EFFECTS OF ARCH HEIGHT OF THE FOOT ON ANGULAR MOTION OF THE LOWER EXTREMITIES IN RUNNING

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Abstract—It has been suggested that a relationship exists between the height of the medial longitudinal arch of the foot and athletic injuries to the lower extremities. However, the functional significance of arch height in relation to injury is not well understood. The purpose of this study was to determine the influence of arch height on kinematic variables of the lower extremities that have been associated with the incidence of injury in running in an attempt to gain some insight into a functional relationship between arch height and injury. The three-dimensional kinematics of the lower extremities were measured during running for 30 subjects using high-speed video cameras. A joint coordinate system was used to calculate the three-dimensional orientation of the ankle joint complex for a single stance phase. Simple, linear regression analyses showed that arch height does not influence either maximal eversion movement or maximal internal leg rotation during running stance. However, assuming that knee pain in running can result from the transfer of foot eversion to internal rotation of the tibia, a functional relationship between arch height and injury may exist in that the transfer of foot eversion to internal leg rotation was found to increase significantly with increasing arch height. A substantial (27%), yet incomplete, amount of the variation in the transfer of movement between subjects was explained by arch height, indicating that there must be factors other than arch height that influence the kinematic coupling at the ankle joint complex. Additionally, the transfer of movement is only one factor of many associated with the etiology of knee pain in running. Therefore, it is suggested that a running-injury-related foot typology based on arch height is not possible at this time.

INTRODUCTION

The height of the medial longitudinal arch of the foot has been suggested as an important structural feature of the foot that may be related to athletic injury, and, being a non-invasive measurement, arch height could have clinical relevance as an indicator of a person’s predisposition to injury. Yet the functional significance of the height of the arch in relation to injury is not well understood. In general, it has been suggested that a greater percentage of athletic injuries occur with flat feet than normal and high arch feet (Subotnick, 1981). However, an increase of activity-related injuries with increasing arch height has been reported more recently in army recruits (Cowan et al., 1989). It may be that the mechanics and types of injuries, and, hence, the influence of arch height on the injuries, differ between activities. Therefore, studies directed toward determining the influence of arch height on injury related variables may provide greater insight into a functional relationship between arch height and injury for a specific activity.

Excessive motion in the ankle joint complex has been associated with injuries in many sports, especially in running (Clement et al., 1981; James et al., 1978; Nigg et al., 1986). In these studies, pronation has been the variable most commonly linked to injury. For example, Clement et al. (1984) found that 56% of runners with Achilles tendon pain (n = 106) had excessive pronation. Similarly, Bahlsen (1988) found that 52% of runners with an assortment of injuries (n = 136) had excessive pronation during running. The most common location of injuries during running is at the knee (Bahlsen, 1988; Clement et al., 1981; James et al., 1978). One hypothesis that has been put forward for the etiology of selected knee injuries is that pronation of the foot is transferred into internal rotation of the tibia, which in turn results in abnormal mechanics of the knee joint (Bahlsen, 1988; James et al., 1978; NIH, 1989). The coupling of pronation—supination movements of the foot to axial rotations of the tibia has been described theoretically (Inman, 1976), as well as in in vitro and in vivo studies of the kinematics of the ankle joint complex (van Langelaan, 1983; Lundberg et al., 1989; Siegler et al., 1988); and it is thought to occur mainly through the subtalar joint (Inman, 1976). Based on the above hypothesis, three kinematic variables can be identified which may be determinants of running injury and which may be functionally related to structural features of the foot such as arch height. These are pronation of the foot, the amount of pronation that is transferred to tibial rotation through the ankle joint complex, and the amount of relative axial rotation at the knee joint.

