Internal Loading of the Foot and Ankle During Impact in Running

Gerald K. Cole, Benno M. Nigg, Gordon H. Fick, and Michael M. Morlock

A 3-D model was used in this study to determine the influence of midsole hardness, as well as the influence of running in shoes in comparison to barefoot, on the contact forces in the joints of the foot and ankle during running. The results showed that there were no statistical differences in the magnitude and rate of joint loading for changing midsole hardness, nor were there any general trends observed in the measured variables. However, both the magnitude and rate of loading in the subtalar and ankle joints during the impact phase were found to be greater in the barefoot condition than the shod condition. The results suggest that if running injuries are assumed to be related to the impact of heel-strike, running in shoes may aid in preventing injuries, whereas it is still questionable whether changes in the midsole hardness have a general influence on the incidence of impact-related injuries.

The high incidence of chronic injuries in running has been assumed to result from the highly repetitive and impulsive nature of the external loading conditions, since this type of loading has previously been associated with injury and degeneration of tissue structures of the body (Robbins & Gouw, 1990). The association between impulsive loading and tissue degeneration is generally based on the results of studies using animal models. For example, mechanically induced, repetitive impulse loading was found to produce subchondral bone stiffening and gross cartilage changes in the knee joints of adult rabbits that were consistent with changes caused by degenerative joint disease (Radin et al., 1973). Radin, Orr, Kelman, Paul, and Rose (1982) later observed decreases in the hexosamine content of cartilage in a group of sheep restricted to walking on a concrete surface in comparison to a control group allowed to walk on a wood chip surface. However, gross pathological changes of severe osteoarthrosis were not evident in the experimental sheep group.

External impulsive loading in heel-toe running occurs during the first 10 to 35 ms following heel strike (Nigg, 1986), leading to the assumption that the...
important time frame of running stance in relation to selected injuries is the so-called impact phase. In support of this hypothesis is the observation that the loading rate of the vertical ground reaction force at heel-strike during walking was 37% higher in subjects with activity-related knee pain than in a group of control subjects (Radin, Yang, Riegger, Kish, & O’Connor, 1991). Additionally, a correlation was reported between a reduced capacity of the musculoskeletal system to attenuate repetitive shock waves resulting from heel-strike in gait and the incidence of low back pain (Voloshin & Wosk, 1982). Substantive proof for the hypothesis is lacking, however, since the external loading conditions do not necessarily reflect the loading of the internal structures of interest. Using inverse dynamics techniques, Scott and Winter (1990) found peak forces in an assortment of chronic running injury sites to be associated with midstance, rather than impact. Furthermore, epidemiological evidence supporting long-term degenerative damage to bone and cartilage from running does not exist (Konradsen, Hansen, & Søndergaard, 1990).

One of the primary concepts that has been incorporated into sport shoe design as a result of the assumption that the impact phase in running can lead to cumulative damage of the musculoskeletal system has been “cushioning” of the impact between the foot and the ground. The constructional feature that is responsible for cushioning is the midsole of the sport shoe (Cavanagh et al., 1985). An assortment of methods have been used to evaluate the cushioning properties of running shoes including measurement of skeletal accelerations with skin-mounted accelerometers (Nigg, Neukomm, & Unold, 1974; Unold, 1973), impact testing of the midsole material of the shoe (Cavanagh, 1980), and measurement of the vertical ground reaction force (Clarke, Frederick, & Cooper, 1983; Nigg, Luethi, Denoth, & Stacoff, 1983). However, a different assessment of cushioning can be obtained depending on the evaluation method used. Additionally, each of these methods provides results that are not well understood, and their relation to the development of injuries is not obvious.

One might suggest that a more appropriate criterion for the evaluation of “cushioning” in running shoes would be the actual forces acting on the internal structures of the body. Direct measurements of tissue forces are not easily performed; however, mathematical models can be used to estimate internal forces. Although the accuracies of such models are difficult to determine, comparisons of forces between different shoe conditions are still valid when the errors in force estimation can be assumed to be systematic between conditions (Nigg & Bobbert, 1990). A mathematical model has been used to determine the timing of the peak internal loading during running (Scott & Winter, 1990); however, the model used was two-dimensional, and no consideration was given to the peak rate of loading in the structures of interest. A three-dimensional model has been used to estimate internal forces of the foot and ankle during running (Burdett, 1982); in this study, the number of structures for which forces were calculated was limited, and the model was not used to determine the influence of different shoe sole properties on the internal loading of the lower extremity.

The purpose of the present study was to determine the influence of midsole hardness of running shoes, as well as the influence of running in shoes in comparison to barefoot running, on the bone-to-bone contact forces and rate of loading in the joints of the foot and the ankle joint complex during the impact phase of running using a three-dimensional model of the foot and the ankle joint
complex. It is suggested that the ankle joint complex is an appropriate location for the study of the internal impact forces in running since little of the external forces would be attenuated at this location, whereas these forces could be attenuated as they move proximally up the lower extremity.

**Methods**

The model of the foot and ankle used for this study was the three-dimensional, six rigid body segment, 12 degree of freedom (12DOF) version developed by Morlock (1989). The model is an extended version of the model introduced by Mann (1986), and it is based on the weight transfer principle proposed by Hamilton and Ziemer (1983). The segments and "joints" of the model are defined in Table 1 and schematically represented in Figure 1. Each joint of the model was given 3 degrees of rotational freedom, except for the ankle joint (Joint 6) and the two metatarsophalangeal joints (Joints 1 and 2), which were defined as hinge joints with 1 degree of freedom. Newton’s equations of motion were formulated for this system and used to calculate the resultant force and resultant moment at each joint. It was assumed that inertial forces of each segment could be neglected (Morlock & Nigg, 1988).

