Late Rearfoot Eversion and Lower-limb Internal Rotation Caused by Changes in the Interaction between Forefoot and Support Surface

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Background: The influence of distal mechanical factors that change the interaction between the forefoot and the support surface on lower-limb kinematics is not well established. This study investigated the effects of the use of lateral wedges under the forefoot on the kinematics of the lower extremity during the stance phase of walking.

Methods: Sixteen healthy young adults participated in this repeated-measures study. They walked wearing flat sandals and laterally wedged sandals, which were medially inclined only in the forefoot. One wedged sandal had a forefoot lateral wedge of 5° and the other wedged sandal had a forefoot lateral wedge of 10°. Kinematic variables of the lower extremity, theoretically considered clinically relevant for injury development, were measured with a three-dimensional motion analysis system. The variables were evaluated for three subphases of stance: loading response, midstance, and late stance.

Results: The 5° laterally wedged sandal increased rearfoot eversion during midstance and the 10° laterally wedged sandal increased rearfoot eversion during mid- and late stances, in comparison to the use of flat sandals. The 10° laterally wedged sandal produced greater internal rotation of the shank relative to the pelvis and of the hip joint, during the midstance, also compared to the use of flat sandals.

Conclusions: Lateral wedges under the forefoot increase rearfoot eversion during mid- and late stances and may cause proximal kinematic changes throughout the lower-extremity kinetic chain. Distal mechanical factors should be clinically addressed when a patient presents late excessive rearfoot eversion during walking. (J Am Podiatr Med Assoc 99(6): 503-511, 2009)

Excessive foot pronation, usually measured as excessive rearfoot eversion, during the stance phase of gait, has been supposed to increase the possibility of developing several musculoskeletal injuries.[1-6] Late excessive rearfoot eversion is a specific altered motion pattern that takes place after the forefoot contacts the support surface and is characterized by increased eversion and decreased inversion angles between the calcaneus and the shank, during the mid- and late stances of walking.[2] This kinematic behavior may lead to increased susceptibility of developing specific foot and ankle pathologies, consequent to the interaction between this pattern and mechanical features of mid- and late stances.[1, 2, 4-6] Furthermore, biomechanical models suggest that an increase in rearfoot eversion, a component of pronation at the subtalar joint, causes an internal rotation increase and an external rotation decrease of the lower extremity, through the coupling between talus and shank.[2, 4] Talus adduction (internal rotation) is one component of the talus motion that occurs during subtalar pronation in

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a closed kinematic chain.[2] Therefore, the occurrence of excessive subtalar pronation and thus of excessive rearfoot eversion, during mid- and late stances, could cause the shank or thigh segments to assume more internally rotated positions during these periods, changing knee or hip joint mechanics.[2, 4, 7] Consequently, excessive mechanical stresses and resultant injuries might also take place at proximal joints and structures.[2, 4] Although these theoretical couplings are frequently accepted,[2, 4] they have rarely been empirically tested.[7, 8]

Rearfoot motion and lower-extremity transverse plane rotation during walking have been frequently explained by the interaction between the foot and the ground reaction force and thus by torques generated at the foot-ankle complex.[4, 9] Late excessive rearfoot eversion is frequently attributed to the presence of excessive varus alignment of the forefoot and first-ray hypermobility, which may increase eversion torques at the foot-ankle complex after the forefoot contacts the support surface.[1, 2, 4, 10] However, rearfoot motion and lower-extremity transverse plane rotation possibly have a multifactorial nature.[2, 11-13] They might be influenced both by forces originated distally, at the foot-ankle complex,[2, 4, 9] and forces originated proximally, mainly at the hip joint.[11-13] Bellchamber and van den Bogert[13] analyzed foot-ankle complex kinetics through inverse dynamics in order to study the contributions of distal and proximal torques to the production of internal and external rotations of the shank during the stance phase of walking. They observed that foot-ankle kinetics alone cannot explain shank rotation patterns during stance. Moreover, this analysis suggested that proximal forces were crucial for the production of shank external rotation and the coupled rearfoot inversion, during late stance.[13] These results raise questions about the influence of distal mechanical factors on late rearfoot motion. Therefore, the influence of distal mechanics in the occurrence of late excessive rearfoot eversion and of the coupled altered motions of proximal segments is still unknown.