In attempting to establish a functional relationship between arch height and injury during locomotion, the influence of arch height on pronation of the foot has been examined previously for both walking and running. The results of these studies have been conflicting. Greater range of motion in the subtalar joint...
has been observed in flat than high arch feet (Close and Inman, 1967), and general statements have been made suggesting that high arch feet are inflexible, while flat feet tend to be hypermobile and susceptible to a large degree of pronation (Cavanagh, 1980; Mann et al., 1981; Subotnick, 1985). However, other studies have suggested that there is no difference in rearfoot motion (in-eversion) between low and high arch feet during running (Hamill et al., 1989) and walking (Kernozek and Ricard, 1990). The influences of arch height on axial rotation of the tibia and on the transfer of pronation to axial rotation of the tibia have not been examined. Yet, these influences may be expected since arch height could be an indicator of the structure of the tarsus which acts as the link between the foot and the tibia.

The purpose of this study is to quantify the relationship between the height of the medial longitudinal arch of the foot and (a) the eversion movement of the foot, (b) the axial rotation of the leg and (c) the transfer of eversion to internal leg rotation during heel-toe running. Such an examination may provide some insight into a possible functional relationship between arch height and injury (in particular, knee pain) in running. Clinically, the results of this study may establish arch height as a risk factor in running injuries, which can then be used as a basis for the design of corrective footwear.

METHODS

Fifteen female and 15 male adult subjects were recruited for the study from the University of Calgary campus. Each subject gave informed consent to participate in the study. Prior to inclusion in the study, the foot types of the subjects were examined cursorily in order to include a wide range of arch heights. In this cursory examination, all subjects reported to have no pain or discomfort during running. The arch height measurement technique has been described and illustrated in detail in Hawes and Sovak (1992) and Hawes et al. (1992). In brief, arch height measurement was taken on the left foot in a full weight bearing position with the right foot resting lightly on a raised (25 cm) platform of a specially constructed measurement table. The ankle joint was placed in a neutral position with the body in a normal upright posture. A modified Mitutoyo digital caliper was used to measure the highest point along the soft-tissue margin of the medial plantar curvature. This point was marked and measured. The method has been reported to have a reliability coefficient of 0.99 and an objectivity coefficient of 0.98 (Hawes et al., 1992). The mean values and ranges for age, body height, body mass, and arch height of the subjects are given in Table 1 in comparison to the values from Hawes and Sovak (1992).

Table 1. Means and standard deviations for subject age, body height, body mass, and arch height for the present study in comparison to Hawes and Sovak (1992).

<table>
<thead>
<tr>
<th></th>
<th>Present study</th>
<th>Hawes and Sovak (1992)</th>
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<tbody>
<tr>
<td>Age (yr)</td>
<td>30.4 (+ 7.6)</td>
<td>35.5 (+ 11.9)</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>66.4 (+ 10.7)</td>
<td>77.3 (+ 10.4)</td>
</tr>
<tr>
<td>Arch height (cm)</td>
<td>2.64 (+ 0.49)</td>
<td>2.12 (+ 0.67)</td>
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Subject kinematics were determined using six light reflective, spherical markers with a diameter of 10 or 20 mm attached to the subject's foot and leg with adhesive tape. The spheres were located according to the specifications in Fig 1. The three-dimensional spatial positions of the markers were collected using four electronically shuttered, high-speed video cameras (NAC MOS-TV, V-14B, Japan) equipped with 12.5-75 mm zoom lenses (Cosmicar, Japan) and a VP310 video processor (Motion Analysis Corporation, Santa Rosa, CA). The cameras were located to the left of the subject in an 'umbrella' configuration, with the first camera located anterior, the second camera located lateral and slightly anterior, the third camera located lateral and slightly posterior, and the fourth camera located posterior to the subject. The sampling frequency of the cameras was set at 200 frames s⁻¹ and the exposure time at 1/2500 s. The raw data were stored on a SUN 3/280 computer.
A volume was calibrated (Expert Vision-3D software, Motion Analysis Corporation, Santa Rosa, CA) near the middle of a 30 m indoor runway, where a force plate (Kistler, type 9287) was mounted flush with the floor. The force plate was connected to an amplifier unit (Kistler, type 9861 A) and force data were sampled at 1000 Hz. All subjects ran using a heel-toe movement and wore the same running shoes. The subjects were given as many practice runs as required to ensure that the left foot landed naturally on the force plate and to achieve a running speed of $4.0 \pm 0.4 \text{ m s}^{-1}$. Two successful running trials were recorded for each subject. Mean running speed was determined based on the time it took for the subject to run between two photocells spaced at a measured distance. The photocells were placed on opposite sides of the force plate at shoulder height. Synchronous data collection for the video system and force plate was triggered as the subject ran past the first photocell. Touchdown of the running stride of interest was defined as the time at which the vertical ground reaction force was greater than zero for at least five successive samplings. Takeoff was defined as the time at which the vertical ground reaction force was zero for at least five successive samplings.

