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The effect of low-mobile foot posture on multi-segment medial foot model gait kinematics

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ABSTRACT

A number of in vitro, invasive in vivo, and non-invasive marker based multi-segment foot models (MSFMs) have reported significant motion in the articulations distal to the calcaneus during gait. Few studies, however, have applied a MSFM to the investigation of the effect of foot posture on gait kinematics. Differences in stance phase kinematics between participants with low-mobile (LMF) ($n = 11$) versus "typical" (TYPF) ($n = 11$) foot postures were investigated using a multi-segment medial foot model. Three-dimensional position and stance phase excursions of four functional articulations (rearfoot complex [RC], calcaneonavicular complex [CNC], medial forefoot, first metatarsophalangeal complex) were quantified using an eight optical camera motion analysis system (Vicon Motus, Vicon Motions Systems, Centennial, CO) and a custom written software program (Matlab 7.0.1, The MathWorks, Natick, MA), respectively. Excursions during four subphases of stance phase (loading response, midstance, terminal stance, pre-swing) at each of the functional articulations were compared using multivariate analyses of variance ($\alpha \leq 0.05$). Results revealed significantly decreased LMF group CNC abduction excursion ($p = 0.047$) during midstance. During pre-swing, LMF group RC inversion excursion was significantly increased ($p = 0.032$) and eversion excursion was significantly decreased ($p = 0.003$) compared to the TYPF group. When these differences are considered in conjunction with the kinematic patterns of other foot/leg segments and functional articulations, the changes may suggest dysfunction of normal leg-calcaneus coupling and the constrained tarsal mechanism associated with low-mobile foot postures.

1. Introduction

The relationship between foot posture, abnormal gait mechanics, and increased risk of lower extremity injury is widely accepted by clinicians. Results of prospective and retrospective studies investigating the relationship between foot structure and lower extremity injury risk, however, have been inconsistent [1–3]. To further elucidate the potential link between foot posture and gait, rearfoot complex models have been utilized to investigate the effect of foot posture on running kinematics [4,5]. Although the models have improved the understanding of the effect of foot posture on gait kinematics, they ignore interactions of the joints distal to the calcaneus. In vitro stereophotogrammetric [6,7], in vivo roentgen stereophotogrammetric [8], and invasive in vivo kinematic [9] studies, however, have reported significant contributions to foot motion from the joints distal to the calcaneus.

Recently, a number of surface marker based multi-segment foot models (MSFMs) have been developed [10–13]. Subsequent studies have reported significant kinematic differences between participants with "normal" foot posture and those with clinical pathologies [14–16], suggesting that MSFMs have the ability to differentiate between "normal" and "abnormal" gait.

Two studies have investigated the effect of foot posture on MSFM gait kinematics. Hunt and Smith [17] utilized a two-segment model to investigate stance phase kinematics in pronated/planus participants versus participants with "no obvious malalignment". The planus group demonstrated significantly decreased forefoot

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adduction at toe-off, decreased forefoot transverse plane range of motion (ROM), and increased rearfoot plantar flexion at 21% of stance. Using a three-segment model, Houck et al. [18] reported significantly increased rearfoot eversion at 28% of stance and greater initial contact and peak forefoot dorsiflexion in abnormally pronated versus normally pronated participants.

Although these studies support the role of MSFMs in advancing the understanding of the effect of foot posture on dynamic function, the inter-tester reliability of the classification systems used to quantify foot posture may somewhat limit the clinical relevance of the studies. While the measure(s) utilized by Hunt and Smith [17] to classify planus foot posture are unclear, the inter-tester reliability of visual observation utilized to classify the normal foot posture has been reported as poor [19]. The measures utilized by Houck et al. [18] have reported moderate-high intra-tester reliability, but low inter-tester reliability [20].

Furthermore, the studies may not have partitioned the foot into the most clinically relevant segments. Neither included a midfoot segment, and Hunt and Smith [17] classified the entire forefoot as a single rigid segment. Results of in vitro and invasive in vivo kinematic studies, however, suggest that significant movement occurs between the navicular and first ray [6,21,22], and that the medial and lateral forefoot function somewhat independently [21,22]. The purpose of this study, therefore, was to investigate the effect of foot posture on walking gait kinematics using a novel multi-segment medial foot model and a foot posture classification system with high intra- and inter-tester reliability. It was hypothesized that kinematics in the articulations distal to the calcaneus would differ significantly between participants with low-mobile versus "typical" foot postures.

