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ABSTRACT

The relationship between a static measure of dorsal first ray mobility and dynamic motion of the first ray, midfoot, and hindfoot during the stance phase of walking was investigated in healthy, asymptomatic subjects who represented the spectrum of static flexibility. Static first ray mobility of 15 subjects was measured by a load cell device and ranged from stiff (3.1 mm) to lax (8.0 mm). Using three-dimensional motion analysis, mean first ray dorsiflexion/eversion and mid-/hindfoot eversion peak motion, time-to-peak, and eversion excursion were evaluated. Subjects with greater static dorsal mobility of the first ray demonstrated significantly greater time-to-peak hindfoot eversion and eversion excursion ($p < .01$), and midfoot peak eversion and eversion excursion ($p < .01$). No significant association was found between static first ray mobility and first ray motion during gait. This research provides evidence that the dynamic response of the foot may modulate the consequences of first ray mobility and that compensory strategies are most effective when static measures of dorsal mobility are most extreme.

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INTRODUCTION

Excessive pronation of the subtalar joint is commonly cited as a factor in lower extremity injuries.[2–5,21] Increased pronation, a delay in the time-to-peak pronation, or an extended period of pronation may contribute to pathology by affecting the distribution of plantar contact pressure, or by affecting the movement of the foot into supination.[9,10,12,23]

Instability of the first ray, due to ligamentous laxity, is believed to be related to excessive pronation.[10,11,27,29] During weightbearing, this ligament laxity causes the first metatarsal to remain in dorsal extension.[19] Hindfoot eversion increases or pronation is prolonged until tension in the first metatarsal plantar ligaments stops/reverses the motion of the first ray.[22] A possible consequence of this hypermobility is that supination of the midtarsal and subtalar joints is delayed, decreasing the rigidity of the foot during terminal stance and adversely affecting push-off mechanics.[20,22] In addition, increased dorsal extension may result in reduced loading of the first ray with the load shifted to the lesser metatarsals.[22,26]

The ability of clinical measures of foot mobility to provide insight into the mechanics of the foot during dynamic activity has not been established. Pertinent to establishing such relationships is the ability to measure objectively the mobility of the first ray as well as documenting the movements of different segments of the foot during dynamic activity. Recent research has described a valid and reliable objective approach for measuring static first ray sagittal plane mobility.[16,17] Associated studies have shown static mobility of the first ray to range between 2 and 8 mm in healthy subjects.[13–15] In contrast, the ability to model the foot during dynamic activities has not completely overcome the inherent complexity of multiple systems and structures that interface during gait. Studies that measured foot segmental movement during walking have uncovered links between movement in the hindfoot, midfoot, and forefoot during the stance phase. Cornwall and McPoil[7,8] reported the navicular follows a similar frontal plane pattern of motion as the calcaneus during walking, but with maximum eversion occurring somewhat later in stance. The major component of motion for the first metatarsal was found to be in the sagittal plane, substantiating the potential of static measures of dorsal mobility to help predict the dynamic response of the foot.[6,8] However, the link between first ray motion and overall foot mechanics remains unclear.

PURPOSE AND HYPOTHESIS

The purpose of this study was to examine the relationship between the static dorsal mobility of the first ray and dynamic motion of the first ray, midfoot, and hindfoot during the stance phase of walking. The null hypothesis tested was that the gait variables (peak motion, total excursion, and time-to-peak motion) of dynamic first ray dorsiflexion and eversion, midfoot eversion, and hindfoot eversion would not differ between subjects having varied amounts of static first ray mobility

METHODS

Subjects

Fifteen subjects (six males, nine females) were selected from a pool of 20 healthy, asymptomatic volunteers. Exclusion criteria included lower extremity pain or pathology that would alter gait mechanics. Static device measures of first ray mobility were used to screen volunteers. By design, the subjects selected had first ray mobility that ranged from stiff to lax. Subjects ranged in age from 18 to 44 years (mean, 28.9). Height (mean, 171 cm) and weight (mean, 73 kg) of subjects were within normal range. The Human Subjects Committee from the University of Iowa approved the study and informed written consent was obtained.