Prior to the running trials, each subject was videoed for 4 s while standing in a calibration frame in order to define a neutral position required for the determination of three-dimensional joint orientations (Fig. 2). The subject stood on the platform of the frame with the posterior heel centered against a small vertical plate. The forefoot was adjusted so that the anteroposterior axis of the calibration frame was in alignment with a line between the first and second ray of the subject's foot. Adjustable bars were positioned so that the line connecting the subject's greater trochanter and lateral maleolus was vertical. Straps were secured around the subject's thigh and leg to prevent motion during videoing. Care was taken to ensure that the straps did not change the position of the skin markers relative to each other or to the underlying bony structures. The angles between the body segments of the left lower limb in this position were defined as zero.

A Turkey running weighted average algorithm implemented in the Expert Vision-3D software was used as a low-pass filter on the three-dimensional spatial coordinates of each marker. The weights were defined by $A(k)=[1+\cos(nk/4)]/2$, where $k=-3, \ldots, -1, 0, 1, \ldots, 3$ defines the data points that are averaged to determine the new value for the data point $k=0$. The cutoff frequency of the filter was 25 Hz.

Four types of coordinate systems were defined for this study: laboratory coordinate system (LCS)—based on the calibration of the motion analysis system; calibration frame coordinate system (FCS); marker coordinate systems (MCS)—one for each body segment; and segment coordinate systems (SCS)—one for each body segment. Calculation of the three-dimensional orientation between body segments using Kintrak software consisted of three steps: static calibration of each MCS to the SCS, calculation of the rotation matrix for each SCS relative to the LCS during the dynamic trial, and the use of a joint coordinate system (Grood and Suntay, 1983) to calculate orientation between body segments.

The calibration frame was used to establish the relationship between the MCS and SCS for each body segment. The FCS was defined based on three markers placed on the frame. The MCS for each segment was defined based on the three skin markers. The SCS for each segment was assumed to have the same orientation as the FCS when the subject was properly placed within the frame (i.e., all angles between SCSs were zero in the neutral position). The transformation matrix, $[T]$, from MCS to SCS was calculated for each segment. The three axes of each SCS were defined in the calibration frame as follows: the $X$ (mediolateral) axis was oriented from left to right, the $Y$ (anteroposterior) axis was oriented anteriorly, and the $Z$ (inferior–superior) axis was oriented verti-cally.

For each dynamic trial, the orientation of each body segment MCS relative to the LCS was calculated from the collected video data. The orientation of the SCS for each segment with respect to the LCS was then calculated using the previously determined transformation matrix, $[T]$. The rotation matrices between SCSs of adjacent body segments were then calculated.

A joint coordinate system (JCS) was used to calculate the three-dimensional orientation between body
segments (Grood and Suntay, 1983). Two different JCSs were used for the ankle joint complex. The first, JCS 1, represented orientation of the foot relative to the leg, and it was constructed with the first body fixed axis (e1) as the mediolateral axis of the leg SCS (i_leg) and the second body fixed axis (e2) as the anteroposterior axis of the foot SCS (i_foot). The floating axis was calculated as e2 = e3 x e1 (Grood and Suntay, 1983). JCS 1 was used to calculate the component of inversion–eversion at the ankle joint complex about the axis i_foot as follows:

\[
\beta = \frac{\pi}{2} - \arccos (i_{\text{foot}} \cdot e_2),
\]

where \(e_2 = i_{\text{foot}} \times i_{\text{leg}}\).

An everted position of the foot was defined as positive and an inverted position as negative.