Ten of the 27 foot muscles and the plantar aponeurosis were included in the model (Table 2). The plantaris muscle was neglected as were the four muscle layers of the plantar surface of the foot. The gastrocnemius and soleus muscles were grouped together as the triceps surae. It was assumed that forces in the

<table>
<thead>
<tr>
<th>Structure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Segment 1</td>
<td>Phalangeals of the three medial rays</td>
</tr>
<tr>
<td>Segment 2</td>
<td>Phalangeals of the two lateral rays</td>
</tr>
<tr>
<td>Segment 3</td>
<td>Metatarsal bone of the three medial rays, three cuneiform bones, and navicular bone</td>
</tr>
<tr>
<td>Segment 4</td>
<td>Metatarsal bones of the two lateral rays and cuboid bone</td>
</tr>
<tr>
<td>Segment 5</td>
<td>Calcaneus</td>
</tr>
<tr>
<td>Segment 6</td>
<td>Talus</td>
</tr>
<tr>
<td>Joint 1</td>
<td>Metatarso-phalangeal joint of the three medial rays (connecting segments 1 and 3)</td>
</tr>
<tr>
<td>Joint 2</td>
<td>Metatarso-phalangeal joint of the two lateral rays (connecting segments 2 and 4)</td>
</tr>
<tr>
<td>Joint 3</td>
<td>Midtarsal joint between navicular and talus (connecting segments 3 and 6)</td>
</tr>
<tr>
<td>Joint 4</td>
<td>Midtarsal joint between cuboid and calcaneus (connecting segments 5 and 6)</td>
</tr>
<tr>
<td>Joint 5</td>
<td>Subtalar joint (connecting segments 5 and 6)</td>
</tr>
<tr>
<td>Joint 6</td>
<td>Ankle joint (connecting segment 6 to the rest of the body)</td>
</tr>
</tbody>
</table>
passive structures of the foot and ankle (i.e., ligaments, skin, etc.) could be neglected. Additionally, it was assumed that the bone-to-bone contact force acted at a single point in each joint, referred to as the joint center.

Equations were formulated for the distribution of the resultant force (Equation 1) and resultant moment (Equation 2) at each joint to the bone-to-bone force at the joint and the individual muscle forces crossing the joint.

\[
F_{ji} = \sum_{m=1}^{N_j} (F_{Mm} + F_{Bj} + \delta_j F_L) \quad (1)
\]

\[
M_{ji} = \sum_{m=1}^{N_j} [(r_{jm} \times F_{Mm}) + \delta_j (F_L \times r_{JL})] \quad (2)
\]

for: \( j = 1, 2, 3, 4, 5, 6 \)
\( N_1 = 4, N_2 = 2, N_3 = 7, N_4 = 5, N_5 = 7, N_6 = 10 \)

\( \delta_j = \begin{cases} 1 & \text{if } j = 1,2,3,4 \\ 0 & \text{if } j = 5,6 \end{cases} \)

where \( i = \text{segment number}, j = \text{joint number}, m = \text{muscle number}, N_j = \text{number of muscles crossing joint } j, F_{Mm} = \text{force in muscle } m, F_L = \text{force in the plantar aponeurosis, } F_{Bj} = \text{bone-to-bone contact force in the joint center } J_j \text{ on the side of segment } i, r_{jm} = \text{moment arm of muscle } m \text{ with respect to the joint center } J_j \text{ of joint } j, \text{ and } r_{JL} = \text{moment arm of the plantar aponeurosis with respect to the joint center } J_j \text{ of joint } j. \)

Taking into account the scalar components of all forces and moments, the general moment Equation 1 yielded a system of 12 equations to solve for 10 unknown muscle forces. This system of equations was solved by minimizing the sum of the absolute differences between the resultant joint moments and the
<table>
<thead>
<tr>
<th>Muscle</th>
<th>Ankle joint (Joint 6)</th>
<th>Subtalar joint (Joint 5)</th>
<th>Lateral midtarsal joint (Joint 4)</th>
<th>Medial midtarsal joint (Joint 3)</th>
<th>Lateral metatarsophalangeal joint (Joint 2)</th>
<th>Medial metatarsophalangeal joint (Joint 1)</th>
<th>Origin</th>
<th>Insertion segment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Extensor hallucis longus</td>
<td>X</td>
<td></td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>Fibula, interosseous membrane</td>
<td>1</td>
</tr>
<tr>
<td>Flexor hallucis longus</td>
<td>X</td>
<td>X</td>
<td></td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>Fibula, interosseous membrane</td>
<td>1</td>
</tr>
<tr>
<td>Extensor digitorum longus</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>Fibula, tibia, interosseous membrane</td>
<td>1+2</td>
</tr>
<tr>
<td>Flexor digitorum longus</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>Tibia</td>
<td>1+2</td>
</tr>
<tr>
<td>Tibialis posterior</td>
<td>X</td>
<td></td>
<td></td>
<td>X</td>
<td></td>
<td></td>
<td>Fibula, tibia, interosseous membrane</td>
<td>3</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>X</td>
<td></td>
<td></td>
<td>X</td>
<td></td>
<td></td>
<td>Tibia</td>
<td>3</td>
</tr>
<tr>
<td>Peroneus tertius</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td></td>
<td></td>
<td>Fibula, interosseous membrane</td>
<td>4</td>
</tr>
<tr>
<td>Peroneus longus</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td></td>
<td></td>
<td>Fibula</td>
<td>3</td>
</tr>
<tr>
<td>Peroneus brevis</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td></td>
<td></td>
<td>Fibula</td>
<td>4</td>
</tr>
<tr>
<td>Triceps surae</td>
<td>X</td>
<td></td>
<td></td>
<td>X</td>
<td></td>
<td></td>
<td>Femur, fibula, tibia</td>
<td>5</td>
</tr>
</tbody>
</table>

