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original research

Custom-Molded Foot-Orthosis Intervention and Multisegment Medial Foot Kinematics During Walking

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Context: Foot-orthosis (FO) intervention to prevent and treat numerous lower extremity injuries is widely accepted clinically. However, the results of quantitative gait analyses have been equivocal. The foot models used, participants receiving intervention, and orthoses used might contribute to the variability.

Objective: To investigate the effect of a custom-molded FO intervention on multisegment medial foot kinematics during walking in participants with low-mobile foot posture.

Design: Crossover study.

Setting: University biomechanics and ergonomics laboratory.

Patients or Other Participants: Sixteen participants with low-mobile foot posture (7 men, 9 women) were assigned randomly to 1 of 2 FO groups.

Intervention(s): After a 2-week period to break in the FOs, individuals participated in a gait analysis that consisted of 5 successful walking trials (1.3 to 1.4 m/s) during no-FO and FO conditions.

Main Outcome Measure(s): Three-dimensional displacements during 4 subphases of stance (loading response, midstance, terminal stance, preswing) were computed for each multisegment foot model articulation.

Results: Repeated-measures analyses of variance (ANOVAs) revealed that rearfoot complex dorsiflexion displacement during midstance was greater in the FO than the no-FO condition ($F_{1,14}=5.24$, $P=.04$, partial $\eta^2=0.27$). Terminal stance repeated-measures ANOVA results revealed insert-by-insert condition interactions for the first metatarsophalangeal joint complex ($F_{1,14}=7.87$, $P=.01$, partial $\eta^2=0.36$). However, additional follow-up analysis did not reveal differences between the no-FO and FO conditions for the balanced traditional orthosis ($F_{1,14}=4.32$, $P=.08$, partial $\eta^2=0.38$) or full-contact orthosis ($F_{1,14}=4.10$, $P=.08$, partial $\eta^2=0.37$).

Conclusions: Greater rearfoot complex dorsiflexion during midstance associated with FO intervention may represent improved foot kinematics in people with low-mobile foot postures. Furthermore, FO intervention might partially correct dysfunctional kinematic patterns associated with low-mobile foot postures.

Key Words: foot structure, pronation, supination, orthotics

Key Points

- Rearfoot complex dorsiflexion displacement during midstance increased after a 2-week custom-molded foot-orthosis intervention in participants with low-mobile foot posture.
- Although the average absolute increase in dorsiflexion displacement associated with custom-molded foot-orthosis intervention was small, the relative increase compared with the total dorsiflexion displacement during midstance might represent a clinically relevant change.
- The increase in dorsiflexion displacement, in conjunction with an observed decreased position of plantar flexion at the beginning of midstance, and an earlier transition from a plantar-flexed to a dorsiflexed position associated with foot-orthosis intervention might represent a correction in gait mechanics.

The clinical effectiveness of foot-orthosis (FO) intervention is assumed to result from restoration of normal foot mechanics or removal of abnormal stress during gait. However, authors investigating the quantitative effect of FO intervention on gait mechanics, specifically 3-dimensional walking gait kinematics, have found somewhat inconsistent results.[1–3] Contributing to the inconsistency might be factors such as the foot model, differences in participants receiving FO intervention, and the FO prescribed.

Most researchers investigating the effect of FO intervention on gait kinematics have used single foot segment or rearfoot complex models. Although these models have improved the understanding of the effect of FO intervention on gait, both ignore the joints distal to the calcaneus. However, authors of in vitro stereophotogrammetric,[4,5] in vivo roentgen stereophotogrammetric,[6] and invasive in vivo kinematic[7] studies reported that the joints distal to the calcaneus contribute to foot motion. Furthermore, in 3 recent multisegment foot model studies,[8–10] investigators revealed gait kinematic differences in the joints distal to the calcaneus among participants with different foot postures.

With respect to the FOs used and the participants receiving the orthoses, prefabricated, semicustom, and custom-molded FO interventions have been investigated in people with abnormal foot posture and associated abnormal foot function.[2,3] In another intervention study, Davis et al[1] enrolled healthy participants but did not include any assessment of foot posture or function. If foot posture influences foot function, the kinematic effects of orthosis intervention might vary depending on the foot posture of the participants. Therefore, to study the mechanical effects of FO intervention, investigating the effect of orthosis intervention in participants with abnormal foot function might be important. However, quantifying foot posture is not without challenges because of the poor intertester reliability of most traditional foot classification systems.[11,12] Poor intertester reliability not only might contribute to the variability in study results but also might affect the clinical relevance of the study results. To ensure that the foot postures being investigated in multiple studies are similar, the measures used to quantify foot posture must have moderate to high intratester and intertester reliability. Therefore, the purpose of our study was to investigate the effects of a 2-week FO intervention on multisegment medial foot kinematics during walking in people with low-mobile foot posture. We hypothesized that 2 weeks of FO intervention compared with a no-FO condition would result in increased calcaneonavicular complex abduction displacement during midstance, decreased rearfoot complex inversion during preswing, and increased rearfoot complex eversion during preswing.