For the purposes of this study, it was also necessary to calculate internal–external rotation of the leg relative to the foot. The rotation about the floating axis of JCS 1 would constitute ad-abduction, and this floating axis only corresponded to the long axis of the leg in the specified neutral position. Therefore, a second JCS was constructed with e1 as the mediolateral axis of the foot SCS (i_foot) and e3 as the inferior–superior axis of the leg SCS (i_leg). Internal–external rotation of the leg was then calculated as a rotation about the long axis of the leg (i_leg) as follows:

\[
\rho = \frac{\pi}{2} - \arccos (i_{\text{leg}} \cdot e_2),
\]

where \(e_2 = i_{\text{leg}} \times i_{\text{foot}}\).

An externally rotated position of the leg was defined as positive and an internally rotated position as negative.

All angles were calculated relative to the neutral position of the subject in the calibration frame. The repeatability of the positioning of a subject in the calibration frame was tested by collecting 10 calibrated positions in the frame and applying them to a single running trial for one subject. The three-dimensional orientation at the ankle joint complex was calculated 10 times for the running trial using each of the calibrations and the methods outlined above. The standard deviation in each of the angular components was calculated with the following results: plantar-dorsiflexion \(\pm 1.5^\circ\), internal–external leg rotation \(\pm 4.8^\circ\), and in-eversion \(\pm 2.4^\circ\).

In order to eliminate systematic errors associated with the determination of the neutral position, variables were defined to represent ranges of motion, rather than single orientations: (a) the maximal eversion movement of the foot with respect to the leg (\(\Delta \beta_{\text{max}}\)) was defined as the difference between the internally rotated position of the leg at the time of maximal eversion and the internally or externally rotated position of the leg 10 ms prior to heel-strike. Based on these definitions, the time frames for \(\Delta \beta_{\text{max}}\) and \(\Delta \rho_{\text{max}}\) were identical. A transfer coefficient (\(T_{\beta}\)) was then defined to describe the transfer of eversion movement of the foot to internal rotation of the leg through the ankle joint complex,

\[
T_{\beta} = \frac{\Delta \rho_{\text{max}}}{\Delta \beta_{\text{max}}}. 
\]

All variables were averaged over the two trials for each subject. Simple, linear regression analysis was used to determine the relationships between angular motion and arch height. The dependent variables chosen for the regression analyses were eversion movement (\(\Delta \beta_{\text{max}}\)), internal leg rotation (\(\Delta \rho_{\text{max}}\)), and transfer coefficient (\(T_{\beta}\)).

**RESULTS**

There was no correlation found between maximal eversion movement (\(\Delta \beta_{\text{max}}\)) and arch height in this study (Fig. 3: \(r^2 = 0.059, p < 0.197\)). This result suggests that a functional relationship between arch height and running injury through the influence of arch height on eversion of the foot does not exist.

A significant correlation was found between maximal internal leg rotation (\(\Delta \rho_{\text{max}}\)) and arch height (Fig. 4: \(r^2 = 0.152, p < 0.033\)), with a trend towards increasing internal leg rotation relative to the foot with increasing arch height. However, the 95% confidence limit for the estimation of the regression line was essentially horizontal, indicating that although the data suggested a slight influence of arch height on internal leg rotation in running, the influence may be negligible.

![Fig. 3. Scattergram of maximal eversion movement (\(\Delta \beta_{\text{max}}\)) versus arch height, with first-order regression curve (solid line) and the 95% confidence intervals for the estimation of the regression line (dashed lines). The graph shows that arch height does not influence the amount of eversion movement of the foot during running.](image-url)
A significant correlation was found between transfer coefficient \( T_{in} \) and arch height (Fig. 5: \( r^2 = 0.267, p < 0.0034 \)). The relationship showed an increasing transfer of foot eversion to internal leg rotation with increasing arch height. The 95% confidence interval for the estimation of the regression line still illustrated the same trend as the regression line, indicating that arch height was a factor that influenced the transfer coefficient. Only 27% of the variance in transfer coefficient was explained by arch height, however, suggesting that other factors were also involved in the transfer of eversion to internal leg rotation. The regression analysis identified one subject as being an outlier (arch = 3.04 cm, \( T_{in} = -0.53 \)). This subject had a substantial influence on the correlation obtained and, if removed, 34% of the variance in transfer coefficient could be explained by arch height. This result suggests that a functional relationship between arch height and running injury through the influence of arch height on the transfer of foot eversion to internal rotation of the tibia may exist.

