respectively point to a well controlled running speed within the total sample. The higher standard variations for the kinetic parameters related to the initial footstrike and the kinematic parameters are due to interindividually different adjustments to the shoe models.

Figure 1 reveals the individual differences for ranges of pronation related to four different shoes of one brand. We can observe inter- and intraindividually different variabilities (displayed as quartildeviations) in dependance of the shoemodels. But the effects of the shoes on range of pronation cannot be regarded as uniform. Shoe one (ASICS Alliance) e.g. shows lowest variability for subject D.A. whereas for subject A.K. it is shoe four (ASICS Gel Exult).

Analyses of Variance revealed that the shoe models could not be told apart by the parameters derived from the active part of the force-time-history. As these parameters must be related to the subjects' propulsion from the ground the latter results can be regarded as a logical consequence of a well controlled running speed. Kinetic parameters related to the initial foot ground contact showed differences between shoe models for more subjects than the "active" kinetic parameters.

But as table 2 reveals there are also parameter patterns for shoe combinations which show no significant differences at all for any of the subjects. The best discriminating abilities can be attributed to the kinematic loading factors. For up to seven subjects significant differences could be found concerning range of pronation (shoe 3 : shoe 5 / shoe 2 : shoe 5) and for up to six subjects concerning pronation velocities. (shoe 3 : schoe 6 / shoe 4 : shoe 7 / shoe 6 : shoe 7)

As shown before differences between shoe models are extremely influenced by intraindividual variabilities and cannot be uniformed. Thus the discriminating abilities of kinematic parameters of load can only be validated for intraindividual conditions. Parameters related to ground reaction forces must be rejected as shoe test parameters.

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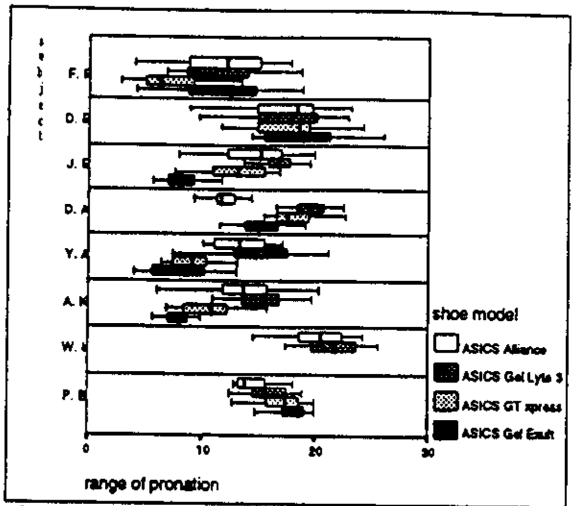

Figure 1: Median, quartildeviation and confidence interval for ranges of pronation for shoes of one brand displayed for every single subject.

Table 1: Mean and standard variation (standard deviation given as % of mean) for kinematic and kinetic loading factors for all trials.

parameter	mean	standard variation
range of pronation (n=1715)	14,4 °	35 %
pronation velocity (n=1715)	163,3 °/s	31,7 %
initial peak force (n=1708)	1119 N	32,5 %
initial force rate (n=1708)	12900 N/s	21,6 %
"active" maximum force (n=1708)	1755 N	13,0 %
total impulse (n=1708)	231 Ns	7,9 %

Table 2: Summary of ANOVA results. The numbers in the table indicate the number of subjects with significant differences between the respective shoes (p<0.1%) related to the parameters: range of pronation / pronation velocity / initial peak force / initial force rate / 2nd maximum force / total impulse. [order of numbers in cells = order of the parameters]

	sh 1	sh 2	sh 3	sh 4	sh 5	sh 6	sh 7	sh 8	sh 9	s10
sh 2	5/3/ 2/1/ 1/1	X								
sh 3	4/3/ 0/0/ 0/2	0/3/ 1/1/ 2/0	X							
sh 4	1/2/ 4/1/ 0/2	5/4/ 1/2/ 0/1	5/3/ 1/2/ 1/1	X						
sh 5	4/3/ 3/4/ 1/3	7/4/ 3/2/ 1/1	7/4/ 1/1/ 0/1	3/4/ 1/1/ 1/1	X					
sh 6	4/3/ 4/2/ 1/2	1/2/ 2/3/ 0/2	1/6/ 2/3/ 1/2	6/5/ 0/0/ 1/1	6/4/ 0/2/ 2/1	X				
sh 7	3/3/ 3/3/ 3/3	5/4/ 4/3/ 2/1	3/1/ 3/4/ 3/1	3/6/ 3/0/ 1/1	3/3/ 2/3/ 1/1	1/6/ 3/1/ 2/2	X			
sh 8	4/3/ 4/4/ 1/3	2/5/ 4/3/ 3/0	3/4/ 3/3/ 1/0	5/3/ 3/3/ 2/0	4/1/ 0/2/ 2/0	1/4/ 3/4/ 1/0	1/5/ 1/3/ 4/1	X		
sh 9	5/5/ 6/3/ 0/1	3/5/ 6/2/ 1/0	3/4/ 3/1/ 0/0	3/4/ 2/1/ 1/0	2/0/ 3/1/ 0/0	3/5/ 2/1/ 1/2	1/5/ 2/2/ 2/1	2/2/ 2/2/ 1/0	X	
sh 10	4/2/ 2/0/ 0/1	5/4/ 2/1/ 1/1	4/3/ 2/1/ 1/1	4/2/ 0/0/ 1/0	2/1/ 1/2/ 0/1	1/2/ 0/0/ 1/0	1/4/ 3/2/ 1/1	2/1/ 2/3/ 1/1	1/3/ 2/1/ 0/1	X
sh 11	5/2/ 3/2/ 1/2	2/4/ 2/2/ 1/1	2/3/ 1/1/ 0/1	5/4/ 0/1/ 1/0	5/0/ 1/0/ 0/1	2/3/ 1/2/ 0/2	2/3/ 2/2/ 2/1	1/1/ 1/1/ 0/0	2/3/ 1/0/ 0/1	4/2/ 1/1/ 0/0

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