METHODS

Participants

We recruited apparently healthy people from Georgia State University and the surrounding community. After initially screening them for current musculoskeletal injuries, we further screened potential participants for eligibility through arch height and foot mobility assessment using the arch ratio in 90% weight bearing and the relative arch deformity (RAD) ratio, respectively.[13] We enrolled 16 participants (7 men, 9 women) with low-mobile foot posture (Table 1). We defined *low arch structure* (arch ratio ≤ 0.287) as an arch ratio equal to or greater than 1 SD below the mean arch ratio assessed in 51 random volunteers (102 feet). We defined a *mobile foot* (RAD ratio > 0.828 10^4/N) as a RAD ratio greater than the mean ratio of the mean assessed in the same 102 feet (Table 1). For the arch ratio, a smaller ratio is associated with a lower arch, and for the RAD ratio, a larger ratio is associated with a more mobile foot. All participants provided written informed consent, and the Institutional Review Board of Georgia State University approved the study.

Table 1. Demographic Data of Participants (Mean ± SD)

Variable	Balanced Foot Orthosis (n = 8)	Full-Contact Foot Orthosis (n = 8)
Age, y	25.4 ± 6.4	25.4 ± 6.7
Height, cm	173.7 ± 10.4	172.2 ± 12.2
Mass, kg	75.3 ± 12.7	72.5 ± 17.4
Arch ratio[a]	0.271 ± 0.009	0.273 ± 0.017
Relative arch deformity ratio, 10^4/N[b]	1.01 ± 0.11	1.18 ± 0.21

[a] Low arch structure (arch ratio ≤ 0.287) was defined as an arch ratio of ≥ 1 SD less than the mean arch ratio assessed in 51 random volunteers (102 feet).

[b] For the relative arch deformity ratio, a larger ratio was associated with a more mobile foot. A mobile foot (relative arch deformity > 0.828 104/N) was defined as a relative arch deformity ratio greater than the mean ratio assessed in 51 random volunteers (102 feet).

Three-Dimensional Motion Analysis

Eight optical video cameras (model TM-6703; PULNiX America, Inc, Sunnyvale, CA) sampling at 120 Hz were used to capture 3-dimensional coordinate data from clusters of 3 or 4 retroreflective markers (8-mm diameter) located on the leg and foot segments of interest. We placed the markers either directly on the skin or mounted on wands constructed from 1.8-mm wire that we fixed to the skin with liquid adhesive and double-sided tape. An AMTI force platform (Advanced Mechanical Technology, Newton, MA) sampling at 960 Hz and an AMTI amplifier (1050-Hz, second-order, critically damped filter with a gain of 1000) were used to determine initial contact and toe-off events. With Peak Performance Motus software (version 8.0; Vicon, Centennial, CO), we synchronized ground reaction force and coordinate data, converted analog signals to digital signals, and filtered the coordinate data with a Butterworth filter using optimal cutoff frequencies determined via residual analysis (range, 2–5 Hz). With custom-written software (Matlab version 7.0.1; The MathWorks, Natick, MA), we performed rigid body-transformation procedures using the calibrated anatomical system technique with a single-value decomposition position and orientation estimator.[14] Next, we computed clinically relevant joint angles between adjacent segments using the joint coordinate system (JCS) technique[15] with positive sagittal-plane, frontal-plane, and transverse-plane rotations defined as plantar flexion, inversion, and adduction of the distal segment on the proximal segment, respectively. The exception was *transverse-plane rotation of the leg segment*, which was defined as medial (positive) rotation of the leg on the calcaneus.[16] Trials for each participant were normalized to 100% of stance and ensemble averaged at 2% intervals. Finally, 3-dimensional displacement within 4 subphases of stance (loading response [0%–16%], midstance [16%–48%], terminal stance [48%–81%], and preswing [81%–100%]) was computed.[17]

Foot Segmentation

Foot segmentation was based on data from in vitro studies,[4] in vivo roentgen stereophotogrammetric studies,[4,6] and the concepts of constrained tarsal mechanism[18] and forefoot twist.[19] The model consisted of 4 functional articulations (rearfoot complex, calcaneonavicular complex, medial forefoot, and first metatarsophalangeal complex). The functional articulations and their local Cartesian coordinate systems are outlined in this subsection, and Cobb et al[8] reported the details of the local reference system computation, reliability of the multisegment foot model, and agreement in the kinematic results between the multisegment foot model and invasive in vivo gait.

Rearfoot Complex. Cartesian coordinate systems defined within the leg and calcaneus segments composed the rearfoot complex (Figure 1). The JCS used to compute sagittal- and frontal-plane rearfoot complex motions was formed by the mediolateral axis of the leg segment, the anteroposterior axis of the calcaneal segment, and a floating axis computed as the cross-product of the calcaneal anteroposterior and leg mediolateral axes. To compute transverse-plane rotation of the leg with respect to the calcaneus, we constructed a separate JCS using the mediolateral axis of the calcaneal segment, the vertical axis of the leg, and a floating axis computed as the cross-product of the calcaneal mediolateral and leg vertical axes. Transverse-plane rotation of the leg relative to the calcaneus then was computed about the vertical axis of the leg.[16]

Calcaneonavicular Complex. Cartesian coordinate systems defined within the calcaneus and the navicular segments formed the calcaneonavicular complex (Figure 1). The JCS used to compute 3-dimensional calcaneonavicular complex movements was formed by the mediolateral axis of the calcaneus segment, the anteroposterior axis of the navicular segment, and a floating axis computed as the cross-product of the navicular anteroposterior and calcaneal mediolateral axes.