*Note. X = muscle that crosses the corresponding joint.*
moments about each joint produced by the force-carrying structures. The force in the plantar aponeurosis was determined using Hicks’s equation (Hicks, 1954). The bone-to-bone contact force at each joint was then calculated using Equation 2.

Anatomical, kinematic, and kinetic information was required for each subject to solve for the forces of interest. The anatomical information was derived from a previous cadaver study (Morlock, 1989). A single cadaver leg was positioned in a fixture (Engsberg, 1987), which allowed for systematic changes of the angles of dorsiflexion/plantar flexion, inversion/eversion, and adduction/abduction at the ankle joint complex. The positions and orientations of the six segments of the model were determined in the cadaver specimen for 175 orientations of the ankle joint complex. Following this, the cadaver was dissected into the six foot segments and the lower leg. The locations of the structures of interest (i.e., muscle insertions and origins, bony prominences defining lines of action of muscles, retinacula, specified points of joints) were determined relative to segment fixed coordinate systems so that these locations could be derived for each of the 175 orientations.

The locations of the anatomical structures of interest were related to the kinematics of each segment in the test subject throughout each running trial. The kinematics for each segment were defined using seven retroreflective, spherical markers placed on the running shoe and three similar markers placed on the lower leg as follows (Figure 2): markers 1, 2, and 3 (segments 1 and 2); markers 3, 4, and 5 (segments 3 and 4); markers 5, 6, and 7 (segments 5 and 6); and markers 8, 9, and 10 (the lower leg). An X-ray of the cadaver foot was used to determine the position and orientation of each segment fixed coordinate system relative to the external markers that defined the kinematics of the segment. The orientation of the foot relative to the leg during the running trial could then be matched to the closest cadaver leg configuration expressed in terms of an Euler angle triple at the ankle joint complex. The assumption was then made that the anatomical information derived from the in vitro study could be used for the test subject for each matched position.

The 3-D spatial positions of the 10 external markers were collected at a

![Figure 2](image-url) — Positioning of the markers on the subject’s foot and leg: 1 = slightly medial to the most anterior aspect of the shoe sole; 2 = midway between markers 1 and 3 on the shoe sole; 3 = most lateral aspect of shoe sole; 4 = above sinus tarsi; 5 = sole of shoe at the anterior border of the calcaneus; 6 = posterior aspect of shoe on shoe sole; 7 = 5 cm superior to marker 6; 8 = tibial crest; 9 = head of fibula; and 10 = lateral malleolus.
frequency of 100 Hz for each running trial using a VP310 video recorder (Motion Analysis Corporation, Santa Rosa, CA) and four electronically shuttered cameras (NAC MOS V-14 Camera 60/220 F/S) equipped with 12- to 120-mm zoom lenses (Angenieux, Paris, France). The cameras were placed in an umbrella configuration. The cameras were zoomed in to a volume of 0.6 m (mediolateral axis), 1.0 m (anteroposterior axis), and 0.65 m (vertical axis). This volume was calibrated using 12 control points and Expert Vision 3-D software (Motion Analysis Corporation, Santa Rosa, CA). The error of the system in calculating the spatial coordinates of the markers has been determined to be less than 1 mm (Koh, Grabiner, & Swart, 1992).

The kinetics for each segment were measured using the thin, flexible EMED pressure distribution insole (novel gmbh, Munich, Germany). Double-sided tape was used to adhere the insole to the top of the existing insole inside the test shoe. To ensure the same alignment of the subject’s foot over the insole for each shoe condition, the EMED insole and the shoe insole were removed together and placed inside the new shoe. For the barefoot condition, the procedure as described above was also used. In addition, double-sided tape was placed on top of the EMED insole. The subject’s foot was placed carefully into the shoe, the EMED insole adhering to the plantar surface of the foot via the tape. This procedure was used to ensure that the same alignment of the insole on the foot was obtained in both shod and barefoot conditions. The foot, EMED insole, and shoe insole were removed simultaneously from the shoe, and then the shoe insole was carefully removed from the EMED insole. The EMED insole measured pressures (forces) normal to the insole and not shear forces. An X-ray of the cadaver foot was aligned with the insole and used to determine the distribution and location of the forces acting on the five plantar segments of the model. Pressure distribution data were sampled during each running trial at the maximum sampling frequency of the insole of 100 Hz.

