A significant volume of scientific literature has been published studying the effects of different types of foot orthoses. Although ample evidence supports the concept that foot orthoses are effective at relieving pain and resolving symptoms, the scientific rationale behind these observations is still controversial, and the exact mechanism of action of foot orthoses is still not well understood.

Most studies investigating the mechanical effects of foot orthoses have focused on their kinematic effects during walking and running. Although some research studies on running have shown significant changes in rearfoot motion with foot orthoses, other studies have not, and the amount of angular change in rearfoot motion with foot orthoses from these studies remains small (1°–3°). On the other hand, research studies on walking have shown more consistent results, with significant changes in rearfoot motion with foot orthoses. As with running studies, the angular changes obtained in walking studies remain small (2°–4°), which has led to the speculation that foot orthoses mechanically act to produce their positive therapeutic effects by means other than by altering rearfoot position and motion.

In addition, despite the significant amount of research regarding the effects of foot orthoses on the kinetics of running, little attention has been given to...
the effects of foot orthoses on rearfoot kinetics during walking. For example, although research has shown that foot orthoses significantly decrease net internal inversion moments during running, this effect has not been widely investigated during walking. Because of the need for more research on the mechanical effects of foot orthoses on the human foot and lower extremity, we have undertaken a study to measure the response of the kinematics and kinetics of the rearfoot to simple rearfoot varus and valgus wedges during walking.

Methods

Participants

Twelve volunteers (six males and six females) were enrolled in the study. Participant ages ranged from 19 to 35 years (mean ± SD, 24.58 ± 5.56 years), and their weights ranged from 47.1 to 92.8 kg (mean ± SD, 62.81 ± 13.7 kg). All of the participants were healthy asymptomatic volunteers who were previously interviewed to determine whether they met the inclusion criteria. The exclusion criteria for the study included the presence of pain or evident gross deformity of the foot and lower extremity, treatment with insoles or braces during the past year, inflammatory diseases or connective tissue disorders, and a history of foot and ankle surgery or past serious trauma to the foot or lower extremities. Written consent was obtained from all of the study participants after verbal and written explanation of the project.

Instrumentation

Kinematic and kinetic data were collected with a seven-camera motion analysis system (Vicon 370; Oxford Metrics Ltd, Oxford, England), sampling at 50 Hz, and two force plates (Advanced Mechanical Technology Inc, Newton, Massachusetts), sampling at 500 Hz. Ground reaction force data were collected with the force plates, which were embedded in the floor and placed in the center of the walkway. Fifteen reflective markers were positioned in the lower limb according to the Helen Hayes marker set described by Kadaba et al.30 Markers were attached to the skin surface of the two anterior superior iliac spines, the sacrum, the lateral midthigh, the lateral femoral epicondyle, the lateral malleolus, the second metatarsal head, and the posterior calcaneus of both lower legs. The markers were spheres 25 mm in diameter covered with retroreflective tape and applied to the skin with double-sided tape. Using the three-dimensional (3-D) position of the markers, the centers of the hip, knee, and ankle joints were calculated together with the local coordinate system for the pelvis, femur, shank, and foot segments. The hip center was calculated from the position of the pelvis markers with the method described by Davis et al31 in which the depth and width of the pelvis are used to calculate the hip joint center. The foot was defined as consisting of a single vector from the ankle joint center to the second metatarsal head.

Lower-limb kinematic and kinetic variables were calculated with Vicon Clinical Manager software (Oxford Metrics Ltd). Angular displacements were calculated with Euler angles derived from joint coordinate systems.32, 33 Inverse dynamics were then used to calculate 3-D moments at the lower-limb joints combining kinetic data from ground reaction forces with segment masses, velocities, accelerations and moments of inertia, and position of the joint centers. The 3-D joint moments were calculated relative to the axes of local coordinate systems of the distal segment. Joint moments were defined as being net internal moments and were normalized to the body mass of each participant.

Protocol

Simple 7° wedges of high-density ethyl vinyl acetate (Shore >65) were used in this study. Each participant was studied in three different walking conditions: 1) barefoot, 2) wearing a 7° rearfoot varus wedge, and 3) wearing a 7° rearfoot valgus wedge. Wedges were 14 cm long and 4 cm wide and were positioned with the thickest border of the wedge under the medial and lateral borders of the heel for the rearfoot varus and valgus wedge conditions, respectively. The rearfoot wedges were firmly adhered to the plantar heel with double-sided tape to secure the wedges so that markers could be clearly visualized and the participant could walk in a normal manner.