Medial Forefoot. The medial forefoot was formed by Cartesian coordinate systems defined within the medial 2 rays[19] and the navicular segment (Figure 1). The JCS used to compute 3-dimensional medial-forefoot motion was formed by the mediolateral axis of the navicular segment, the anteroposterior axis of the medial ray segment, and a floating axis computed as the cross-product of the anteroposterior and navicular mediolateral axes of the medial rays.

First Metatarsophalangeal Complex. The first metatarsophalangeal complex (1MTP) was formed by Cartesian coordinate systems defined within the hallux and medial ray segments (Figure 1). The JCS used to compute 3-dimensional motions of the 1MTP was formed by the mediolateral axis of the medial ray segment, the anteroposterior axis of the hallux segment, and a floating axis computed as the cross-product of the hallux anteroposterior and 1MTP mediolateral axes.

Custom-Molded Foot Orthoses

We used a balanced traditional orthosis (BALORT) constructed with rearfoot and forefoot posting (Foot Levelers, Inc, Roanoke, VA) and a full-contact orthosis (FCORT) that provided support through the medial longitudinal arch with no rearfoot or forefoot posting (Sole Supports, Inc, Lyles, TN) (Figure 2). We chose the 2 orthoses because, although both are designed to correct abnormal foot mechanics, the methods used to affect foot function are very different. The casting procedure for both orthoses involved capturing an impression of the participant's feet in a foam box. To create a cast for the BALORT, we instructed the participant to step into the foam box. Casts for the FCORT were created using the casting procedure of the maximum arch subtalar stabilization position theory of the manufacturer. With the participant seated, we positioned the foot in a foam box. Next, we captured the impression of the participant's foot by pressing down on the thigh, along the lateral border of the foot, on all 5 toes, and on all 5 metatarsal heads. After completing the casting procedures, we sent the impressions to