Ground reaction forces were measured using a force platform (Kistler, type 9287). The force plate was connected to an amplifier unit (Kistler, type 9861A), and ground reaction force data were sampled at 1000 Hz. The EMED insole did not fit exactly to the shape of the running shoe insole and, therefore, did not measure all of the forces acting on the plantar surface of the foot. A comparison of the vertical ground reaction force to the total normal force measured by the insole was used as an estimate of the quality of the insole measurement. This comparison was done by expressing the difference between EMED and platform forces as a percentage of the platform measurement for each sampling of the EMED insole and then averaging this value over the entire stance phase of each running trial.

Data collection was triggered manually by an external switch to the EMED storage unit. This caused a signal to be sent from the EMED unit to trigger data collection of the force plate and motion analysis system. Synchronization error between data collection devices was no greater than 1 ms.

Seven physically active male subjects were chosen for the study based on foot size. The mass of each subject was as follows:

<table>
<thead>
<tr>
<th>Subject 1, 75 kg</th>
<th>Subject 5, 91 kg</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject 2, 62 kg</td>
<td>Subject 6, 76 kg</td>
</tr>
<tr>
<td>Subject 3, 83.5 kg</td>
<td>Subject 7, 75 kg</td>
</tr>
<tr>
<td>Subject 4, 82 kg</td>
<td>( M = 77.8, \ SD = 9.1 )</td>
</tr>
</tbody>
</table>
A measurement was taken from the ptemion on the calcaneus to the end of the first toe. This measurement had to match the same length for the cadaver specimen within ±0.4 mm. Based on data collected by Hawes and Sovak (in press), this foot length criterion would ensure that each subject’s foot would match the 23 anthropometric measurements of the cadaver foot within 1 cm, the criterion used in the original application of the model (Morlock, 1989).

Three pairs of shoes (size 9 1/2) were used for the study. The shoes were identical in construction, differing only in midsole hardness (shore C 35, 54, and 75).

The test protocol required the subjects to perform heel–toe running along a 30-m runway. Each subject performed as many practice runs as required to feel comfortable with each shoe condition. The average running speed was measured by two photocells that were mounted 0.8 m on either side of the video calibrated volume at shoulder height. The speed of each trial was controlled to 4.5 ± 0.2 m/s. Any trials falling outside these limits were rejected. Each subject had to perform three successful trials for each shoe condition for which marker position data, pressure distribution data, and force plate data were collected.

The kinematic and pressure distribution data were reduced to the stance phase in the video calibrated volume for each trial. The force in each muscle, the force in the plantar aponeurosis, and the bone-to-bone contact force in each of the six joints of the model were calculated for each trial and normalized according to stance time.

The following variables were calculated for each normalized running trial for analysis in the study:

- \( F_{jk} \): maximal bone-to-bone force acting at each joint \((j = 1, ..., 6)\) during the first 20% of stance time for each shoe condition (shore C hardness 35, 54, and 75, and barefoot)
- \( G_{jk} \): maximal loading rate of the bone-to-bone force at each joint \((j = 1, ..., 6)\) during the first 20% of stance time for each shoe condition (shore C hardness 35, 54, and 75, and barefoot), calculated as the average loading rate over the 10-ms time intervals between force calculations

In this study, the impact phase was arbitrarily defined as the first 20% of stance time. The mean and standard deviation over the three trials per shoe condition per subject were calculated for the above variables.

The following statistical comparisons were made for each joint using a \( t \) test, significant differences between conditions being determined if \( p < .05 \): (a) influence of midsole hardness (35, 54, 75) on the maximal impact loading, \( F_{jk} \), and on the maximal impact loading rate, \( G_{jk} \), and (b) influence of shod (shore C 54) versus barefoot running on the maximal impact loading, \( F_{jk} \), and on the maximal impact loading rate, \( G_{jk} \). For each variable and joint combination, a multiple-regression model was formulated that included subjects, trial order, and footwear condition (Armitage & Berry, 1987). For each model, a test of significance was made to compare the conditions described above. Accordingly, the comparison between conditions was performed after removing subject and order differences. No adjustments for multiple comparisons were made. The statistical design of the study was such that small differences between conditions might not be detected. As a result, for variables where no statistical differences were
found between footwear conditions, a post hoc power analysis was performed. The magnitude of differences that could be detected with power $1 - \beta = 0.8$ were calculated and expressed as a percentage of the mean value for the shore C 54 shoe.

**Results**

Changing the midsole hardness of the running shoe had a minimal effect on the magnitude of the bone-to-bone contact force at the subtalar and ankle joints during the impact phase in running (Figure 3). Similar results were observed for the other four “joints” of the model. Any observed differences between shoe conditions were not found to be statistically significant (subtalar joint, $p = .30$; ankle joint, $p = .70$), and there were no general trends in joint loading for changes in midsole hardness. The post hoc power analysis showed that there was an 80% probability of detecting a difference in impact force between the shore C 35 and 75 shoes of 37% at the subtalar joint and 43% at the ankle joint. There was a considerable amount of variability in results between trials (Table 3), however, as observed by the wide ranges obtained for each shoe condition and by the number of outlier trials. An examination of force in the ankle joint for each subject showed that three of the outlier trials for the shore C 54 shoe were responsible for a substantial increase in joint contact force in this shoe in compari-

![Figure 3](image.png)