The aim of the present study was to investigate the effects of the use of lateral wedges under the forefoot on the kinematics of the lower extremity while in the stance phase of walking.

Methods

Participants

Sixteen young healthy participants (8 male and 8 female) from the Universidade Federal de Minas Gerais community volunteered to participate in the study. The subjects had maximum varus alignments of 4°, 7°, and 4°, for the calcaneus, forefoot, and tibia, respectively.[2, 14, 15] They did not have any valgus alignment of these structures. They also had normal ranges of motion of eversion/inversion at the ankles and internal/external rotation at the hips.[16, 17] Furthermore, participants with lower-extremity length differences greater than 1 cm or who have experienced any pain or pathology in the feet, ankles, hips, pelvis, and spine were excluded from the study. Participants' characteristics are shown in Table 1.

Instrumentation

Kinematic data about the right lower extremity and pelvis were obtained with a four-camera motion analysis system (ProReflex; Qualisys Medical AB, Gothenburg, Sweden). This system has an accuracy of 0.6 mm as specified by the manufacturer. Anatomical markers were placed on the following anatomical locations: iliac tubercles, greater trochanters, lateral and medial epicondyles, lateral and medial malleoli, and the heads of the first and fifth metatarsals.[18] Clusters, containing tracking markers, were attached to the participant's calcaneus, shank, and thigh of the right lower extremity and pelvis. Pelvic cluster consisted of a rigid basis containing three tracking markers and was attached to the pelvis on the sacrum posterior surface with elastic fastenings, which involved the entire segment. The clusters of the thigh were nonrigid, comprising neoprene girdles with three tracking markers each.[19] The clusters of the shank were rigid, consisted of rigid plates attached to the neoprene girdles, with three tracking markers.[20] The clusters were firmly secured to the distal half of the segment by Velcro (Velcro USA, Manchester, New Hampshire). These clusters and their positions were chosen in an attempt to accomplish better estimates of femur and tibial rotations, regarding the soft-tissue artifacts present in transverse plane kinematic data of the lower limb.[19, 20] The cluster of the calcaneus consisted of three rigid wands, each with a tracking marker fixed on their tip,

Table 1. Descriptive Data of the Study Participants (N = 16)

Variable	Mean	SD	Range
Age (y)	23.4	2.4	20.0–30.0
Mass (kg)	63.8	7.6	49.5–80.0
Height (m)	1.7	0.06	1.6–1.8
Rearfoot (°)[a]	2.3	1.7	0.0–4.0
Forefoot (°)[a]	5.0	2.1	0.0–7.0
Tibia (°)[a]	3.4	1.0	0.0–4.0

[a]Positive values indicate varus alignment.

firmly attached to the metallic basis, which involved the calcaneus posteriorly, laterally, and medially.[18] These clusters were attached just below the insertion of the tendon of the calcaneus (Fig. 1).

Procedures

This research was approved by the Universidade Federal de Minas Gerais Ethics in Research Committee and the participants signed an informed consent form agreeing to participate in the study.

A reference position was obtained with the participant in a relaxed standing position, with the anatomical markers and clusters. Subsequently, the anatomical markers were removed and kinematic data were collected in three different conditions: 1) control condition: the participant walked wearing flat sandals; 2) 5° experimental condition (5° EC): the participant walked wearing a flat sandal on the left foot and a sandal with a 5° lateral wedge in the forefoot of the right foot; and 3) 10° experimental condition (10° EC): the participant walked wearing a flat sandal on the left foot and a sandal with a 10° lateral wedge in the forefoot of the right foot. The wedged sandals were flat at the rearfoot and laterally wedged (medially depressed) under the forefoot, beginning at approximately 3 cm before the metatarsal heads (Fig. 2). The sequence of these conditions was randomized and the right lower limb was chosen for standardization of the experimental conditions.