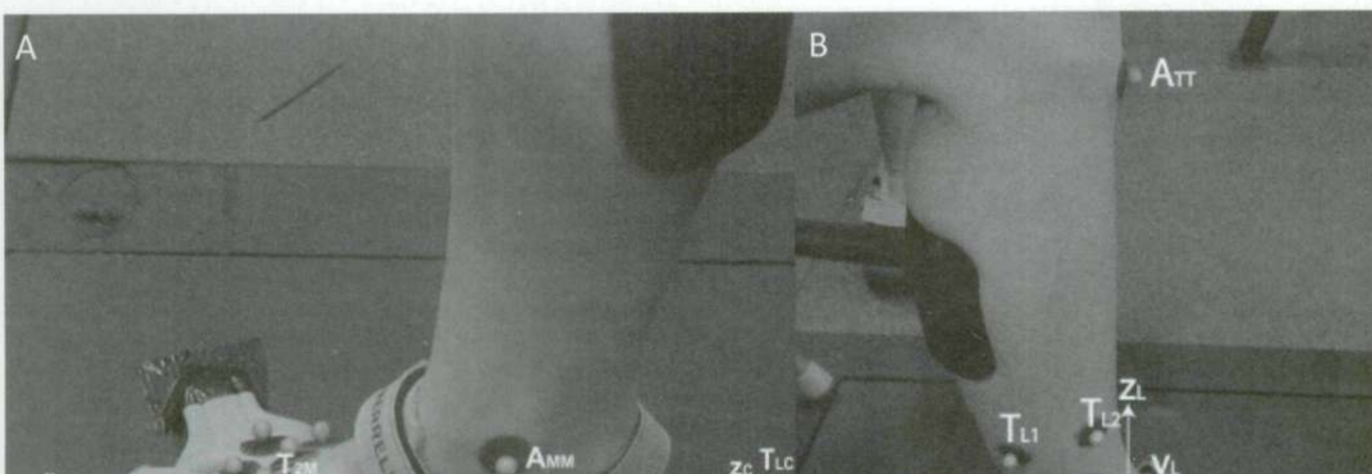

Figure 1. A, Calcaneus (medial technical marker [T_{MC}], lateral technical marker [T_{LC}], apex technical marker [T_{AC}]), navicular (proximal technical marker [T_{PN}], distal technical marker [T_{DN}], apex technical marker [T_{AN}]), medial rays (medial cuneiform technical marker [T_{MCN}], first metatarsal technical marker [T_{1M}], second metatarsal technical marker [T_{2M}], first metatarsal head anatomical marker [A_{1MH}], second metatarsal head anatomical marker [A_{2MH}]), and hallux (medial technical marker [T_{MH}], lateral technical marker [T_{LH}], apex technical marker [T_{AH}]) segment marker clusters. Calcaneus (x_C, y_C, z_C), navicular (x_N, y_N, z_N), medial rays (x_{MR}, y_{MR}, z_{MR}), and hallux (x_H, y_H, z_H) anatomical Cartesian reference systems. Abbreviation: A_{MM}, medial malleolus anatomical marker. B, Leg segment (leg technical marker 1 [T_{L1}], leg technical marker 2 [T_{L2}], leg technical marker 3 [T_{L3}], leg technical marker 4 [T_{L4}], A_{MM} [not shown; see Figure 1A], lateral malleolus anatomical marker [A_{LM}], tibial tuberosity [A_{TT}]) anatomical marker clusters. Leg segment anatomical Cartesian reference systems (x_L, y_L, z_L). The original model also included lateral forefoot and cuboid segments (the additional lateral foot markers); however, because of difficulties with reconstruction of the lateral segment marker clusters, only the medial segments are presented.

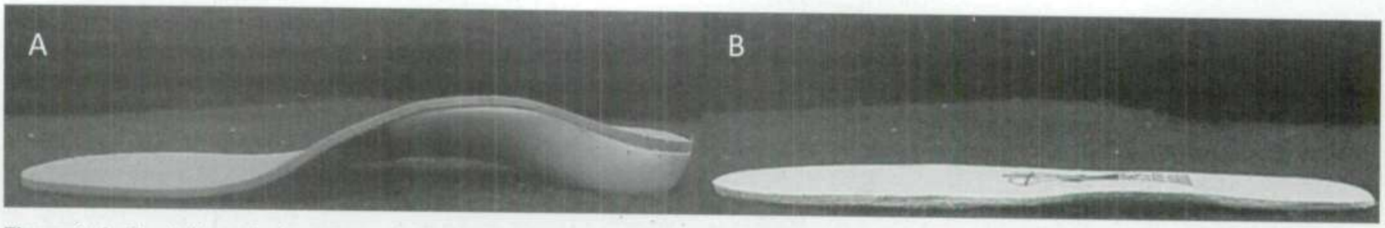

Figure 2. A, The full-contact custom-molded orthosis (Sole Supports, Inc, Lyles, TN) is a polyethylene composite material orthosis constructed to control midfoot motion. B, The balanced custom-molded foot orthosis (Foot Levelers, Inc, Roanoke, VA) is a leather and composite material orthosis based on the Root Functional orthotic.

the manufacturers, who constructed custom-molded orthoses from the casts. The shells of both orthoses extended from the calcaneus to the level of the metatarsal heads and were covered with a full-length vinyl topcover (Figure 2). Upon receipt of the orthoses from the manufacturers, we randomly assigned participants to either the BALORT or FCORT group and provided break-in instructions. Participants were provided with only the assigned FO at the beginning of the study, so they were blinded to differences between the orthoses. We instructed participants to gradually increase the time during which they wore their FOs until they could wear them comfortably for a continuous 8-hour period. After a 2-week break-in period, the participants reported to the university's Biomechanics and Ergonomics Laboratory for gait assessment.

Procedures

Before data collection, we performed dynamic camera calibration (volume=0.5 m×0.4 m×0.9 m). We then applied technical marker clusters and anatomical landmarks to each segment on the participant's right foot and leg and performed an anatomical calibration procedure (Figure 1). All participants wore the same model sandal (Merrell Waterfall; Wolverine World Wide, Inc, Rockford, MI) for testing with and without orthoses. The FOs were secured to the sandals using double-sided carpet tape to prevent slippage during the orthosis condition trials. During the anatomical calibration procedure, the participant was seated with the leg oriented vertically and the midpoint of the calcaneus and second metatarsal aligned parallel to the direction of progression. We chose a semi–weight-bearing reference position because in a weight-bearing position, compensatory motions of the foot and leg already have occurred, so differences between the foot posture groups might be masked. Segmental angles computed during the no-FO anatomical calibration procedure were used as zero reference angles for the dynamic trials. After the anatomical calibration procedure, anatomical landmarks were removed, and participants performed 5 successful walking trials across a 10-m walkway at a speed of 1.3 to 1.4 m/s. We monitored walking speed using a handheld digital timer and defined a *successful trial* as one in which walking speed was within the appropriate range and right-limb initial contact and toe-off occurred on the force platform. Because of marker dropout during some trials, we could not reconstruct 5 trials for all participants. Therefore, 3 trials were averaged for subsequent analysis. For participants with 5 complete trials, we selected the 3 trials with the least number of marker dropouts.