Instrumentation

Static first ray mobility was measured using a load cell device described elsewhere.[13–17] A controlled, dorsally directed standard force of 55 N was imposed on the plantar aspect of the head of the first metatarsal. Dorsal displacement of the first ray was measured by a linear variable differential transformer (LVDT) with 0.01 mm precision.

The Optotrak motion analysis system (model 3020, Northern Digital Inc., 403 Albert St., Waterloo, Ontario, Canada N2L 3V2) was used to track infrared emitting diodes (IREDs) on the foot and leg during the stance phase of gait. Three-dimensional coordinate data for each IRED has been reported to have a RMS accuracy of 0.1 mm given the configuration used in this study.[24] Surface markers were tracked at 60 Hz and subsequently filtered at 6 Hz using a zero phase lag, 4th order, Butterworth low-pass filter. A previously performed residual analysis technique determined that a 6-Hz cutoff frequency was appropriate for walking trials.[30] Initial contact and toe-off during stance were determined using a Kistler force plate (Kistler Instrument Corp., Amherst, NY, model 9865B). Force plate data were sampled at 300 Hz and filtered at 8 Hz.

Procedures

Static first ray mobility of the right foot was measured with subjects seated. A custom immobilizer boot stabilized the hindfoot, a dorsal compression clamp immobilized the lesser metatarsals, and a separate platform supported the first metatarsal head. After two preconditioning loads,[16,17] a 55-N dorsal directed force was imposed and first ray dorsal mobility was measured. A single investigator performed all first ray measurements. Same-day repeat measures were taken on 11 subjects to assess reliability.

A four-segment model of the lower extremity was used to estimate angular kinematics of the foot. One segment consisted of the tibial shank identified by a set of three noncollinear IRED markers (Fig. 1). The foot segment was subdivided into the hindfoot, midfoot, and first ray. Hindfoot motion was tracked from a set of three IREDs placed on the calcaneus. One IRED on the navicular and two on the fifth metatarsal modeled midfoot motion. To monitor first ray motion, an IRED triad mounted on a 2-cm post was attached to an orthoplast base mounted over the first metatarsal.

Following marker placement, subjects stood with their feet positioned parallel to the plane of motion while 13 bony landmarks on the foot and tibia were digitized. The digitized points represented the skeletal system relative to the IRED markers and defined anatomically referenced axes in order to calculate Euler angles. Motion of the hindfoot, midfoot, and first ray segments was expressed relative to the proximal segments. Movement about the x-axis was defined as inversion/eversion, about the y-axis as abduction/adduction, and about the z-axis as dorsiflexion/plantarflexion. Intra- and interrater reliability for two investigators was established by digitizing the bony landmarks on a subject twice within one session.

Subjects walked along a 10-m walkway until comfortable with the applied markers. Speed was regulated by

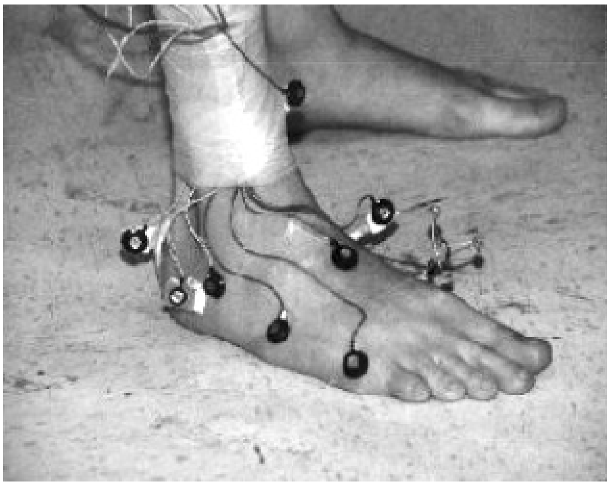

Fig. 1: IRED marker placement for tracking segmental motion. Marker 7 was an additional reference marker for system tracking.