**Figure 3** — Box plot of the maximal force during the impact phase in (a) the subtalar joint and (b) the ankle joint for shoes with midsole hardnesses shore C 35, 54, and 75. Adjustments for subject differences were made by calculating the mean force over all trials for each subject (zero line) and plotting the difference between each trial value and the mean. Middle line of each box is the median, outer lines are the quartile range, bars are the absolute range, and dots are the outlier trials for all trials and all subjects.
Table 3  Inter- and Intrasubject Variability

<table>
<thead>
<tr>
<th></th>
<th>Ankle joint</th>
<th></th>
<th>Subtalar joint</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Force (N)</td>
<td>Loading rate (kN/s)</td>
<td>Force (N)</td>
<td>Loading rate (kN/s)</td>
</tr>
<tr>
<td>Intersubject</td>
<td>2,772</td>
<td>110</td>
<td>1,170</td>
<td>47</td>
</tr>
<tr>
<td>Intrasubject</td>
<td>818</td>
<td>56</td>
<td>355</td>
<td>26</td>
</tr>
<tr>
<td>Mean</td>
<td>2,882</td>
<td>173</td>
<td>1,523</td>
<td>89</td>
</tr>
</tbody>
</table>

Note. Variability is expressed as the standard deviation in the measurements of joint contact force and loading rate for the ankle and subtalar joints. Mean values are also presented as a reference. The values presented are for the three shoe conditions with midsole hardnesses of shore C 35, 54, and 75.

Figure 4 — Mean maximal force in the ankle joint during the impact phase in running for 7 subjects and three shoe conditions: midsole hardness shore C 35, 54, and 75.

Figure 4 — Mean maximal force in the ankle joint during the impact phase in running for 7 subjects and three shoe conditions: midsole hardness shore C 35, 54, and 75.

In addition, Subject 1 showed a clear trend toward a higher impact force in the ankle joint in the shore 75 shoe, while Subject 5 showed a trend toward higher force in both the shore 35 and 75 shoes.

There were no significant differences found in the rate of loading in the
subtalar joint \((p = .07)\) and ankle joint \((p = .45)\) during impact in running for changes in midsole hardness (Figure 5). Similar results were found for the other joints of the model. The post hoc power analysis showed that there was an 80% probability of detecting a difference in impact loading rate between the shore 35 and 75 shoes of 48% at the subtalar joint and 46% at the ankle joint. There was substantial variability in the rate of joint loading during impact between trials (Table 3), particularly in the subtalar joint with the shore 75 shoe. On an individual basis, Subjects 1, 5, and 6 showed a clear trend toward increased rate of subtalar joint loading in the shore 75 shoe compared to the other two shoe conditions, while Subject 7 showed a substantially greater rate of loading in the shore 54 shoe (Figure 6).

The differences in the magnitudes of the contact force during the impact phase were not significant in the subtalar joint \((p = .07)\), but were significant in the ankle joint \((p = .03)\) between the shod and barefoot conditions (Figure 7). In general, there was a trend toward greater contact force when running barefoot. The rates of loading in the ankle and subtalar joints during the impact phase were significantly greater when running barefoot in comparison to running in shoes \((p < .001\) in both cases) (Figure 8).

**Discussion**

The purpose of this study was to determine if the midsole hardness of a running shoe had an influence on the internal loading of the joints of the foot and ankle.
during impact in running and, furthermore, to compare the internal impact loading for shod and barefoot running. A model adapted from Morlock (1989) was used to estimate the bone-to-bone contact forces in these joints. Errors in the results of the model fell into five main categories: errors in the estimation of anatomical positions, errors in the kinematic data acquisition, errors in relating the cadaver data to the movement data, errors in kinetic data acquisition, and errors in the solution of the distribution problem. These are all discussed in detail in Section 8.3 of Morlock (1989). Some of these errors are discussed in further detail next.

Morlock (1989) discussed the error in kinematic measurements due to relative movement between motion analysis markers. The possible errors due to movement of the foot within the shoe or due to movement of the skin-mounted markers relative to the underlying bony structures were not discussed. The magnitude of the error in determining the actual bone position could not be quantified. Because the shoes were of identical construction except for the midsole hardness, it was assumed that these errors were systematic between shoe conditions for each subject. However, the errors due to relative marker movement are probably different for skin-mounted and shoe-mounted markers so that the errors are no longer systematic for comparisons of shod and barefoot running. The magnitudes of these errors are unknown and cannot presently be quantified.

The anatomy of the cadaver and the anatomy of each subject were different, and the differences were not quantified. These differences were assumed to be systematic between shoe conditions for each subject. Additionally, external
Figure 7 — Box plot of the maximal force during the impact phase in (a) the subtalar joint and (b) the ankle joint for barefoot running and shod running (shore C 54). Adjustments for subject differences were made by calculating the mean force over all trials for each subject (zero line) and plotting the difference between each trial value and the mean. Middle line of each box is the median, outer lines are the quartile range, bars are the absolute range, and dots are the outlier trials for all trials and all subjects.

marker coordinates were represented in a coordinate system based on the three markers placed on the lower leg. Although these markers were placed on the same anatomical landmarks for all subjects and the cadaver, the relative positions would be different. The orientation of the leg coordinate system differed between subject leg and cadaver leg, which introduced some error into the matching of in vivo ankle joint complex orientations to in vitro orientations. However, this error was systematic between shoe conditions for each subject.