Statistical Analysis

We performed repeated-measures multivariate analyses of variance (MANOVAs) for each of the functional articulations during the loading response, midstance, terminal stance, and preswing subphases. The between-groups factor in the repeated-measures MANOVAs was insert (BALORT, FCORT), and the within-group factor was insert condition (no FO, FO). Dependent variables were plantar-flexion, dorsiflexion, inversion, eversion, abduction, and adduction displacements within each subphase for each functional articulation. We computed displacement in each direction within a plane (ie, plantar-flexion and dorsiflexion displacements were computed in the sagittal plane). Follow-up repeated-measures analyses of variance (ANOVAs) were performed to investigate repeated-measures MANOVA omnibus F ratios that were different. The α level for all analyses was set at .05 (version 15.0; SPSS Inc, Chicago, IL). In addition, we computed partial η^2 to facilitate interpretation of the clinical meaningfulness of the results. The partial η^2 was interpreted based on recommendations by Cohen[20] for small (0.01), medium (0.06), and large (0.14) effects. We analyzed gait kinematics within the subphases of stance using definitions of loading response, midstance, terminal stance, and preswing established by Perry and Burnfield.[17]

RESULTS

Loading Response

Repeated-measures MANOVA results did not reveal insert between-groups main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}=1.25$, $P=.36$, Wilks $\Lambda=0.55$), calcaneonavicular complex ($F_{6,9}=0.86$, $P=.56$, Wilks $\Lambda=0.64$), medial forefoot ($F_{6,9}=0.16$, $P=.98$, Wilks $\Lambda=0.90$), or 1MTP ($F_{6,9}=0.93$, $P=.52$, Wilks $\Lambda=0.62$). We also did not find insert condition within-group main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}=0.67$, $P=.68$, Wilks $\Lambda=0.69$), calcaneonavicular complex ($F_{6,9}=1.30$, $P=.35$, Wilks $\Lambda=0.54$), medial forefoot ($F_{6,9}=0.98$, $P=.49$, Wilks $\Lambda=0.61$), or 1MTP ($F_{6,9}=0.89$, $P=.54$, Wilks $\Lambda=0.63$). Finally, we did not find insert-by-insert condition interactions for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}=1.34$, $P=.33$, Wilks $\Lambda=0.53$), calcaneonavicular complex ($F_{6,9}=1.13$, $P=.42$, Wilks $\Lambda=0.57$), medial forefoot ($F_{6,9}=0.57$, $P=.75$, Wilks $\Lambda=0.72$), or 1MTP ($F_{6,9}=1.50$, $P=.28$, Wilks $\Lambda=0.50$).

Midstance

Midstance repeated-measures MANOVA results revealed an insert condition within-group main effect for rearfoot complex displacement ($F_{6,9}=4.71$, $P=.02$, Wilks $\Lambda=0.24$). Follow-up repeated-measures ANOVA analysis revealed that dorsiflexion displacement was greater during the FO than the no-FO condition ($F_{1,14}$=5.24, P=.04, partial η^2=.27) (Table 2). Participants entered midstance in a plantar-flexed position with the rearfoot complex dorsiflexing during both the no-FO and FO conditions. Upon entering midstance, participants continued to dorsiflex in a similar pattern through the entire subphase during both insert conditions (Figure 3). Although the patterns of motion were similar, participants entered midstance in a greater plantar-flexed position and transitioned from a plantar-flexed to a dorsiflexed position later during the no-FO condition (approximately 40% stance) than the FO condition (approximately 30% stance) (Figure 3). We found no insert between-groups main effects for the functional articulations of the rearfoot complex ($F_{6,9}$=0.89, P=.54, Wilks Λ=0.63), calcaneonavicular complex ($F_{6,9}$=0.61, P=.72, Wilks Λ=0.71), medial forefoot ($F_{6,9}$=1.52, P=.27, Wilks Λ=0.50), or 1MTP ($F_{6,9}$=1.63, P=.25, Wilks Λ=0.48). In addition, we found no insert condition within-group main effects for the functional articulations of the calcaneonavicular complex ($F_{6,9}$=1.05, P=.45, Wilks Λ=0.59), medial forefoot ($F_{6,9}$=2.03, P=.16, Wilks Λ=0.43), or 1MTP ($F_{6,9}$=3.04, P=.07, Wilks Λ=0.33). Finally, we did not find insert-by-insert condition interactions for the functional articulations of the rearfoot complex ($F_{6,9}$=2.60, P=.1, Wilks Λ=0.37), calcaneonavicular complex ($F_{6,9}$=1.67, P=.24, Wilks Λ=0.47), medial forefoot ($F_{6,9}$=1.67, P=.24, Wilks Λ=0.47), and 1MTP ($F_{6,9}$=1.08, P=.44, Wilks Λ=0.58).

Terminal Stance

Repeated-measures MANOVA results for terminal stance revealed insert-by-insert condition interactions for 1MTP ($F_{6,9}$=6.34, P=.007, Wilks Λ=0.19) and rearfoot complex ($F_{6,9}$=3.75, P=.04, Wilks Λ=0.29) displacement. Follow-up repeated-measures ANOVA analysis revealed an insert-by-insert condition interaction for abduction displacement of the 1MTP ($F_{1,14}$=7.87, P=.01, partial η^2=.36). However, additional follow-up analyses did not reveal differences between the no-FO and FO conditions for the BALORT ($F_{1,7}$=4.32, P=.08, partial η^2=.38) or FCORT ($F_{1,7}$=4.10, P=.08, partial η^2=.37) (Table 2).