**DISCUSSION**

The purpose of this study was to determine if arch height has an influence on certain kinematic variables that have been suggested to be related to injuries in running. The quantitative description of three-dimensional orientation between body segments used in this study depends, among other factors, on the definitions of the segment coordinate systems (Pennock and Clark, 1990) and the (arbitrary) definition of the neutral position. As a result, comparisons of absolute variables between subjects or between studies are not always possible. The variables chosen for analysis in this study reflect differences between joint orientations during the stance phase of running and, therefore, are not influenced by systematic errors in absolute values.

Results for three-dimensional joint angular motion may also depend on the placement of the markers and relative movement between the markers and the underlying bone. In this study, foot marker \( F \) was placed at the head of the fifth metatarsal and marker \( E \) was placed on the distal calcaneus, both positions where little relative movement between the foot and shoe would be expected. It was also assumed that the sole of the running shoe was sufficiently rigid to prevent relative movement between these two markers. Marker \( D \) was placed at the dorsum of the foot, and the relative movement between the foot and shoe was not known at this location. It was assumed that errors in representing motion of the dorsum of the foot were systematic and not influenced by arch height. This assumption is supported by the finding that although it has been shown that the rearfoot moves relative to the shoe during running, the error was systematic so that the motion determined from shoe markers was highly correlated with rearfoot motion \( r^2 = 0.817 \) based on data from Stacoff et al., 1992). Similarly, with respect to the determination of leg motion, it was not expected that arch height would influence the relative motion between the skin markers and the underlying bone. Therefore, it is suggested that any visible relationships between arch height and the measured variables were real results.

The use of the transfer coefficient \( T_{in} \) assumes that there is a direct coupling between foot eversion and internal rotation of the tibia during running, an assumption that is supported by theoretical considerations (Inman, 1976), and both in vitro (Siegler et al., 1988) and in vivo studies of ankle joint complex kinematics (Lundberg et al., 1989). The coupling of eversion to internal rotation of the tibia has not been measured in running, however. The results of a simple
Fig. 6. Mean values and standard error (dotted lines) for in-eversion ($\beta$) and leg rotation ($\rho$) normalized over stance time for all subjects ($n = 30$). $\beta > 0^\circ$ is eversion and $\beta < 0^\circ$ is inversion. $\rho > 0^\circ$ is external leg rotation and $\rho < 0^\circ$ is internal leg rotation.

Regression analysis of inversion-eversion versus axial leg rotation (Fig. 6) using the data from this study (Fig. 7) showed a correlation between the mean values of the two variables ($r^2 = 0.991, p < 0.0001$). The use of mean data to illustrate the coupling effect is not ideal. However, the extremely high correlation found using the data from this study supports the assumption that foot eversion is coupled to internal rotation of the tibia during the first half of running stance.

The measured values for maximal internal leg rotation in this study are reasonable in comparison to measurements taken in other studies. Direct measurements of tibial rotations during running are not presently available for comparison with the results of this study. However, mean internal tibial rotations of $19.2^\circ$ ($n = 12$) have been measured during walking using bone pins (Levens et al., 1948). The mean result of $21.8 \pm 8.4^\circ$ for internal leg rotation relative to the foot in this study ($n = 30$) was within the acceptable range of motion for the tibia based on the walking measurements. Axial rotation of the leg relative to the foot was assumed in this study to be an indicator of the relative motion at the knee joint since methodological problems negate the possibility for quantifi-
ation of axial rotation of the tibia relative to the femur. It was not possible to determine the validity of this assumption.