2. Materials and methods

2.1. Participants

Following an initial screening for current musculoskeletal injuries, potential participants were further screened for eligibility through arch height and foot mobility assessment using the arch ratio in 90% weightbearing and the relative arch deformity ratio, respectively [23]. Eleven participants ($m = 4$; $f = 7$) with low-mobile foot posture (LMF) and 11 individuals ($m = 8$; $f = 3$) with "typical" arch height and foot mobility (TYPF) were enrolled in the study (Table 1). Prior to testing, all participants provided written informed consent in accordance with institutional guidelines, and the equipment and procedures of the study were explained.

2.2. Three-dimensional motion analysis

Eight optical cameras (Vicon Motus, Vicon Motions Systems, Centennial, CO) sampling at 120 Hz were used to capture three-dimensional coordinate data from marker clusters of 3–4 retroreflective markers (8 mm diameter) located on the leg and foot segments of interest. The markers were either placed directly on the skin or mounted on wands constructed from 1.8 mm wire and fixed to the skin. An AMTI force platform (Advanced Mechanical Technology, Newton, MA) sampling at 960 Hz was used to determine initial contact and toe-off events. Peak Performance Motus software (Version 8.0) was used to synchronize ground reaction force and coordinate data, convert analog signals to digital signals, and filter the coordinate data with a Butterworth filter using optimal cut-off frequencies determined via residual analysis (range: 2–5 Hz). Custom written software (Matlab 7.0.1, The MathWorks, Natick, MA) was used to perform rigid body transformation procedures using the calibrated anatomical system technique with a single value decomposition position and orientation estimator [24]. Clinically relevant joint angles between adjacent segments were then computed using the joint coordinate system (JCS) technique [25]. Positive sagittal, frontal, and transverse plane rotations were 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 [26]. Trials for each participant were then normalized to 100% of stance and ensemble averaged at 2% intervals. Finally, three-dimensional excursions (computed as the absolute difference between successive time frames in the appropriate rotation direction) within four subphases of stance [loading response (0–16%), midstance (16–48%), terminal stance (48–81%), pre-swing (81–100%)] were computed [27].

Table 1
Mean (SD) participant descriptive data.

	Typical foot posture ($n = 11$)	Low-mobile foot posture ($n = 11$)
Age (years)	25.2 (3.2)	24.5 (6.1)
Height (cm)	176.9 (12.1)	172.3 (10.5)
Mass (kg)	84.8 (22.1)	72.3 (15.2)
Arch ratio[a]	0.325 (0.01)	0.272 (0.01)
Relative arch deformity ratio ($10^4 \times N^{-1}$)[b]	0.637 (0.195)	1.158 (0.18)

[a] "Typical" arch structure (arch ratio = 0.301–0.343) was defined as arch ratios ranging between −0.5 and 1 SD of the mean arch ratio assessed from 51 random volunteers (102 feet). Low arch structure (arch ratio ≤ 0.287) was defined as an arch ratio of ≥1 SD below the mean.

[b] For the relative arch deformity(RAD) ratio, a larger ratio is associated with a more mobile foot. For participants with "Typical" arch structure, "Typical" foot mobility (RAD ratio = $(0.378–1.053) \times 10^4 \times N^{-1}$) was defined as RAD ratios ranging between and -1 and 0.5 SD of the mean assessed from the same 102 feet. For participants with a low arch structure, a mobile foot (RAD ratio $\geq 0.828 \times 10^4 \times N^{-1}$) was defined as a RAD ratio greater than the mean ratio of the 51 random volunteers.

2.3. Foot segmentation

Foot segmentation was based on data from in vitro studies [6], in vivo roentgen stereophotogrammetric studies [6,8], and the constrained tarsal mechanism [28] and forefoot twist [29] concepts.