Follow-up repeated-measures ANOVA revealed an insert-by-insert condition interaction for eversion displacement of the rearfoot ($F_{1,14}$=6.64, P=.02, partial η^2=.67). However, additional follow-up analysis did not reveal differences between the no-FO and FO conditions for the BALORT ($F_{1,7}$=3.31, P=.11, partial η^2=.32) or FCORT ($F_{1,7}$=4.51, P=.07, partial η^2=.39) (Table 2).

We found an insert condition within-group main effect for the 1MTP ($F_{6,9}$=6.27, P=.008, Wilks Λ=0.19), but the difference was not investigated further because of the insert-by-insert condition interaction. We did not find insert between-groups main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}$=1.75, P=.21, Wilks Λ=0.46), calcaneonavicular complex ($F_{6,9}$=0.77, P=.61, Wilks Λ=0.66), medial forefoot ($F_{6,9}$=0.67, P=.68, Wilks Λ=0.69), or 1MTP ($F_{6,9}$=1.53, P=.27, Wilks Λ=0.50). We found no insert-condition within-group main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}$=2.81, P=.08, Wilks Λ=0.35), calcaneonavicular complex ($F_{6,9}$=3.17, P=.06, Wilks Λ=0.32), or medial forefoot ($F_{6,9}$=3.14, P=.06, Wilks Λ=0.32). Finally, we did not find insert-by-insert condition interactions for the variables within the functional articulations of the calcaneonavicular complex ($F_{6,9}$=2.73, P=.09, Wilks Λ=0.36) or medial forefoot ($F_{6,9}$=3.29, P=.053, Wilks Λ=0.31).

Preswing

Repeated-measures MANOVA results did not reveal insert between-groups main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}$=2.12, P=.15, Wilks Λ=0.41), calcaneonavicular complex ($F_{6,9}$=0.17, P=.98, Wilks Λ=0.90), medial forefoot ($F_{6,9}$=0.37, P=.88, Wilks Λ=0.80), or 1MTP ($F_{6,9}$=0.62, P=.71, Wilks Λ=0.71). We also found no insert condition within-group main effects for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}$=1.68, P=.23, Wilks Λ=0.47), calcaneonavicular complex ($F_{6,9}$=1.99, P=.17, Wilks Λ=0.43), medial forefoot ($F_{6,9}$=1.42, P=.31, Wilks Λ=0.51), or 1MTP ($F_{6,9}$=1.25, P=.37, Wilks Λ=0.55). Finally, we did not find insert-by-insert condition interactions for any of the variables within the functional articulations of the rearfoot complex ($F_{6,9}$=0.46, P=.82, Wilks Λ=0.77), calcaneonavicular complex ($F_{6,9}$=0.56, P=.75, Wilks Λ=0.73), medial forefoot ($F_{6,9}$=0.23, P=.96, Wilks Λ=0.87), or 1MTP ($F_{6,9}$=1.83, P=.20, Wilks Λ=0.45).

DISCUSSION

We hypothesized that 2 weeks of custom-molded FO intervention would increase calcaneonavicular complex abduction displacement during midstance and would decrease rearfoot complex inversion and increase rearfoot complex eversion dur-

Table 2. Functional Articulation Excursion (°) During No Foot-Orthosis and Foot-Orthosis Conditions (Mean ± SD [95% CI])

	Condition			
Stance	No–Foot Orthosis		Foot Orthosis	
Midstance				
Rearfoot complex dorsiflexion	9.22 ± 2.20 (8.01, 10.43)[a]		10.10 ± 3.00 (8.49, 11.71)[a]	
	No Balanced Foot Orthosis	No Full-Contact Foot Orthosis	Balanced Foot Orthosis	Full-Contact Foot Orthosis
Terminal stance				
Rearfoot complex eversion	0.02 ± 0.04 (0.43, 0.46)[b]	0.85 ± 0.83 (0.41, 1.30)[b]	0.18 ± 0.26 (0.12, 0.47)[b]	0.28 ± 0.48 (0.01, 0.57)[b]
First metatarsophalangeal complex abduction	0.32 ± 0.34 (0.04, 0.68)[b]	0.78 ± 0.58 (0.41, 1.14)[b]	0.46 ± 0.45 (0.07, 0.85)[b]	0.54 ± 0.58 (0.15, 0.93)[b]

[a] Indicates difference was found with follow-up analysis of variance of omnibus insert main effect F ratio ($P \leq .05$).

[b] Indicates no difference was found with follow-up analysis of variance of omnibus insert–insert condition F ratio ($P > .05$).