Segment	Marker #	Location
First ray	1,2,3	First ray
Midfoot	4	Base 5th metatarsal
	5	Navicular tubercle
	6	Head 5th metatarsal
Reference	7	Base 2nd metatarsal
Hindfoot	8	Inferior lateral calcaneus
	9	Posterior calcaneus
	10	Anterior lateral calcaneus
Shank	11	Distal tibia
	12 (not shown)	Medial tibia
	13 (not shown)	Superior tibia

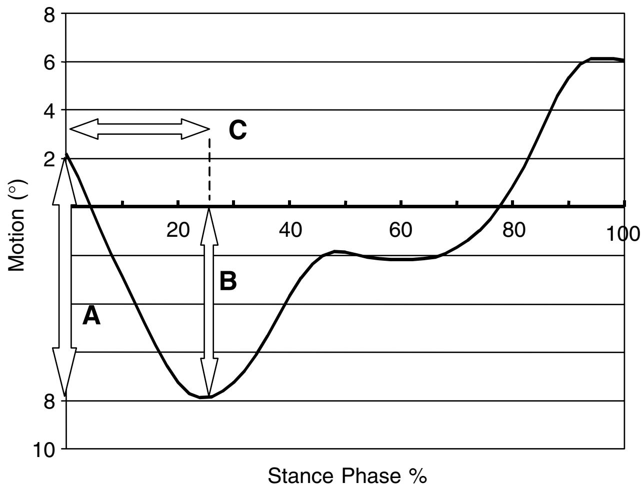

Fig. 2: Gait variables determined from walking trials. **A**, *Total excursion* is the degrees of motion from initial movement into eversion or dorsiflexion until peak motion. **B**, *Peak motion* is the maximum eversion or dorsiflexion during stance. **C**, *Time-to-peak* motion is the percent of stance phase when peak motion was achieved.

an overhead tracking system with streamers to assist subjects in maintaining a constant 3 mph (1.88 m/s) pace. Data were collected on five successful trials. A trial was considered successful if the subject's entire test foot landed on the force plate concealed in the walkway and there were no obvious observable gait deviations.

Data Analysis

Intraclass correlation coefficients (ICC 2,1) assessed intrarater reliability of the static measure of first ray mobility obtained by the device. Data collected by the Optotrak motion analysis system were processed using the KinGait 3 software package (KinGait 3, Mishac, Inc., University of Waterloo, Waterloo, Ontario, Canada). Angular movement of the first metatarsal relative to the midfoot, the midfoot relative to the hindfoot, and the hindfoot relative to the tibia was calculated and used for data analysis. ICCs (2,1) were calculated to determine the inter- and intrarater reliability for bony landmark digitization.

Analysis of stance phase dependent variables included peak motion, time-to-peak, and total excursion into eversion or dorsiflexion. Peak motion was defined as the maximum level of motion recorded during stance. Time-to-peak values were recorded as the percent of stance phase at which peak motion occurred. Total excursion was identified as the movement occurring within a plane of motion from early contact to peak motion (i.e., peak hindfoot inversion to peak hindfoot eversion). The mean and standard deviations of peak motion, time-to-peak, and total excursion (Fig. 2) of first ray dorsiflexion and eversion, midfoot eversion, and hindfoot eversion were calculated from a subject's set of five walking trials.

Pearson correlation coefficients (r) were calculated to assess the degree of relationship between the static measure of first ray dorsal mobility and dynamic measurements of first ray, midfoot, and hindfoot motion. An alpha level of .05 was accepted. Coefficients of determination (r^2) were used to estimate the amount of variance accounted for by the independent variable of static first ray mobility.

RESULTS

Static first ray mobility ranged from 3.1 mm to 8.0 mm. The ICC for intrarater reliability coefficient for the measure was .96. The range of values for dynamic measures of the first ray, midfoot, and hindfoot are summarized in Table 1. Inter- and intrarater ICC reliability coefficients were .99 for digitization of bony landmarks by two investigators.