2.3.1. Rearfoot complex

Cartesian coordinate systems defined within the leg and calcaneus segments comprised the rearfoot complex (RC) (Fig. 1). The JCS used to compute sagittal and frontal plane RC motions were 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, a separate JCS was constructed 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 was then computed about the vertical axis of the leg [26] (Appendix B).

2.3.2. Calcaneonavicular complex

Cartesian coordinate systems defined within the calcaneus and the navicular segments formed the calcaneonavicular complex (CNC) (Fig. 1). The JCS used to compute three-dimensional CNC 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 (Appendix B).

2.3.3. Medial forefoot

The medial forefoot was formed by Cartesian coordinate systems defined within the medial two rays [29] and the navicular segment (Fig. 1). The JCS used to compute three-dimensional medial forefoot motion was formed by the mediolateral axis of the navicular segment, the anteroposterior axis of the medial rays segment, and a floating axis computed as the cross-product of the medial rays anteroposterior and navicular mediolateral axes (Appendix B).

2.3.4. First metatarsophalangeal complex

The 1st metatarsophalangeal complex (1MTP) was formed by Cartesian coordinate systems defined within the hallux and medial rays segments (Fig. 1). The JCS used to compute three-dimensional motions of the 1MTP was formed by the mediolateral axis of the medial rays 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 (Appendix B).

2.4. Procedures

Prior to data collection, dynamic camera calibration (0.5 m (H) × 0.4 m (W) × 0.9 m (L) volume) was performed. Technical marker clusters and anatomical landmarks were then applied to each segment on the subject's right foot and leg and an anatomical calibration procedure was performed (Fig. 1). During the anatomical calibration procedure, the participant was in a seated position with the leg oriented vertically and the midpoint of the calcaneus and second metatarsal aligned parallel to the direction of progression. Segmental angles computed during the anatomical calibration procedure were used as zero reference angles for the dynamic trials. Following the anatomical calibration procedure, anatomical landmarks were removed and participants performed five successful walking trials across a 10 m walkway at a speed of 1.3–1.4 m s^{-1}. Participants wore the same style sandal (Merrell