The height of the lateral surface of the experimental sandals' laterally wedged portion was equal to the height of the flat portion surface of the same sandals. This configuration distanced the anterior sandal surface from the forefoot medial border, when the subtalar joint was placed in neutral position, similar to the presence of forefoot varus.[2, 4] The sandal bases were manufactured from high-density ethylene vinyl ac-

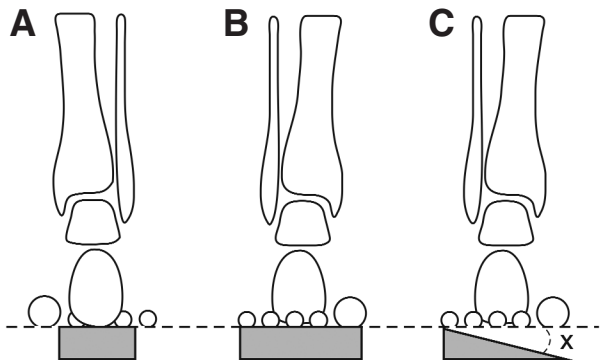


Figure 2. Schematic representation of the surface inclinations of the sandals' bases. A, Posterior view showing the horizontally aligned surface of the rearfoot portion of the sandals' bases (control and experimental). B, Anterior view of the horizontally aligned surface of the forefoot portion of the control sandal's base. C, Anterior view of the medially inclined surfaces of the experimental sandals' bases. X represents the angle of medial inclination (5° or 10°); the broken line represents the horizontal reference.





Figure 1. Placement of the reflective markers. A, Posterior view showing the anatomical and tracking markers. B, Detailed view of the right foot markers, rearfoot cluster, and sandals markers.

etate (EVA) and were attached to the participants' feet with Velcro. Each right sandal had two tracking markers attached to its lateral surface to enable identification of specific events and duration of the stance phase of walking.[21]

For each condition, the participants walked along an 8-m walkway at their natural preferred speed. Ten trials in each condition were carried out, with a data collection frequency of 120 Hz. Before data collection in each study condition, the participants walked on a 100-m walkway wearing each sandal for familiarization.

Data Processing

The data were processed through the Visual 3D Motion Analysis Software (C-Motion, Inc, Rockville, Maryland). The reference position data were used for creating the rigid bodies corresponding to the pelvis, shank, thigh, and foot segments and to determine the neutral position of the joints. The positions of the anatomical markers were used for attributing coordinate systems for each segment.[18] The X-axis was defined as anteroposterior, the Y-axis as mediolateral, and the Z-axis as proximal-distal. The clusters were associated with each corresponding segment, and the data were smoothed using a zero-lag fourth-order Butterworth lowpass filter with a cut-off frequency of 6 Hz.[22]

The duration of the stance phase of each trial was determined according to Ghoussayni et al.[21] This procedure enabled the identification of the following events: heel contact (HC), forefoot contact (FC), heel rise (HR), and toe off (TO). The stance phase was then divided into three subphases: loading response (period between HC and FC), midstance (period between FC and HR), and late stance (period between HR and TO).

Definition of Variables

Angular variables were calculated as the motion of a distal segment relative to a proximal segment. Rearfoot eversion/inversion was calculated as the calcaneus motion relative to the shank in the frontal plane (X-axis). Shank internal/external rotation in relation to the pelvis was calculated as the shank movement relative to the pelvis in the transverse plane (Z-axis); and the hip internal/external rotation was calculated as thigh motion relative to the pelvis in the transverse plane (Z-axis).

Each mean angle was calculated for each of the three stance subphases. Mean angular values for entire periods were chosen instead of peak or other discrete angular variables in order to obtain quantity estimates of the continuous angular displacements that took place during the analyzed subphases.

Statistical Analysis

Two-way repeated measures analyses of variance (ANOVA) with two within-subjects factors and each factor with three levels (experimental conditions: control, 5° EC, 10° EC; stance subphases: loading response, midstance, late stance) were used for statistical analyses. One ANOVA was carried out for each dependent variable (angles). When the ANOVA identified significant interaction effects, preplanned contrasts were made to test for significant differences in specific comparisons (control × 5° EC and control × 10° EC) for each stance subphase. The α level for the analyses was established at 0.05.

In order to investigate the influence of possible measurement errors on the results, the standard errors of measurement were determined for the variables that showed significant differences. Two trials of the control condition were randomly selected for the calculation of the intraclass correlation coefficients (ICC) correspondent to the inter-trial reliabilities of each of those variables. A standard error of measurement value larger than the mean difference observed for a variable indicates that such difference may be attributable to inherent measurement errors related to the procedures.[23]

Results

Interaction effects between experimental conditions and stance subphases were significant for all variables ($P < .05$). Means, standard deviations, and P values are presented in Table 2. Significant mean differences, ICCs, and standard error of measurements are presented in Table 3.