A possible source of error in the kinetic data acquisition in this study was the limitation in sampling frequency of 100 Hz imposed by the currently available equipment. This sampling frequency probably resulted in underestimation of the peak pressure/force amplitudes since data collection and peak force would not necessarily occur at the same time. With a large number of trials per footwear condition, this error would statistically become systematic between conditions. This error may have been quite random in the present study since only three trials per footwear condition were used, and this would make differences between conditions more difficult to detect.

As an estimate of the accuracy of the EMED system in measuring the force acting on the plantar surface of the foot during running, the total force measured by the insole was compared to the vertical ground reaction force. The EMED force measurement was found to vary between about 60 and 80% of the vertical ground reaction force measurement depending on the subject; this value herein is referred to as the force difference. It was speculated that the force difference
Figure 8 — Box plot of the maximal loading rate during the impact phase in (a) the subtalar joint and (b) the ankle joint for barefoot running and shod running (shore C 54). Adjustments for subject differences were made by calculating the mean maximal loading rate over all trials for each subject (zero line) and plotting the difference between each trial value and the mean. Middle line of each box is the median, outer lines are the quartile range, bars are the absolute range, and dots are the outlier trials for all trials and all subjects.

was due to the inability of the insole to entirely cover the plantar surface of each subject’s foot, particularly in the toe area, as well as the fact that small areas of the insole did not have sensors. The variation in force difference between shoe conditions was within 10% for each subject, suggesting that shoe type did not influence force measurement errors. These errors would then cancel out in the comparisons made in this study and would not influence the results. The force difference varied between barefoot and shod running by less than 7% for 4 of the subjects, but was 15 to 21% greater in the barefoot condition for the other 3 subjects. It is not known if this was a real result, or if there were some differences in measurement error between the shod and barefoot conditions.

Another possible error in the measurement of force acting on the plantar surface of the foot resulted from the fact that the EMED insole only measured forces normal to the insole (i.e., no shear forces). However, it was speculated that the maximal internal forces of interest occurred at the time of the impact peak in the vertical ground reaction force when the anterior–posterior and medial–lateral components of the ground reaction force were small in comparison to the vertical component. Thus, it was assumed that the shear forces acting on the plantar surface of the foot at the times of interest were negligible in comparison to the normal force.

The EMED insole may have provided a certain amount of cushioning during barefoot running trials. The thickness of the insole was 2.5 mm, so any cushioning effects were assumed to be small. Furthermore, cushioning during
the barefoot trials might have been expected to mask differences between shod and barefoot conditions. Since significant differences in joint impact loading were still found between the two conditions, these differences may be even larger in normal barefoot running.

Distribution of the resultant joint forces and moments to the internal force-carrying structures has some inherent error associated with it. Speculations on the accuracy of the distribution method used in the 12DOF version of the model can be made by comparing the time history of the calculated muscle forces to published EMG data for the same muscles (Figure 9). (Figure 9 illustrates the activation patterns of the muscles during running stance. Maximal values of the mean normalized forces close to 1 suggest consistent activation patterns across trials and subjects, whereas maximal values substantially less than 1 suggest considerable intersubject or intertrial variability in activation pattern.)

The predicted activation patterns of the triceps surae and tibialis anterior were very similar to the EMG activation patterns. Unlike the EMG data, the estimated tibialis anterior force showed no initial activation. This was due to the quasi-static solution method of the model and the neglect of the inertial forces of the foot. There was some variation in the estimated activation of the peroneus longus between subjects. The majority of the peroneus longus activation occurred at midstance, a trend that was also seen in the EMG data. Although some variation existed in predicted muscle force activation patterns in comparison to EMG, the reported EMG activation patterns are only for a single subject and are not necessarily the same for all subjects. We are not aware of published EMG data for the remaining muscles included in the model.

The 12DOF model used in this study did not incorporate physiological muscle force limitations in the solution of the distribution problem. However, in a comparison of the 12DOF model to a 6DOF model that included physiological muscle characteristics, Morlock and Nigg (1991) showed that even though the absolute magnitude of the estimated forces varied drastically, the relative differences in the forces between controlled conditions were the same for the two models.

In summary, errors existed in the muscle forces and bone-to-bone forces estimated by the model. The present study was a comparative study, and it was not our goal to estimate absolute muscle and joint forces. Most of the errors in the model were systematic for each subject. It was assumed that systematic errors from the model cancelled out so that the relative intrasubject differences between test conditions were still apparent (Nigg & Bobbert, 1990).

It has been suggested that runners adapt their mechanics in response to changes in footwear, these adaptations taking the form of changing kinematics or muscle activation. The fact that adaptations may have occurred in response to the different footwear conditions used in this study is not seen as a limitation, as it was the purpose of the study to detect changes in internal loading regardless of how those changes were produced. We consider adaptations to be neurologically controlled changes resulting from some type of feedback mechanism and not purely mechanically induced changes. Furthermore, since the impact phase has been described as being passive, adaptations that influence the impact would have to be observed immediately prior to heel-strike. There is the possibility that the extreme conditions used in this study induced adaptations that would not normally be observed for the range of shoes that are presently on the market.
Figure 9 — Mean, normalized force in the triceps surae, tibialis anterior, and peroneus longus during running stance in comparison to normalized EMG data for the same muscles. Mean values are for all 7 subjects using the shore C 54 shoe. Muscle forces for each running trial were normalized to the maximal muscle force for each trial before being averaged. EMG data taken from Scott and Winter (1990).