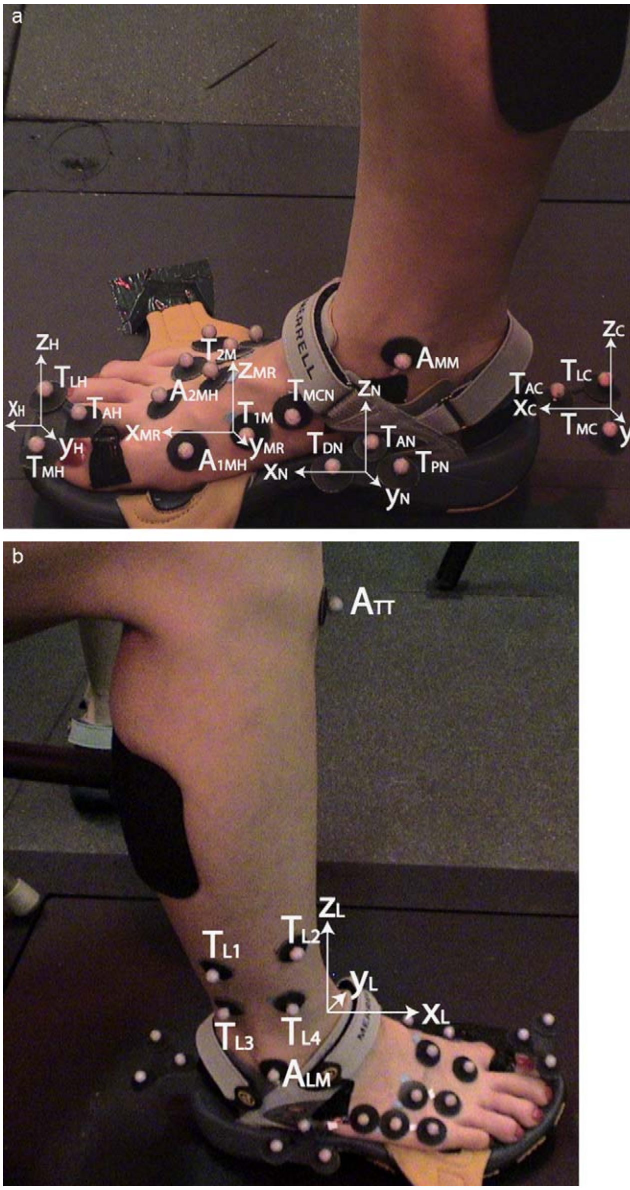

Fig. 1. (a) Calcaneus (medial technical marker [T_{MC}], lateral technical marker [T_{LC}], apex technical marker [T_{AC}]), navicular (proximal technical marker [T_{PN}], distal technical marker [T_{DN}], apex technical marker [T_{AN}]), medial rays (medial cuneiform technical marker [T_{MCN}], 1st metatarsal technical marker [T_{1M}], 2nd metatarsal technical marker [T_{2M}], 1st metatarsal head anatomical marker [A_{1MH}], 2nd 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. All marker wands were positioned parallel to the floor with the markers utilized to compute the x (anteroposterior) axis aligned parallel to the line of progression. (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}], medial malleolus anatomical marker [A_{MM}] (see a), lateral malleolus anatomical marker [A_{LM}], tibial tuberosity [A_{TT}] anatomical marker) 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) due to difficulties with reconstruction of the lateral segment marker clusters, however, only the medial segments are presented.

Waterfall, Wolverine World Wide, Inc. Rockford, MI) and walking speed was monitored using a hand held digital timer. A successful trial was defined as one in which right limb initial contact and toe-off occurred on the force platform and walking speed was within the appropriate range. Due to marker drop-out during some trials, five trials could not be reconstructed for all participants. As a result, three trials were averaged for subsequent analysis. For participants with five complete trials, the three trials with the least number of marker drop-outs were chosen.

2.5. Statistical analysis

Multivariate analyses of variance (MANOVA) for each of the functional articulations during the loading response, midstance, terminal stance, and pre-swing subphases were performed with a level of significance established at $\alpha = 0.05$ (SPSS v 15.0, Chicago, IL). The between group factor in the MANOVAs was foot posture, and the dependent variables were plantar flexion, dorsiflexion, inversion, eversion, abduction, and adduction excursion within each subphase for each functional articulation. Due to the novelty of the MSFM, within-session coefficients of multiple correlation were computed across the TYPF groups' three trials to examine model reliability. Coefficients of multiple correlation >0.70 were considered very repeatable [30].

3. Results

3.1. System accuracy and model reliability

The average calibration wand standard deviation and camera calibration residual over the data collection period were 0.42 and 0.80 mm, respectively. With respect to MSFM reliability, sagittal, frontal, and transverse plane coefficients of multiple correlation were >0.83 for all of the functional articulations (Appendix B).

3.2. Loading response and terminal stance

MANOVA results did not reveal significant group differences for any of the variables within the functional articulations.

3.3. Midstance

Midstance MANOVA results revealed a significant group difference for CNC excursion ($F_{6,15} = 3.80$, $p = 0.017$). Follow-up univariate ANOVA analysis revealed significantly decreased abduction excursion ($F_{1,20} = 4.49$, $p = 0.047$) in the LMF (mean: $-0.9 \pm 0.7^{\circ}$) vs. TYPF (mean: $-1.8 \pm 1.1^{\circ}$) group. The TYPF group entered midstance in an abducted position, abducted until ≈30% of stance, leveled off until ≈40%, and then gradually adducted through the remainder of the subphase. The LMF group entered midstance in a similar position of abduction, but then adducted through the majority of midstance (Fig. 2).

3.4. Pre-swing

Pre-swing MANOVA results revealed a significant group difference for RC excursion ($F_{6,15} = 5.23$, $p = 0.004$). Follow-up univariate ANOVA analysis revealed significant inversion ($F_{1,20} = 5.31$, $p = 0.032$) and eversion ($F_{1,20} = 11.04$, $p = 0.003$)