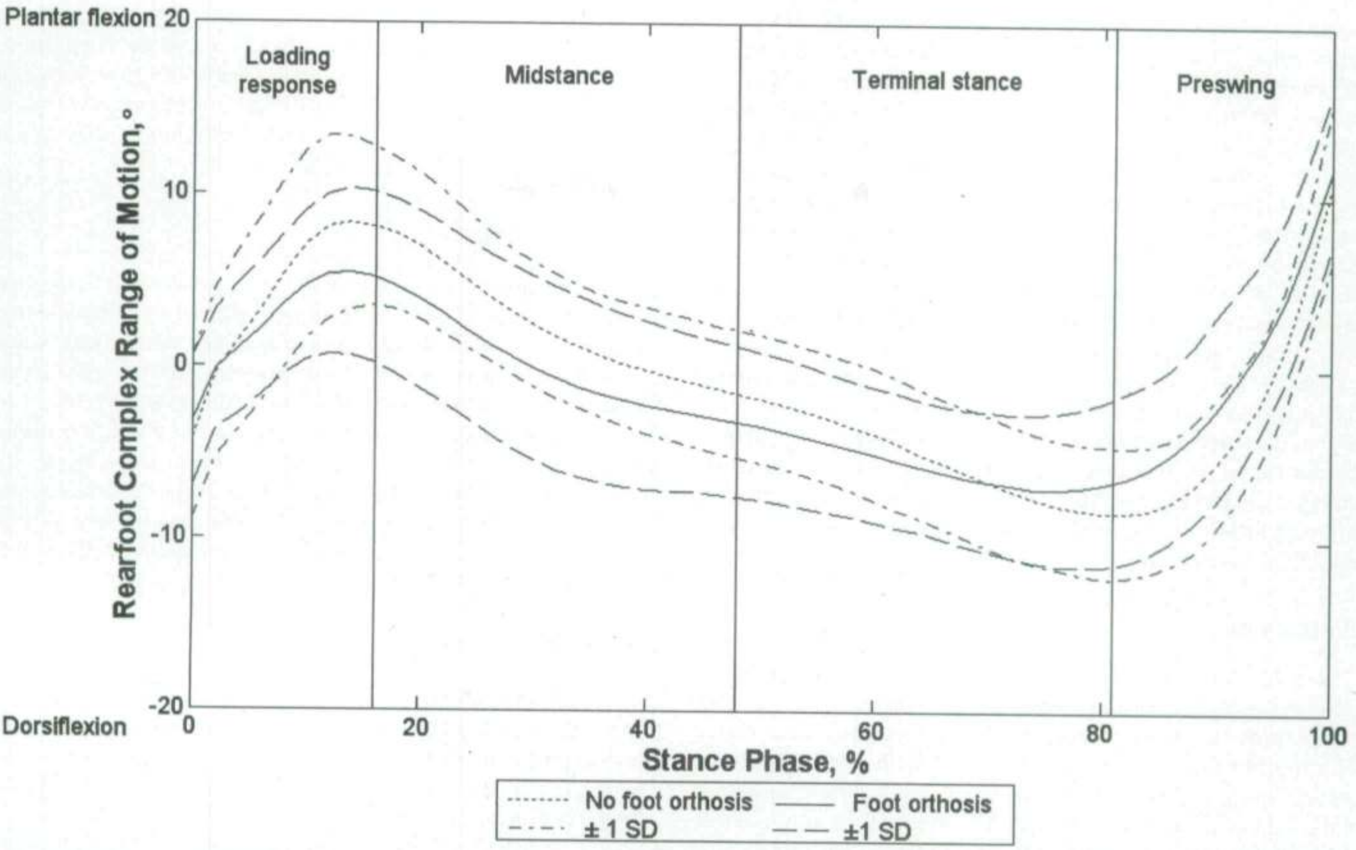

Figure 3. Sagittal-plane, rearfoot complex, and stance-phase kinematics for no–foot-orthosis and foot-orthosis conditions (mean ± 1 SD). Vertical lines represent the partition points for the loading response, midstance, terminal stance, and preswing subphases.

ing preswing. Although FO intervention did affect walking gait kinematics, the effects were not those hypothesized. No significant difference in calcaneonavicular complex abduction displacement occurred during midstance; in fact, displacement was less during the FO than during the no-FO condition. With respect to the rearfoot complex, inversion displacement increased and eversion displacement decreased during preswing in the FO versus the no-FO condition, but the changes were not significantly different.

The differences that did occur between the FO and no-FO conditions were in the sagittal plane of the rearfoot complex during midstance. Specifically, rearfoot complex dorsiflexion displacement was greater in the FO condition (10.21° ± 2.9°) than in the no-FO condition (9.1° ± 2.2°). Although the average absolute increase in dorsiflexion displacement in the FO condition was small (1.1°), the total dorsiflexion displacement in the no-FO condition was approximately 9°. Therefore, the relative increase (approximately 12%) might represent a clinically relevant change. Furthermore, the increase in dorsiflexion displacement in conjunction with observed decreased position of plantar flexion at the beginning of midstance and an earlier transition from a plantar-flexed to dorsiflexed position in the FO versus no-FO condition might represent a correction in gait mechanics (Figure 3). These observed kinematic changes in the FO condition resulted in a pattern very similar to that previously reported in participants with typical foot posture.[8] In addition, although Cobb et al[8] did not report a difference, midstance dorsiflexion displacement was less in participants with low-mobile foot posture than in those with typical foot postures.