Rearfoot Eversion/Inversion

The use of the 5° laterally wedged sandal caused a mean (±SD) increase in rearfoot eversion of 0.88° ± 1.13° (95% confidence interval [CI] = 0.27–1.49; F = 9.65; $P = .007$) compared to the control condition, during the midstance. The use of the 10° laterally wedged sandal generated increases in rearfoot eversion of 1.35° ± 1.03° (95% CI = 0.80–1.90; F = 27.78; $P < .001$), during the midstance, and of 0.83° ± 1.45° (95% CI = 0.06–1.61; F = 5.29; $P = .036$), during the late stance, in comparison to the control condition (Fig. 3A). The standard error of measurement of this vari-

Table 2. Means and SDs of the Dependent Variables and *P* Values of the Comparisons Among the Experimental Conditions and the Control Condition, in each Subphase of Stance

Variable	Loading Response			Midstance			Late Stance		
	Control	5° EC	10° EC	Control	5° EC	10° EC	Control	5° EC	10° EC
Rearfoot eversion/inversion (°)	0.06 (4.65)	0.41 (4.77) *P* = .144	0.21 (4.56) *P* = .61	–2.88 (3.49)	–3.77[a] (3.88) *P* = .007	–4.24[a] (3.82) *P* < .001	0.51 (3.36)	–0.09 (4.20) *P* = .143	–0.32[a] (3.39) *P* = .036
Shank versus pelvis internal/external rotation (°)	–2.15 (6.70)	–2.01 (6.81) *P* = .751	–1.75 (6.31) *P* = .484	1.12 (7.44)	1.96[a] (7.81) *P* = .016	2.53[a] (7.34) *P* = .001	–0.09 (8.41)	0.19 (8.56) *P* = .467	0.41 (8.07) *P* = .162
Hip internal/external rotation (°)	6.50 (7.35)	6.87 (7.11) *P* = .319	6.79 (6.21) *P* = .571	5.46 (7.63)	6.18[a] (7.92) *P* = .022	6.85[a] (7.65) *P* = .001	–0.67 (8.42)	–0.19 (8.41) *P* = .208	–0.08 (7.66) *P* = .141

Abbreviations: 5° EC, experimental condition with the 5° lateral wedge under the forefoot; 10°EC, experimental condition with the 10° lateral wedge under the forefoot.

Note: Positive values refer to inverted or internally rotated positions. Negative values refer to everted or externally rotated positions.

[a]Significantly different from control condition (*P* < .05).

able in the midstance was 0.35° and in the late stance was 0.47°. These values were smaller than the mean changes generated by both wedged sandals.

Shank Internal/External Rotation in Relation to the Pelvis

The use of the 5° laterally wedged sandal generated an increase in shank internal rotation relative to the pelvis of 0.83° ± 1.24° (95% CI = 0.17–1.50; F = 7.31; *P* = .016) and the use of the 10° laterally wedged sandal caused an increase in shank internal rotation relative to the pelvis of 1.41° ± 1.36° (95% CI = 0.68–2.13; F = 17.03; *P* = .001), in comparison to the control condition, only during the midstance (Fig. 3B). The standard error of measurement of this variable was 1.28° and thus larger than the mean change generated by the 5° laterally wedged sandal and smaller than the mean change generated by the 10° laterally wedged sandal.

Hip Internal/External Rotation

The use of the 5° laterally wedged sandal generated an increase of 0.71° ± 1.11° (95% CI = 0.11–1.30; F = 6.47; *P* = .022) and the use of the 10° laterally wedged sandal caused an increase in hip internal rotation of 1.38° ± 1.33° (95% CI = 0.67–2.10; F = 17.35; *P* = .001), in comparison to the control condition, only during the midstance subphase (Fig. 3C). The standard error of measurement of this variable was 1.2° and thus larger than the mean change generated by the 5° laterally wedged sandal and smaller than the mean change generated by the 10° laterally wedged sandal.