The same tester (J.P.H.) performed all of the marker placements. Once skin markers were positioned, a static anatomical landmark calibration trial was performed with the motion analysis system to determine the reference angles and anatomical landmarks. Each participant then walked along a 20-m walkway at a self-selected speed. After 5 min of warm-up walking, ten to 12 trials were captured in each of the three walking conditions (ie, barefoot, 7° rearfoot varus wedge, and 7° rearfoot valgus wedge) so that one foot landed on one of the force plates first and then the next foot landed on the second force plate. No trials were accepted in which there was evidence of force plate targeting by the participants. Three hundred gait trials were collected during the study (12 participants × 10 trials × 3 conditions).
Statistical Analysis

Three gait parameters were measured and analyzed: the foot progression angle, the foot rotation angle, and net ankle inversion moments. The foot progression angle was defined as the transverse plane angle between the longitudinal foot vector and the laboratory sagittal axis. For statistical analysis, the maximum foot progression angle (ie, abduction of the foot relative to the line of progression) was used. The foot rotation angle was defined as the transverse plane angle between the longitudinal foot vector and the sagittal axis of the shank and measured internal and external rotation of the tibia relative to the foot (Fig. 1). For statistical analysis, the maximum foot rotation angle (ie, abduction of the foot relative to the tibia) was used during the stance phase. Net ankle inversion moments were calculated relative to the ankle joint center at the point of peak internal tibial rotation for each participant. The mean values from six to eight trials in each participant in each of the three conditions were used for statistical analysis. To avoid statistical error, only the right limb of each participant was included in the analysis.34, 35

Because of the small sample, data were considered to be nonnormally distributed, and the Wilcoxon matched pairs signed rank test was used to calculate differences among the three conditions for the parameters measured. Differences were considered significant at $P \leq .05$. Results are expressed as mean (range). Statistical analysis was performed with a software program (SPSS version 12.0; SPSS Inc, Chicago, Illinois).

Results

Table 1 summarizes the results of foot progression, foot rotation, and ankle inversion moment analysis. There were statistically significant differences in the foot progression angle comparing walking barefoot ($10.93^\circ$; $0.69^\circ$ to $15.99^\circ$) with walking with the rearfoot varus wedges ($9.29^\circ$; $0.38^\circ$ to $16.38^\circ$) ($P = .015$) and comparing walking with the rearfoot varus wedges ($9.29^\circ$; $0.38^\circ$ to $16.38^\circ$) with walking with the rearfoot valgus wedges ($10.56^\circ$; $1.91^\circ$ to $17.22^\circ$) ($P = .008$). No significant differences were found comparing walking barefoot with walking with valgus wedges ($P = .937$).

There were no statistically significant changes regarding foot rotation (peak internal tibial rotation during stance) in the three conditions (Table 1). There were significant differences in net ankle inversion moments comparing walking barefoot ($-0.019$; $-0.070$ to $0.077$) with walking with varus wedges ($0.007$; $-0.046$ to $0.113$) ($P = .012$) and comparing walking with varus wedges ($0.007$; $-0.046$ to $0.113$) with walking with valgus wedges ($-0.035$; $-0.080$ to $0.015$) ($P = .003$). There were no significant changes comparing walking barefoot with walking with valgus wedges ($P = .147$).

Discussion

The present study did not show significant differences in rearfoot kinematics with the use of $7^\circ$ rearfoot varus and valgus wedges during normal walking. Peak internal tibial rotation during the stance phase of walking occurred between 13% and 19% of the whole gait cycle in the study participants, and neither the rearfoot varus nor the valgus wedges significantly increased or decreased peak internal tibial rotation (Fig. 1). These results are different from those of previous research that studied the effect of insoles in walking in which a small but consistent change in rearfoot kinematics was found. However, these other studies also used insoles that differed from the present simple rearfoot wedges either in longitudinal arch contour21, 23, 25, 27, 28 or in rearfoot varus or valgus angle ($10^\circ$).24, 35 There were, however, differences in the foot progression angle (the angle between the foot axis and the laboratory fixed axis) between the rearfoot varus and valgus wedge conditions. The reasons for this finding are not clear, and the changes obtained were so small (<1.5°) that their clinical application is uncertain.
Joint moments calculated by the system were defined as net internal joint moments and represent the body’s response to the loads associated with gait, including ground reaction forces, segment weight, and segment inertia. Results of the present study show significant changes in net ankle inversion moments between the barefoot and the rearfoot varus wedge conditions ($P = .012$) and between the rearfoot varus and valgus wedge conditions ($P = .003$) during the initial timing of the stance phase (Fig. 2). The 7° rearfoot varus wedge produced a significant decrease in net internal ankle inversion moments compared with the barefoot and the 7° rearfoot valgus wedge conditions, which means that the tissues that cause internal ankle inversion moments (e.g., the posterior tibial muscle and the deltoid ligament) were likely subjected to decreased stress when the individual walked with a varus wedge compared with walking barefoot and walking with a valgus wedge. In other words, compared with the varus wedge condition, the barefoot and rearfoot valgus wedge conditions produced a significant increase in net internal ankle inversion moments, which increased the stress in the tissues that produce varus moments at the ankle-subtalar complex during the initial timing of the stance phase of gait.