However, in another study conducted in our laboratory, there were no changes observed in the kinematics of the body immediately prior to heel-strike between the shore C 35 and shore C 75 shoes (Koning, Jacobs, & Nigg, in press). A similar result was found by Nigg, Bahlsen, Luethi, and Stokes (1987) for shoes of differing midsole hardness, although midsole construction was shown to significantly influence the initial rearfoot angle in another study (Nigg & Bahlsen, 1988).

The possibility exists that the normal adaptations to midsole hardness
develop over a longer period of time than the warm-up period used in the present study. If this is the case, our study did not detect the normal influence of changing midsole hardness on joint impact loading. This might explain the conflicting results of the other studies, as the study that did detect a significant kinematic adaptation to changing midsole hardness (Nigg & Bahlsen, 1988) was the only one that reported a substantial warmup period with each shoe type (10 min). However, further research is required to make any conclusions on this topic.

In this study we were looking for general trends between footwear conditions that could be observed across subjects. Our ability to detect any trends was limited by the number of trials per footwear condition, the variability observed between trials, and the number of subjects. The number of trials per footwear condition limited the ability to detect differences between conditions within each subject, and the number of subjects limited the ability to extrapolate the results of the study to the general population. To reiterate the results of the post hoc power analysis, the design of the study was such that there was a very good chance of detecting differences of 40–50% between conditions if they existed, but small differences would probably not have been detected. In the discussion of the results to follow, where it is suggested that no general trends were observed between footwear conditions, this more specifically refers to trends in which the difference in the measured variable between footwear conditions would be at least 40%. This is not to say that differences between footwear conditions of less than 40% are not important; however, it is our opinion that precisely what constitutes a relevant difference in the measured variables is not presently known and is a question that requires much further research.

There was no general trend found in the magnitude of joint loading at impact for shoes of different midsole hardness among the subjects of this study. This suggests that there would be no expected response of the joint loading in the general population to changes in midsole hardness. However, this general result was not consistent for all subjects of this study, suggesting that individual responses to changes in midsole hardness may be possible. In addition, given the sensitivity of the model used in this study to estimate joint loading on impact, it is possible that small differences between the various midsoles were not detected.

The result that there was no systematic change in the impact loading of the joints of the foot and ankle for changes in running shoe midsole hardness is in agreement with other studies that have examined the influence of midsole hardness on the external loading to the body. Most of the previous studies found no significant differences in the impact peak of the vertical ground reaction force during both walking and running (Clarke et al., 1983; Lafortune & Hennig, 1992; Nigg et al., 1987) for changing midsole hardness. One study did surprisingly find an increased impact peak for softer midsoles (Nigg & Bahlsen, 1988), although the midsoles used in this study were made out of different materials altogether, making it difficult to fully interpret the result. In contrast, substantial reductions in peak tibial accelerations on impact have been reported with the use of "cushioned" footwear and viscoelastic shoe inserts (Lafortune & Hennig, 1992; Voloshin and Wosk, 1981).

A theoretical explanation can be given for these apparently conflicting results. Bobbert, Schamhardt, and Nigg (1991) have shown that the inertial contribution of the lower leg to the impact peak in the vertical ground reaction...
force is small (200-N contribution to a 1,600-N peak for one subject), which is
due to the small mass of the lower leg. A change in the acceleration of the tibia
reflects only a small change in the internal forces acting on the tibia since, by
nature of its small mass, the change in the inertia of the tibia would be small.
For example, if the mass of the tibia is assumed to be 1 kg and its peak vertical
acceleration on heel-strike changes from 8 g to 5 g using a cushioned shoe, this
represents a change in the net forces acting on the tibia of only 29.5 N. With
regard to the ground reaction force studies, it is possible that the changes in tibial
acceleration on impact due to "cushioning" were reflected in the vertical ground
reaction force, but that the effect was too small to be detected based on the
sample sizes used and the interstrike variability in the ground reaction forces
(Clarke et al., 1983; Nigg et al., 1987). It is also possible that the vertical ground
reaction force does remain the same, but there is a redistribution in the segmental
contributions to the ground reaction force in response to the different shoe
conditions (Bobbert et al., 1991). Furthermore, Stacoff, Denoth, Kaelin, and
Stuessi (1988) have shown theoretically that there is a cushioning effect provided
by pronation of the foot after heel-strike and that "pronation cushioning" in-
tcreases with increasing midsole hardness as a result of changes in the moment
arm of the ground reaction force about the subtalar joint axis. Their results suggest
that cushioning is a three-dimensional problem and is not so easily influenced
by simply changing the midsole hardness.

The results of the present study suggest that there would be no expected
response to changes in midsole hardness in the rate of joint loading during impact
in heel–toe running in the general population. However, the result that there was
no statistical difference in subtalar joint rate of loading between midsoles (p = .07) is questionable considering the limitations of the study; there may have been
a trend toward increasing rate of subtalar joint loading with increasing midsole
hardness. This trend was certainly evident in the individual responses of 3 of
the 7 subjects who exhibited increased rate of loading in the subtalar joint when
running in the shore 75 shoe in comparison to the shore 35 and 54 shoes. As
discussed, it is also possible that small differences in the rate of joint loading
between the various midsoles were not detected due to the sensitivity of the
model used in this study.