Eversion motion measured in this study (28 ± 7.2°) was about 10° greater than that measured in previous studies (Nigg, 1986). This was due partly to the two-dimensional eversion determination used in the previous study (Areblad et al., 1990), as well as the fact that, with the marker placement used in this study, ‘total foot’ eversion was measured rather than ‘rearfoot’ eversion as in the previous study. The result of the present study that maximal eversion motion was not influenced by arch height is in agreement with other recent studies (Hamill et al., 1989; Kernozek and Ricard, 1990). However, this result is in contrast to the common clinical notion that subjects with high arches have more rigid feet and subjects with flat arches have hypermobile feet in terms of their inversion–eversion range of motion (Cavanagh, 1980; Subotnick, 1981, 1985).

It was not possible to directly compare the results of the transfer coefficient from this study to the results of other studies since quantification of the coupling of foot eversion to internal leg rotation in running has not been done previously. The mean transfer coefficient found in this study of 0.76 ± 0.16 was substantially higher than a similar coefficient representing the ratio of eversion of a foot platform to internal tibial rotation: 0.20 (Lundberg et al., 1989). However, in the in vivo study of Lundberg and coworkers, the calcaneus was not fixed to the footplate which resulted in relative rotation between calcaneus and footplate of 37° over a range of motion in inversion–eversion of the footplate of 40°. Therefore, the coefficient of 0.20 represented the transfer of forefoot eversion to internal rotation of the tibia, and it is suggested that this coefficient is not representative of ankle joint complex kinematics in running, where considerable rearfoot eversion has been shown to occur (Nigg, 1986).

Approximately 27% of the intersubject variation in the transfer of foot eversion to internal leg rotation was accounted for by arch height, a relationship which can be explained based on the mechanism for the transfer of in-eversion to tibial rotation. Eversion of the foot during running is first transferred from the calcaneus and navicular to the talus. Theoretically, it may be argued that the amount of movement transfer is determined primarily by the inclination of the subtalar joint axis, the amount of talar rotation per degree of in-eversion increasing as the axis becomes more vertical in the sagittal plane (Inman, 1976). The movement of the talus is then transferred to the tibia at the ankle joint. This transfer results from bony contact or by tension in the ligaments surrounding the ankle joint, and it can be influenced by the laxity that is known to exist in the joint (van Langelaan, 1983). Additionally, plantar-dorsiflexion movement of the talus will result in a small amount of axial rotation of the tibia due to the obliqueness of the ankle joint axis (Inman, 1976).

It may be speculated that the height of the medial longitudinal arch of the foot reflects the construction and shape of the tarsus of the foot and may be an indication of the inclination of the subtalar joint axis in a sagittal plane, a high arch foot having a more vertical axis and a low arch foot having a more horizontal axis. A relationship between arch height and movement transfer from eversion to internal leg rotation would then be expected since the subtalar joint is theorized to be one of the main components of the transfer mechanism. A high correlation between arch height and movement transfer would not be expected since the other factors involved in the transfer mechanism are probably not related to arch height. Furthermore, a low arch may be an indication of disfunction in the support mechanism of the arch (Funk et al., 1986; Mann et al., 1985; Subotnick, 1981), rather than a different bony construction in comparison to a normal arch. In such cases, the inclination of the subtalar joint axis may or may not change relative to the normal arch.

The results of this study suggest that a functional relationship between arch height and injuries in running does not exist through the influence of arch height on either foot eversion or internal leg rotation. However, such a relationship may exist through the influence of arch height on the amount of foot eversion that is transferred to internal leg rotation at the ankle joint complex. This speculated relationship is based on the hypothesis that anterior knee pain is caused by excessive tibial rotation at the knee joint, which is a result of excessive pronation and the transfer of movement through the ankle joint complex (Bahlsen, 1988; James et al., 1978; NIKE, 1989).
internal leg rotation with increasing arch height was found in this study, suggesting that a high arch foot may be at risk to knee pain in running.

Arch height had a substantial influence on the transfer of eversion to internal leg rotation; however, this movement transfer is only one factor of many that might influence the axial rotation of the tibia relative to the femur. Furthermore, the movement transfer can only be associated with the etiology of knee pain at this time, although it is possible that it could be related to injuries resulting from torsional loading of the tibia. Therefore, arch height may be one of many risk factors associated with running injuries, but it cannot be used clinically to define a general foot type that is at risk to injury.

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