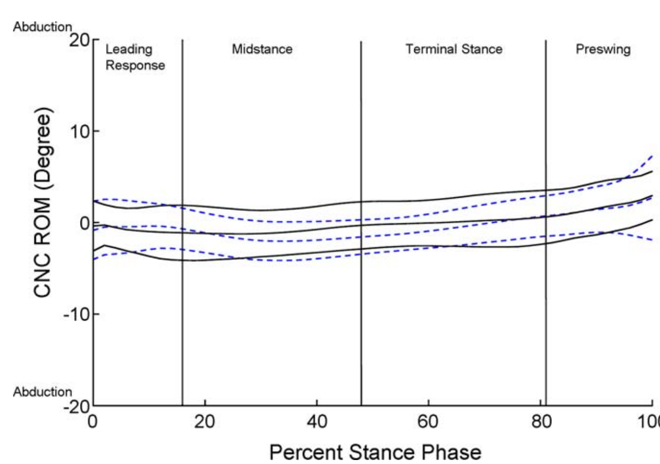

Fig. 2. TYPF (dotted lines) and LMF (solid lines) transverse plane calcaneonavicular complex stance phase kinematics (mean ± 1 SD). Vertical lines represent the partition points for the loading response, midstance, terminal stance, and pre-swing subphases.

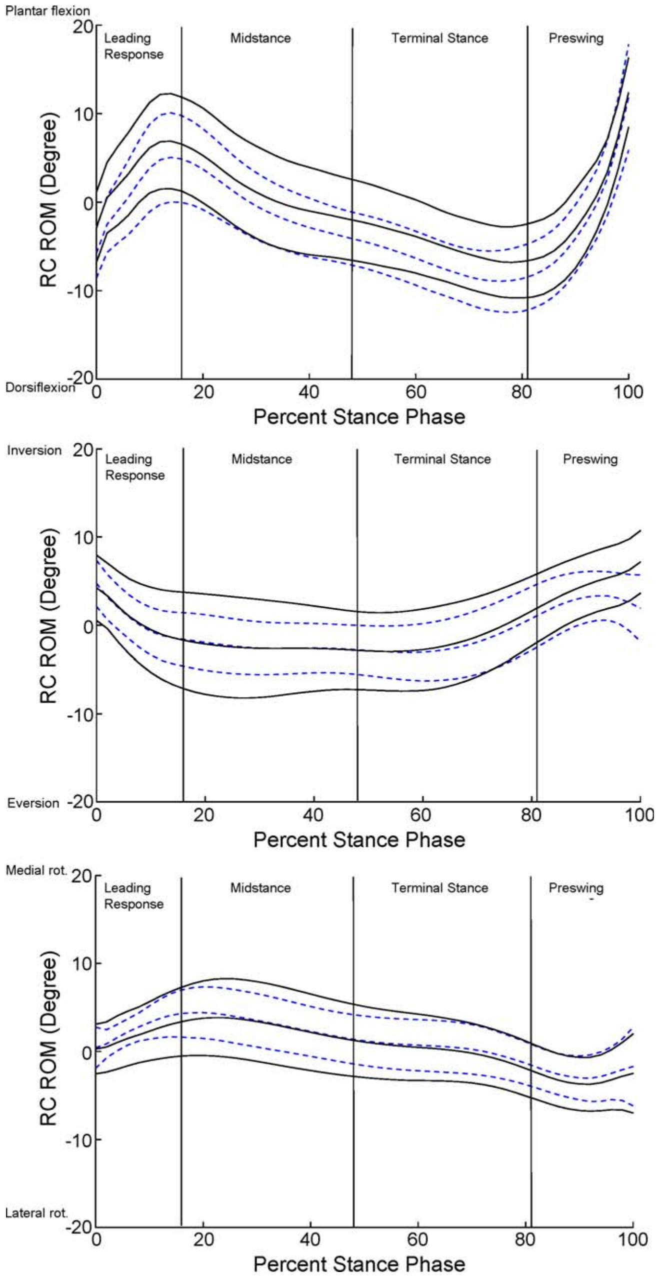

Fig. 3. TYPF (dotted lines) and LMF (solid lines) sagittal, frontal, and transverse plane rearfoot complex stance phase kinematics (mean ± 1 SD). Vertical lines represent the partition points for the loading response, midstance, terminal stance, and pre-swing subphases.

excursion group effects. The LMF group exhibited significantly increased RC inversion excursion (LMF: $5.8 \pm 3.1^\circ$, TYPF: $3.1 \pm 2.3^\circ$) and significantly decreased eversion excursion (LMF: $-0.2 \pm 0.4^\circ$, TYPF: $-2 \pm 1.7^\circ$) during the subphase. Both groups entered pre-swing with similar angles of inversion and both continued to invert at a similar rate until ≈87% of the subphase. The rate of TYPF group inversion then began to decrease until approximately 94% of stance, at which time the RC began to evert through the remainder of the subphase. The LMF group, however, continued to invert throughout the remainder of pre-swing (Fig. 3).