Table 3. Mean Differences, Inter-trial ICCs, and Standard Error of Measurements for the Variables Showing Significant Differences

Variable	Comparison	Mean Difference (°)	ICC (3,1)	Standard Error of Measurement (°)
Rearfoot eversion/inversion	Control versus 5° EC (midstance)	0.88	0.99	0.35
	Control versus 10° EC (midstance)	1.35		
	Control versus 10° EC (late stance)	0.83	0.98	0.47
Shank versus pelvis internal/external rotation	Control versus 5° EC (midstance)	0.83	0.97	1.29
	Control versus 10° EC (midstance)	1.41		
Hip internal/external rotation	Control versus 5° EC (midstance)	0.71	0.98	1.21
	Control versus 10° EC (midstance)	1.38		

Abbreviations: 5° EC, experimental condition with the 5° lateral wedge under the forefoot; 10° EC, experimental condition with the 10° lateral wedge under the forefoot; ICC, intraclass correlation coefficient (correspondent to the inter-trial reliability of the control condition).

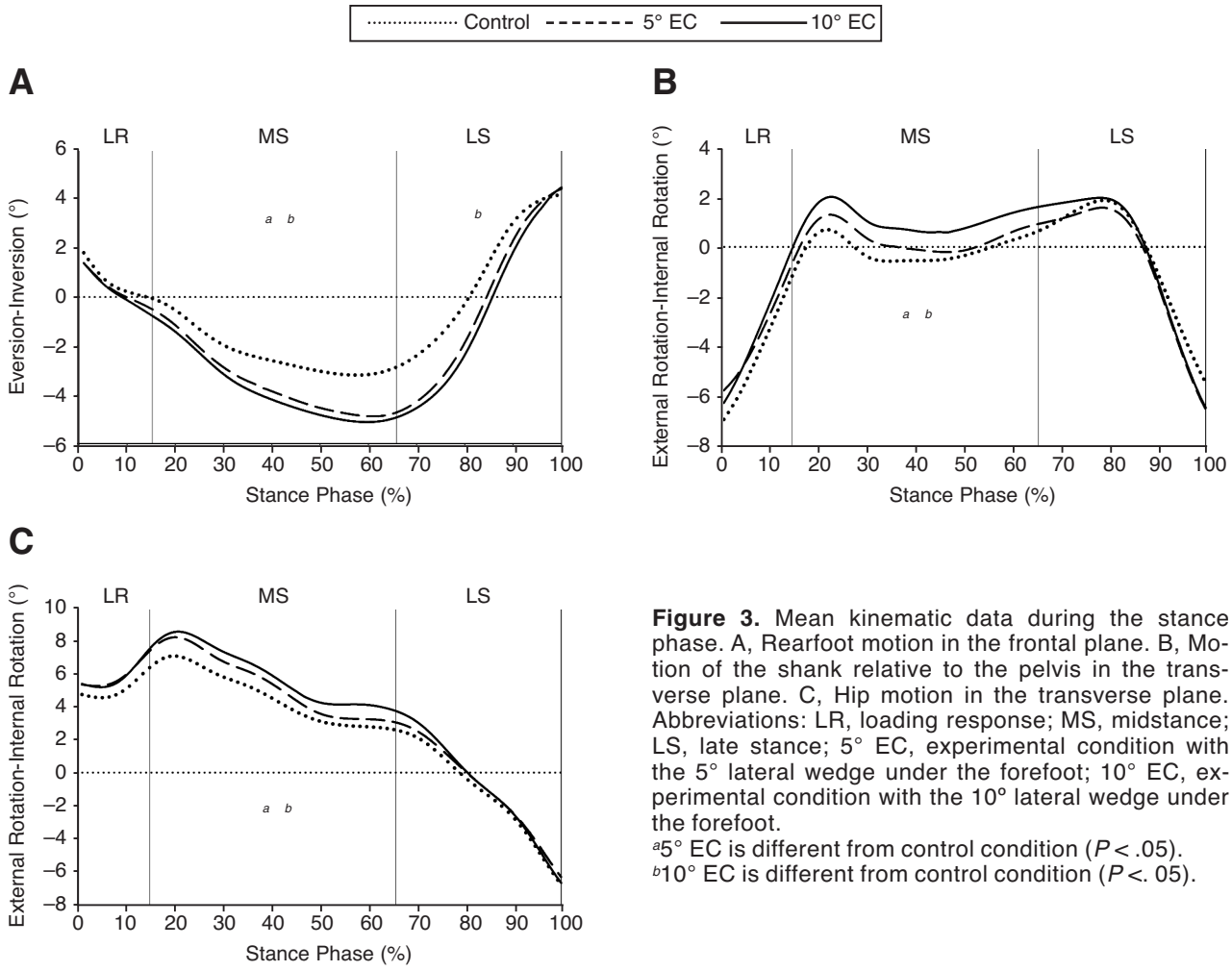


Figure 3. Mean kinematic data during the stance phase. A, Rearfoot motion in the frontal plane. B, Motion of the shank relative to the pelvis in the transverse plane. C, Hip motion in the transverse plane. Abbreviations: LR, loading response; MS, midstance; LS, late stance; 5° EC, experimental condition with the 5° lateral wedge under the forefoot; 10° EC, experimental condition with the 10° lateral wedge under the forefoot.
[a]5° EC is different from control condition ($P < .05$).
[b]10° EC is different from control condition ($P <. 05$).

Discussion

The use of sandals laterally wedged under the forefoot caused kinematic changes in the lower limb. Significant modifications took place only after the forefoot touched the sandal surface, that is, only during the mid- and late stances. The 5° and 10° laterally-wedged surfaces increased rearfoot eversion and internal rotations of the shank in relation to the pelvis and of the hip joint, during the midstance. Only the 10° laterally wedged surface caused increases in rearfoot eversion during late stance. Such effects were greater than the standard error of measurements calculated for each variable and were thus not produced by possible measurement errors related to the study procedures.[23] These kinematic changes demonstrated that distal factors that possibly increase eversion torques at the foot-ankle complex during mid- and late stances may affect lower-extremity mechanics during walking, especially during midstance. Excessive varus alignment of the forefoot, first-ray hypermobility, and inadequate footwear are candidates to generate these torques and, consequently, late excessive rearfoot eversion and the coupled altered movements observed in the present study.[1, 10, 24, 25] Donatelli et al[24] investigated the differences between forefoot alignment of professional baseball players with and without an everted rearfoot position during the whole stance phase of walking. They observed that subjects who maintained the rearfoot everted had greater magnitudes of varus alignment of the forefoot. This result is in accordance with our findings and emphasizes the possible participation of distal mechanical factors in the presence of late excessive rearfoot eversion.