Table 1: Range of first ray, midfoot and hindfoot dynamic motion and percent stance where peak dorsiflexion or eversion occurred

Segmental Motion	Peak (°)	Stance at Peak (%)	Total Excursion (°)
First ray dorsiflexion	1–8	23–95	3–9
First ray eversion	0–14	74–99	3–12
Midfoot eversion	2–23	49–98	6–35
Hindfoot eversion	4–12	22–56	5–21

Table 2: Relationship between static first ray mobility and midfoot and hindfoot eversion (EVR) motion

Static Ray Mobility vs. Dynamic:	*r*	*p*	r^2
Midfoot peak eversion	.59	<.05	.35
Midfoot excursion	.61	<.05	.37
Hindfoot time-to-peak EVR	.72	<.01	.52
Hindfoot excursion	.73	<.01	.53

The association between the static measure of first ray mobility and dynamic first ray motion was not significant (r values ranged from $-.37$ to. 47, $p > .05$). Poor to moderate association was also found between static first ray mobility and midfoot time-to-peak eversion ($r = .16$) and hindfoot peak eversion ($r = .15$). Significant correlation (Table 2) was found between the static measure of first ray mobility and midfoot peak eversion ($r = .59$), midfoot eversion excursion ($r = .61$), hindfoot eversion excursion ($r = .73$), and hindfoot time-to-peak eversion ($r = .72$).

DISCUSSION

This is the first study to examine the relationship between a static measure of first ray mobility and dynamic motion of the first ray, midfoot, and hindfoot during the stance phase of walking in healthy subjects. A direct relationship was found between first ray mobility and midfoot and hindfoot eversion variables. Surprisingly, the static measure of first ray mobility did not have a strong relationship with dynamic frontal/sagittal plane movements of the first ray in this nonsymptomatic population. The investigators predicted that static dorsal laxity of the first ray would manifest itself as increased or prolonged first ray dorsiflexion during weightbearing activity.[19] The inability to establish a strong relationship between static first ray mobility and dynamic movement of the first ray underscores the complexity of segmental movement and associated mechanical linkages. During dynamic, unrestricted movement, the influence of first ray mobility will be modulated by mechanical linkages in the midfoot to hindfoot and forefoot to midfoot. These factors may be subject dependent and difficult to model. Along with ligamentous and fascial connections, the neuromuscular system is active under dynamic conditions and another source of compensatory influence and individual variation.

Pronation of the subtalar joint lowers the first ray to the ground during early stance.[18] As body weight moves forward during mid to late stance, supination acts to stabilize the medial arch, with the first ray acting as a beam of support for the increasing rigidity of the foot.[18,28] During supination, the peroneus longus, which inserts onto the lateral plantar base of the first metatarsal and medial cuneiform, everts and plantarflexes the first ray.[20,25] The windlass mechanism of the plantar fascia as described by Hicks[18] also plantarflexes the first ray. Together, the peroneus longus and plantar fascia stabilize the medial column of the foot during mid to late stance,[20] which allows for subtalar joint supination and formation of the rigid foot lever needed for effective forward propulsion.

It has been proposed that hypermobility of the first ray reduces the rigidity of the framework of the medial arch as supination occurs.[18] Without a rigid lever system, hindfoot pronation may be prolonged past midstance. In addition, the peroneous longus is at a mechanical disadvantage when the medial arch is in a pronated position and less able to effectively stabilize the first ray, contributing further to a less rigid foot structure during mid to late stance.