Ankle inversion moments showed no significant differences in the rearfoot valgus wedge and the barefoot conditions ($P = .147$). These results are contrary to those of Kakihana et al. in which a significant increase in valgus moments of the subtalar joint was found with a 6° valgus wedge compared with being barefoot. One possible explanation could be derived from the calculation of ankle inversion moments at

Table 1. Comparison of Rearfoot Kinematics and Kinetics Measured with the Three Studied Conditions in 12 Feet

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Intervention</th>
<th>$P$ Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot progression angle (peak foot</td>
<td>Barefoot</td>
<td>.015</td>
</tr>
<tr>
<td>abduction during stance) (°)</td>
<td>Varus wedge</td>
<td></td>
</tr>
<tr>
<td></td>
<td>10.93 (0.69 to 15.99)</td>
<td>9.29 (0.38 to 16.38)</td>
</tr>
<tr>
<td>Barefoot</td>
<td>Valgus wedge</td>
<td>.937</td>
</tr>
<tr>
<td>10.93 (0.69 to 15.99)</td>
<td>10.56 (1.91 to 17.22)</td>
<td></td>
</tr>
<tr>
<td>Varus wedge</td>
<td>Valgus wedge</td>
<td>.008</td>
</tr>
<tr>
<td>9.29 (0.38 to 16.38)</td>
<td>10.56 (1.91 to 17.22)</td>
<td></td>
</tr>
</tbody>
</table>

Foot rotation angle (peak internal tibial rotation during stance) (°)

<table>
<thead>
<tr>
<th>Intervention</th>
<th>Barefoot</th>
<th>Varus wedge</th>
<th>.308</th>
</tr>
</thead>
<tbody>
<tr>
<td>20.80 (29.50 to 10.50)</td>
<td>21.49 (28.51 to 11.97)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Barefoot</td>
<td>Varus wedge</td>
<td>.638</td>
<td></td>
</tr>
<tr>
<td>20.80 (29.50 to 10.50)</td>
<td>21.00 (31.13 to 9.98)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Varus wedge</td>
<td>Valgus wedge</td>
<td>.814</td>
<td></td>
</tr>
<tr>
<td>21.49 (28.51 to 11.97)</td>
<td>21.00 (31.13 to 9.98)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Net ankle inversion moments (Nm/kg)

<table>
<thead>
<tr>
<th>Intervention</th>
<th>Barefoot</th>
<th>Varus wedge</th>
<th>.012</th>
</tr>
</thead>
<tbody>
<tr>
<td>−0.019 (−0.070 to 0.077)</td>
<td>0.007 (−0.046 to 0.113)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Barefoot</td>
<td>Varus wedge</td>
<td>.147</td>
<td></td>
</tr>
<tr>
<td>−0.019 (−0.070 to 0.077)</td>
<td>−0.035 (−0.080 to 0.015)</td>
<td></td>
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<tr>
<td>Varus wedge</td>
<td>Valgus wedge</td>
<td>.003</td>
<td></td>
</tr>
<tr>
<td>0.007 (−0.046 to 0.113)</td>
<td>−0.035 (−0.080 to 0.015)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

$^a$Results are expressed as mean (range). Abduction, internal rotation, and varus are expressed as positive values.

$^b$Wilcoxon matched pairs signed rank test.

Figure 2. Net ankle internal inversion moments in the three walking conditions for one participant. The vertical lines indicate the end of the stance phase for each condition.
the point of peak internal tibial rotation, which is different from that in the studies by Kakihana et al. Because the peak ankle inversion moment occurred relatively early in the gait cycle (ie, 13%–19%), soon after heel contact, the point of lateral heel contact during walking was not changed significantly with the valgus wedge. However, because use of a varus wedge could shift the location of the center of pressure more medially compared with the barefoot condition, it is not surprising that the rearfoot varus wedge showed more significant changes in ankle kinetics than did the rearfoot valgus wedge compared with the barefoot condition.