The sandal with the 5° inclination did not increase rearfoot eversion during late stance. These findings suggest that the eversion forces possibly produced were not sufficient to change the sum of forces acting on the shank and rearfoot during this period. Most of

the forces generated by this condition could have been dissipated through motions at the midfoot joints.[26] These findings also suggest the possible crucial role of proximal forces in the kinematic pattern of the foot-ankle complex during late stance.[13] In contrast, the 10° inclination was sufficient to increase rearfoot eversion during late stance, suggesting that this condition generated forces that altered the sum of proximal and distal forces acting on the foot-ankle complex. An increase in shank internal rotation (relative to the foot), and a concurrent increase in knee internal rotation, may have occurred since the knee reaches its maximum flexion position[9] and allows greater transverse plane motion to occur during the late stance.[4, 17] It is noteworthy that, in this condition, the rearfoot continued to invert during late stance, and thus the kinematic changes can be interpreted as a decrease in rearfoot inversion or as a placement of the rearfoot in a less inverted position. This condition was not sufficient to produce a late eversion movement or even prevent late inversion. Therefore, although the 10° laterally wedged surface affected late rearfoot motion, the proximal forces acting on the foot-ankle complex possibly had a relevant role during this period.[13] These findings emphasize the multifactorial nature of rearfoot motion and are in accordance with the observed absence of changes in the external rotation of the hip joint and of the shank in relation to the pelvis, during the late stance.

The kinematic changes observed in the present study are related to immediate interventions that may not have allowed for significant tissue adaptations in the lower extremity, especially at the hip joint. Since there is no activation of the gluteus maximus during mid- and late stances,[12] the external rotation torque possibly generated at the hip during these periods may be, in great part, attributed to forces produced by the passive structures that resist internal hip rotation. The long-term presence of distal factors that cause increased late rearfoot eversion and lower-extremity internal rotation may decrease the stiffness of these structures, and thus is likely to produce greater kinematic changes than those found in the present study.[11]