4. Discussion

Differences in foot segmentation, foot posture classification, walking speed, footwear, and discrete variables investigated make direct comparisons of the current study results with the previous MSFM studies investigating the effect of static foot posture on walking kinematics [17,18] difficult. Differing foot segmentation methods may not be expected to yield the same kinematic patterns and because the foot classification methods differed, the studies may not have been comparing the same foot postures. With respect to walking speed, the current study controlled speed (range: 1.3–1.4 m s^{-1}) based on the average speed for the age group being investigated [31], whereas participants in the Hunt and Smith [17] (range: 1.25–1.9 m s^{-1}) and Houck et al. [18] studies walked at self-selected paces. Because the purpose of the study was to compare kinematics between individuals, the fixed range was chosen to minimize kinematic differences due to walking speed. Finally, Hunt and Smith [17] and Houck et al. [18] investigated peak angles at specific instances within stance and total stance ROM, whereas the current study compared excursions within stance subphases. Peak angles were not utilized because group differences may be the result of marker placement or reference position/offset rather than true kinematic differences. Total stance ROM was not utilized because the single measure may mask between group differences within the subphases. Although total stance ROM was not utilized in the statistical model, it was computed for the TYPF group for comparison purposes and is presented in Table 2 along with the available data from Hunt and Smith [17], Houck et al. [18], and two recently published invasive in vivo MSFM studies [9,21]. Finally, participants in the current study wore sandals whereas participants in the previous studies walked barefoot. Because the effect of shoe wear on MSFM kinematics has not been investigated, the potential influence due to the footwear differences is unknown. Considering the previously mentioned methodological differences and also variability in the methods utilized to compute relative joint angles, kinematic differences between the current study and previous studies were relatively small (range: 0.1° and 7.6°) (Table 2).

Table 2
Mean (SD) stance phase ROM from relevant multi-segment foot models.

	RC			CNC			Medial forefoot		
	Sag	Fron	Tran	Sag	Fron	Tran	Sag	Fron	Tran
Cobb	21.5 (4.2)	9.1 (2.6)	9.0 (3.6)	9.7 (4.5)	12.9 (5.4)	6.5 (2.0)	14.5 (3.7)	12.8 (4.7)	6.3 (2.0)
Hunt [17]	22	8	10	NA[a]	NA[a]	NA[a]	12	5	10
Houck [18]	NR[b]	9.2 (1.6)	NR[b]	NA[a]	NA[a]	NA[a]	18.5 (4.2)	NR[b]	NR[b]
Nester [21][c]	13.9 (3.2)	8.6 (2.6)	6.1 (1.8)	6.1 (3.0)	9.5 (2.7)	11.3 (5.6)	11.6 (3.5)	11.0 (2.4)	5.4 (3.7)
Lundgren [9][d]	17.0 (2.1)	11.3 (3.5)	7.3 (2.4)	NA[a]	NA[a]	NA[a]	NA[a]	NA[a]	NA[a]

[a] NA is not applicable, the study did not define the functional articulation.
[b] NR is not reported, the study did not report the data for the plane.
[c] The kinematic data are computed from bone mounted markers. The RC and CNC functional articulations were defined using the same segments as those in the current study. The medial forefoot was defined as the articulation between the navicular and first metatarsal.
[d] The kinematic data are computed from bone mounted markers.