The sagittal-plane effect associated with orthosis intervention during midstance is inconsistent with the only other 3-dimensional walking study in which researchers investigated sagittal-plane kinematics. Eng and Pierrynowski[2] did not reveal differences associated with FO intervention in participants with "excessive" forefoot varus or calcaneal valgus. The inconsistency between the studies might result from different methods of foot posture quantification, foot models, or variable definitions. The investigators might not have been comparing the same abnormal foot postures because of the different methods of foot posture quantification. With respect to the foot models, Eng and Pierrynowski[2] modeled the entire foot as a single, rigid segment, whereas we defined the *rearfoot complex* as the functional articulation between the calcaneus and leg. Modeling the entire foot as a single, rigid segment might have masked differences occurring at the rearfoot complex. Finally, Eng and Pierrynowski[2] computed midstance displacement as the total sagittal-plane motion in the subphase, but we computed displacement in each direction (plantar flexion and dorsiflexion) within midstance. A potential disadvantage of using the total sagittal-plane displacement is that the same value could be recorded if plantar flexion increased and dorsiflexion decreased in one condition and plantar flexion decreased but dorsiflexion increased in the other condition.

Although repeated-measures MANOVA results revealed differences in rearfoot complex and 1MTP kinematics during terminal stance, follow-up analyses of 1MTP abduction and rearfoot complex eversion did not reveal differences between the no-FO and FO conditions. Our rearfoot complex results are inconsistent with those of previous 3-dimensional walking gait studies. Davis et al[1] reported less eversion excursion computed over the entire stance phase during semicustom FO versus custom-molded FO (mean excursion = 0.9°) and no-FO (mean excursion = 1.6°) conditions. Similarly, Zifchock and Davis[3] reported that eversion displacement (maximum eversion – heel contact position) was less in custom-molded orthosis (mean decrease = 1°) and semicustom orthosis (mean decrease = 1°) conditions than a no-orthosis condition in high-arched and low-arched participants. Finally, Eng and Pierrynowski[2] reported less frontal-plane foot displacement (mean decrease = 1.8°) during midstance with FO intervention. Differences in the methods of foot-posture quantification, foot models, or variable definitions again might have contributed to the different results between our study and previous investigations.

Finally, the omnibus insert-by-insert condition interactions also suggested that the BALORT and FCORT had different effects on rearfoot complex and 1MTP walking gait kinematics during terminal stance. As stated, however, the follow-up analyses did not reveal differences, suggesting that the effect of different orthosis designs on walking gait kinematics warrants additional investigation.

Limitations

Before conclusions are drawn about the effect of FO intervention for people with low-mobile foot posture, the limitations of our study should be considered. First, the changes associated with the FO intervention in participants with low-mobile foot posture were assumed to be corrective because of the similarity in the kinematic patterns between the low-mobile foot posture group in the orthosis condition in our study and previously collected data from participants with typical foot posture. However, because the participants with low-mobile foot posture in our study were asymptomatic, we could not determine whether long-term use of the orthosis would prevent or potentially contribute to the development of lower extremity injury. To further elucidate the effect of the mechanical changes associated with orthosis intervention, orthosis intervention in participants with abnormal foot posture and symptomatic lower extremity pathologic conditions or long-term orthosis intervention in asymptomatic participants with abnormal foot posture should be studied.

A second potential limitation to consider was the performance of the no-FO condition trials after the 2-week break-in period. We assumed that wearing the FOs during the break-in period would not affect gait kinematics during the no-FO condition. Although we believe the assumption is reasonable, a future study comparing no-FO condition trials before and after a period of FO intervention might be warranted.

Several other factors also deserve consideration in the planning and conduction of future studies in which the effects of FO intervention on gait kinematics are investigated. First, although we quantified foot structure and mobility using a method with moderate to high intratester and intertester reliabilities, we do not know where gait kinematics might change along the continuum of structure or mobility. Furthermore, we do not know whether foot structure, mobility, or potentially a combination thereof is related more strongly to gait function. To answer these questions, researchers need to investigate the relationship between the foot-posture measures and multisegment foot-model kinematics. Second, the influence of other factors, such as the strength of the lower extremity musculature acting as dynamic stabilizers, warrants additional investigation. Third, the effect of foot posture might become more apparent and important in situations when the lower extremity dynamic stabilizers are compromised (ie, fatigued). Fourth, although our approach partitioned stance into subphases, the statistical model continues to rely on few discrete variables to represent gait function. Future researchers also should investigate alternative approaches, such as dynamic system techniques, that might better capture the continuous nature of gait.

CONCLUSIONS

Two weeks of custom-molded FO intervention affected multisegment medial foot walking kinematics. Specifically, rearfoot complex dorsiflexion displacement during midstance was increased after orthosis intervention. Although the absolute change in dorsiflexion was small, the change relative to the subphase displacement during the no-FO condition might represent a clinically relevant difference. Of potentially greater clinical relevance might be the correction of the gait kinematic pattern during the custom-molded FO condition compared with that of previously collected data from participants with typical foot structure and mobility.[8]

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