The literature is not clear about the magnitudes of rearfoot kinematic alterations that may be sufficient to cause pathologic conditions. Viitasalo and Kvist[6] compared rearfoot running kinematics of subjects with and without shin splints and found that this condition was associated with a delayed beginning of rearfoot inversion, suggesting the presence of late excessive rearfoot eversion in these subjects. Furthermore, the presence of shin splints was also associated with a mean increase in maximum eversion of approximately 1°.[6] The present study found that the lateral wedges under the forefoot led to similar mean increases in rearfoot eversion during the midstance. Since the loads imposed upon the musculoskeletal tissues during walking are smaller in comparison to running, it may be argued that the isolated kinematic effects observed in the present study are not likely to produce pathologies. However, it should be considered that the occurrence of overuse injuries has a multifactorial etiology and, hence, other factors such as frequency and duration of the activity and the individual's capability to deal with mechanical stresses must be addressed.[11, 27] Studies that investigated the effects of foot orthoses in the treatment of patellofemoral pain found positive clinical outcomes with a mean decrease in rearfoot eversion of 1.8° during walking.[28, 29] These findings suggest that small changes in lower-limb walking kinematics may constitute clinically relevant changes.

Our results showed kinematic change values between 0.71° and 1.41°, which seem to be extremely small effects. It is important to stress that the variables were defined as the mean values of angle curves within specific stance periods. Such definition generates angle values that do not reveal the magnitude of the differences between peak values of experimental conditions. For instance, the maximum mean difference in rearfoot eversion peaks between the experimental conditions and the control condition were found to be approximately 2° (Fig. 3A), whereas the maximum difference observed for the variables used was 1.41°. Although differences between angle peaks may be larger (Fig. 3), our variable definition was used to capture the continuous joint behavior within defined periods instead of identifying differences between discrete values that may occur at different moments of stance.

The laterally wedged sandals also led to increases in internal rotation of the hip joint and of the shank relative to the pelvis, during the midstance. These findings are in agreement with the theoretical suggestions about the coupling between rearfoot motion and lower-limb transverse rotations in walking[30] and about the possible association of late excessive rearfoot eversion with proximal overuse injuries such as patellofemoral pain,[3, 4] iliotibial band syndrome,[2] trochanteric bursitis,[2] and hip degenerative processes in older adults.[5] We considered the transverse plane movement of the shank in relation to the pelvis to be a possible clinically relevant variable. Increased shank internal rotation relative to the pelvis is often said to strain the structures that link these segments, such as the rectus femoris muscle, iliotibial band, and sartorius muscle, and is theoretically associated to specific injury mechanisms.[2, 31, 32] Even though the effects of

changing rearfoot motion on shank rotation have been previously reported,[7, 33] our results demonstrated that these changes also induce more proximal compensatory motions.

The observed kinematic effects must be interpreted with caution in light of the inaccuracy of quantitative measures of lower-extremity kinematics obtained with external markers, attributable to soft-tissue artifacts.[20, 34] However, cluster configurations and placement were carried out according to recent recommendations.[19, 20, 35] It should be also noted that the shapes of rearfoot eversion/inversion curves measured both with external and bone-anchored markers have good to excellent agreement,[34] which suggests that the quantitative differences between such measures constitute systematic errors. Thus the changes in rearfoot motion, generated in our study, are probably real changes, although the absolute angle values may not correspond to the true bone positions.

In addition to the quantitative inaccuracy,[20] the shank and hip transverse plane motions obtained with external markers[8, 20] usually show poor curve shape agreement with the motion measured with bone-anchored markers.[20, 36] Furthermore, in the present study, the mean changes produced by the 5° laterally wedged sandal on the motion of the shank relative to the pelvis and of the hip were smaller than the standard error of measurements of these variables, indicating that these results could have been influenced by possible inherent measurement errors related to the study procedures.[23] However, the effects of using the 10° laterally wedged sandal on the same variables were greater than the corresponding standard error of measurements. Further, the observed hip motion curve (Fig. 3C) was noticeably similar to the curve pattern obtained with bone-anchored markers.[36] Considering such limitations and features, it is possible to argue that our results showed that distal factors that may produce late excessive rearfoot eversion have proximal mechanical effects and the direction rather than the real magnitude of such proximal kinematic changes was revealed.

Conclusions

The findings of the present study suggest that, although it has been previously demonstrated that proximal forces acting on the foot-ankle complex are predominant during late stance, late excessive rearfoot eversion and lower-limb internal rotation may be a result of the presence of distal mechanical factors that change the interaction between the forefoot and the support surface. The occurrence of these kinematic changes depends on the magnitude of such mechanical factors.

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Conflict of Interest: None reported.

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