4.1. Rearfoot complex

The increased inversion excursion and decreased eversion excursion during pre-swing may suggest that the LMF group was not sufficiently inverted at 87% of stance to provide a stable base for push-off and therefore continued to invert through the remainder of the subphase. If this was the case, either increased eversion excursion earlier in stance, or similar excursion with initial contact in a less inverted position may also be expected in the LMF group. RC eversion excursion did not differ significantly between the groups during any of the earlier subphases and both groups made initial contact in a similar degree of inversion (Fig. 3). It is possible that the LMF group did make initial contact in a lesser degree of inversion, but that the difference was masked by the reference position. A semi-weightbearing reference position was chosen because in a weightbearing position, compensatory motions of the foot and leg have already occurred and as such, differences between the foot posture groups may be masked. It is still possible, however, that the LMF group was more everted in the reference position. If this was the case, using the reference angle as the offset may have shifted the entire stance curve vertically. This possibility may be further supported by the results of Houck et al. [18] which revealed significantly increased rearfoot eversion angles at 21% stance when using a subtalar neutral reference position, but not with a weightbearing reference position. Of potentially greater clinical relevance than the relatively small inversion (2.7°) and eversion (1.8°) excursion differences between the groups may be the frontal plane rearfoot and transverse plane leg coupling difference during pre-swing. Both groups demonstrated lateral leg rotation until $\approx$87% of stance followed by medial rotation for the remainder of the phase (Fig. 3). The TYPF group coupling of rearfoot eversion with leg medial rotation and vice versa supports the expected rearfoot-leg coupling pattern. The LMF group rearfoot-leg coupling pattern was also as expected until $\approx$87% of stance, when the rearfoot continued to invert but the leg began to medially rotate. This may suggest inappropriate leg-rearfoot coupling associated with the low-mobile foot posture during pre-swing, which others have suggested may be more important clinically than differences in the quantity of rearfoot or leg motion [4,26].

4.2. Calcaneonavicular complex

The significantly decreased LMF group CNC abduction excursion during midstance, which resulted in prolonged adduction, may be associated with prolonged lowering of the medial longitudinal arch. In addition, transverse plane motion of the CNC and of the calcaneus relative to the leg (the inverse of the leg relative to calcaneus graph) in the TYPF group is consistent with the constrained tarsal mechanism proposed by Huson [28]. The concept suggests the talus, calcaneus, navicular, cuboid, and their associated articulations act as a constrained mechanism (i.e. inversion of any one of the articulations within the constrained mechanism will cause inversion of the other articulations) (Figs. 2 and 3). The CNC adduction and RC abduction during early midstance exhibited by the LMF group, however, may suggest dysfunction of the constrained tarsal mechanism (Figs. 2 and 3). Once again, altered coupling may be of greater clinical relevance than the significant, but somewhat small (0.9°), CNC abduction excursion difference.

Although significant kinematic differences between the foot posture groups were revealed, additional research to further clarify the potential link between foot posture and dynamic function is warranted. First, the relationship between foot measures with moderate-high intra- and inter-tester reliability and MSFM gait kinematics requires additional study. In moving forward, utilization of the foot posture measure(s) most strongly associated with gait kinematics will be important. Second, alternative data analysis techniques such as continuous relative phase analysis, that may capture the continuous nature of gait and the joint coupling among multiple foot segments better than discrete variable analysis, requires exploration. Third, additional study is needed to determine the potential physiological relevance of disrupted coupling in the joints distal to the calcaneus. Finally, although foot posture may play a significant role in dynamic function, it is likely one of a composite of factors [26]. Therefore, studies investigating the relationship between foot posture and function using MSFMs may benefit from inclusion of other potentially important variables (i.e. lower extremity strength).

5. Conclusion

Statistically significant, but relatively small, differences in gait kinematic excursions exist in the joints distal to the calcaneus in participants with low-mobile foot posture versus participants with typical foot posture. Of greater clinical relevance may be the differing midstance and pre-swing stance subphases coupling patterns which may suggest constrained tarsal mechanism and leg-calcaneus coupling dysfunction associated with low-mobile foot posture. Furthermore, the results, along with those reported by Hunt and Smith [17] and Houck et al. [18], suggest that it will be important for future studies investigating the relationship between foot posture and dynamic function to address the joints distal to the calcaneus.

Acknowledgement

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Conflicts of interest statement

There are no conflicts of interest regarding this work and the authors.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.gaitpost.2009.06.005.

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