Previous kinematic studies that have investigated changes in skeletal alignment with foot orthoses have shown mixed results that suggest that foot orthoses could be therapeutically effective by methods other than by producing changes in skeletal alignment. In running research, foot orthoses have also shown mostly a kinetic effect rather than a kinematic effect. In walking research, valgus wedges have also been shown to have mostly a kinetic effect and have little to no kinematic effect. The present results support the hypothesis that the main effect of foot orthoses during walking is their ability to alter kinetics rather than kinematics.

Analyzing joint kinetics allows direct access to information regarding the forces and moments that act internally in the foot and lower extremity during weightbearing activities and, as such, allows a better understanding of the stresses in the tissues that may become injured during these activities. The “tissue stress model,” first proposed by McPoil and Hunt and further discussed more recently by Kirby and Fuller, suggest that mechanical therapy should be directed toward resolving tissue stress rather than toward preventing compensation for foot deformities. The goal of this approach is to remove the pathologic stress from the injured structure to optimize healing and gait and to prevent new musculoskeletal pathologic abnormalities in the future. Because varus wedges decreased net ankle inversion moments in the present study, they also decreased the stress in the tissues that cause subtalar and ankle joint inversion moments. In this context, it could be hypothesized that foot orthoses work by decreasing stress by primarily changing the kinetics, with kinematics playing a secondary role.

This study has some limitations that may make clinical correlation difficult. First, the mechanical intervention orthoses used were 7° simple rearfoot wedges, not foot orthoses. Extrapolation of these results to other types of orthoses, such as custom-made orthoses, should be performed cautiously. Mündermann et al showed that the effect of isolated posting varied during running when combined with molding of the arch. It would be helpful to study the effect observed in this study with other types of custom foot orthoses during normal walking at different moments during the stance phase. Second, the study comprised asymptomatic volunteers, and effects of rearfoot wedges on symptomatic individuals may be different. Because we used asymptomatic volunteers, it is unknown whether the differences observed in ankle inversion moments were clinically relevant in terms of pain reduction or alleviation of symptoms. Finally, the study of joint moments has shown significant differences in joint moment profiles with alternative reference frames, especially in the frontal and transverse plane. Moments were reported in the reference frame of the segment distal to the joint because it was thought that it would be more useful to examine the effect of ankle moments on the foot by reporting moments relative to the cardinal planes of the foot. Therefore, the results and the ankle frontal plane moment profiles obtained herein cannot be directly compared with those of other studies using different reference frames.

One additional potential problem could be that some marker sets, such as the Helen Hayes marker set used in this study, model the foot as a single vector and not as a segment. Therefore, foot kinematics could be calculated in the present study only in the sagittal plane (movement of the foot vector to the tibia in the sagittal plane) and the transverse plane (movement of the foot vector to the tibia in the transverse plane) but not in the frontal plane. Other studies have also used tibial rotation in the transverse plane as an indirect measure of subtalar joint motion. Indeed, it has been suggested that measurement of transverse plane motion between the foot and the shank may provide a more complete reference for rearfoot function during weightbearing activities because it provides a description of the composite function of the midfoot, midtarsal, rearfoot, and ankle joints. Although foot kinematics could be measured only in the sagittal and transverse planes in the present study, foot kinetics (referred to as ankle moments) could be measured in all three planes. Therefore, transverse plane tibial rotation relative to the foot was used to measure rearfoot kinematics, and ankle inversion/eversion moments were used to measure rearfoot kinetics in the frontal plane. Whether the choice to use the Helen Hayes marker set to determine joint kinematics and kinetics in this study influenced the results is unknown.
Conclusions

This study found that 7° rearfoot varus and valgus wedges had no effect on peak tibial internal rotation during the stance phase of normal walking, although there were significant differences in ankle inversion moments in the rearfoot varus condition compared with the barefoot and valgus wedge conditions. This finding suggests that wedged insoles may have a much more significant effect on rearfoot kinetics than on rearfoot kinematics and may explain why foot orthoses that are therapeutically effective may not significantly alter foot position and motion during walking.

Acknowledgment: The Department of Human Anatomy and Embryology of Universidad Complutense de Madrid, the Gait Analysis Laboratory of Clinica Universitaria de Podología (Universidad Complutense de Madrid), and Ernesto Maceira Suarez, MD, for support in the development of this work.

Financial Disclosure: None reported.

Conflict of Interest: None reported.

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