The rates of loading in the joints of the foot and ankle during the impact
phase of running have not been published previously. The results of studies of
the ground reaction forces in running tend to support the idea that there may be
individual responses to midsole hardness in terms of an increasing rate of joint
impact loading with increasing midsole hardness. Clarke and co-workers (1983)
found that although the magnitude of the impact peak in the vertical ground
reaction force did not change, the time to the impact peak decreased with increas-
ing midsole hardness. Their results suggest that the average rate of loading of the
vertical ground reaction force during the impact phase increases with increasing
midsole hardness of the running shoe. The average rate of loading of the vertical
ground reaction force was also shown to change significantly between a variety
of different running shoes (Snel, Dellemann, Heerkens, & Ingen Schenau, 1985),
although the shoes were not limited strictly to differences in midsole hardness.
In their theoretical study, Stacoff and co-workers (1988) also suggested that
although increasing midsole hardness would have little effect on the peak vertical
ground reaction force during impact, it would result in a decreased time to
the peak. In contrast, Nigg and co-workers (1987) did not find any significant differences in the maximal loading rate of the vertical ground reaction force during running for shoes of different midsole hardness.

The finding of this study that there is a trend toward increasing peak magnitude and rate of joint impact loading in the foot and ankle when running barefoot in comparison to shoes is well supported by other research. Theoretically, it has been shown that the midsole materials used in running shoes are capable of absorbing energy when loaded (McCullagh & Graham, 1985). In addition, Jørgensen and Bojsen-Moller (1989) have shown that the shock absorbency of the human heel pad is increased by confining it within a shoe. The shock absorbency effect of the shod condition has been observed in studies of the ground reaction forces in walking where the peak vertical ground reaction force, force loading rate, tibial acceleration, and tibial jerk at heel-strike were significantly greater for the barefoot than the shod condition (Lafortune & Hennig, 1992).

The results for running are consistent, at least in part, with those of walking. The average loading rate of the vertical ground reaction force during impact was measured to be significantly greater for running barefoot than with shoes (Snel et al., 1985), as was the peak axial acceleration of the tibia (Valiant, 1990). All of the above results are in sharp contrast, however, to a proposed theory that the impact associated with heel-strike in running, and its potential effects on the musculoskeletal system, should be decreased in the barefoot condition due to feedback from plantar sensations that are attenuated in the shod condition (Robbins & Gouw, 1991). The theory is supported in part by a study that identified changes in the kinematics of the body immediately prior to heel-strike when running barefoot in comparison to running in the identical shoes used in the present study (Koning et al., in press). The results of the present study, however, show that even if these potential feedback mechanisms induce adaptation, the impact is still more severe when running barefoot than in shoes.

It is interesting to note that although the intent of the present study was to look for general trends between footwear conditions, there were sometimes different responses to these conditions between subjects. Previous studies of adaptations to changing midsole hardness only looked for trends that were generalized across all subjects (Koning et al., in press; Nigg et al., 1987). Support for the possibility of adaptations on an individual level comes from one previous study which showed that runners adapted their angle of knee flexion on surfaces of different hardness and that the strategy of adaptation differed between the runners (Nigg & Denoth, 1980). It is difficult at this time to be confident that the different trends observed in the present study were real results, due to the statistical limitations of the study. Further research may be warranted to determine if there are individual responses to changes in running shoe midsole hardness since the existence of such responses would require a more subject-specific approach to running shoe design.

It has been assumed that impact in running and appropriate cushioning of that impact are associated with injury (Radin et al., 1991; Robbins & Gouw, 1990). The magnitude of change in the internal impact forces that might be required to influence the integrity of biological tissues is not known. Further study in this area is required before the results of the present study can be related to the potential for injury reduction in running. The influence of midsole hardness on internal impact forces appears to be subject dependent, suggesting that the
use of cushioning as an injury prevention mechanism may have to be evaluated on an individual basis rather than applied universally, although the statistical limitations of this study prevent any definitive conclusions on this idea. Epidemiological studies into the effects of cushioning on the reduction of injuries to the lower extremities have given conflicting results. Viscoelastic inserts have been shown to have no effect on the incidence of stress reactions of bone (Gardner et al., 1988, \( N = 3,025 \) military recruits; Milgrom et al., 1990, \( N = 22 \) rabbits). However, in another study, the use of neoprene foot inserts was found to significantly decrease the incidence of tibial stress syndrome in military recruits (\( N = 1,388 \)) (Schwellnus, Jordaan, & Noakes, 1990). The results of these studies are contradictory and inconclusive, as some results may be due to kinematic effects rather than a cushioning effect. In addition, cushioning (concrete vs. wood-chip surface) has been shown to be a relevant parameter in changes to the cartilage of sheep (Radin et al., 1982). Yet, the assumption that this result is representative of runners is questionable since it is not supported by epidemiological evidence (Konradsen et al., 1990; Lane et al., 1986; Panush et al., 1986). Furthermore, the rate of impulsive loading applied to a joint has been suggested to be the critical parameter in terms of stimulating degenerative changes to the cartilage. If it is assumed that runners are at risk of developing osteoarthritis in the joints of the lower extremities, the results of this study suggest that running in shoes may prevent its onset through a decrease in the rate of loading in the joints of the lower extremities, while the influence of midsole hardness of the running shoe remains questionable.

References


Acknowledgments

This work was supported by the Alberta Heritage Foundation for Medical Research (AHFMR), the Natural Sciences and Engineering Research Council (NSERC), and Adidas Research Center Lucerne.