Pronation of the foot, while being complex, is perhaps most purely represented in the movement of the calcaneus. The results of this study show a strong correlation between increased static dorsal laxity of the first ray and increased time-to-peak hindfoot eversion and eversion excursion. Subjects with increased static dorsal laxity of the first ray reached peak pronation later in stance phase and had greater degree of movement from inversion to peak eversion, with subjects starting in a greater degree of inversion rather than moving to greater peaks of eversion. These results support the concept that dorsal laxity of the first ray alters

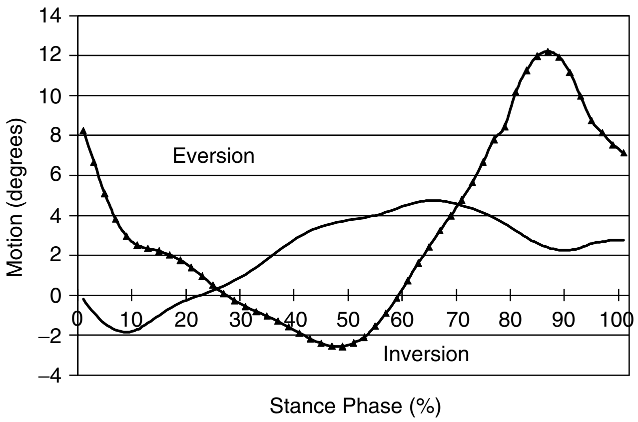

Fig. 3: Comparison of frontal plane motion of the midfoot between three subjects with stiff (—) and three with lax (–▲–) static first ray measures.

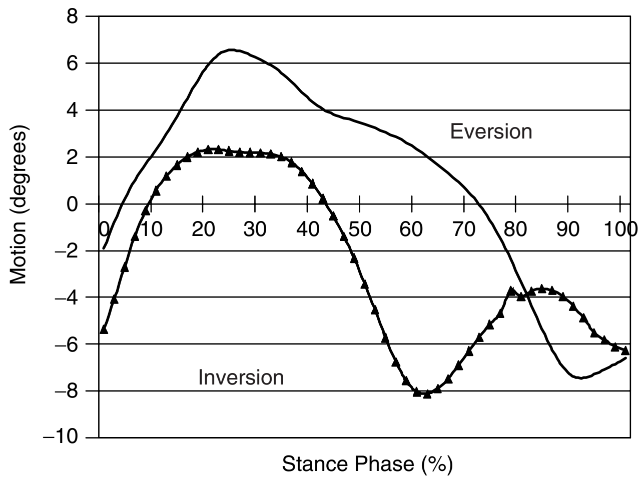

Fig. 4: Comparison of frontal plane motion of the hindfoot between three subjects with stiff (—) and three with lax (–▲–) static first ray measures.

the pronation mechanics of the foot. Our subjects were asymptomatic; however, over time it is possible that individuals with increased first ray mobility could be susceptible to associated problems with push-off mechanics and abnormal loading in the forefoot.

The possibility that individuals at either extreme of first ray mobility might exhibit a response that was less variable was investigated with post hoc analysis. The first ray, midfoot, and hindfoot movement patterns of three subjects with the most lax and three subjects with the stiffest static measures of first ray mobility were compared (Figs. 3–5). The mean for first ray static measures for these two subgroups was 3.4 mm (stiff) and 7.7 mm (lax). The figures support the results of this study. Figures 3 and 4 suggest noted differences in midfoot and hindfoot motion patterns between groups. Comparison of first ray dynamic movement patterns suggest that as the extremes of static first ray mobility are approached, the dynamic response of the foot is less variable, especially in regard to first ray dorsiflexion (Fig. 5). This may help to explain why static dorsal laxity of the first ray did not translate into increased dynamic motion of the first ray, yet midfoot and hindfoot motion was affected. Additional studies comparing stiff versus lax groups would be beneficial to identify the strength of these relationships at the extremes of first ray mobility.

Manual static mobility testing of the first ray is a commonly used clinical test.[1] The pretense of this test is that it provides insight into function of the first ray during gait; however, the utility of manual assessment with regard to dynamic function of the first ray has not been investigated. The advantage of the mechanical device used to assess static mobility of the first ray is the ability to lock out secondary movements in the forefoot and midfoot, which contributes to both the reliability and validity of the measure.[16] The weak correlation

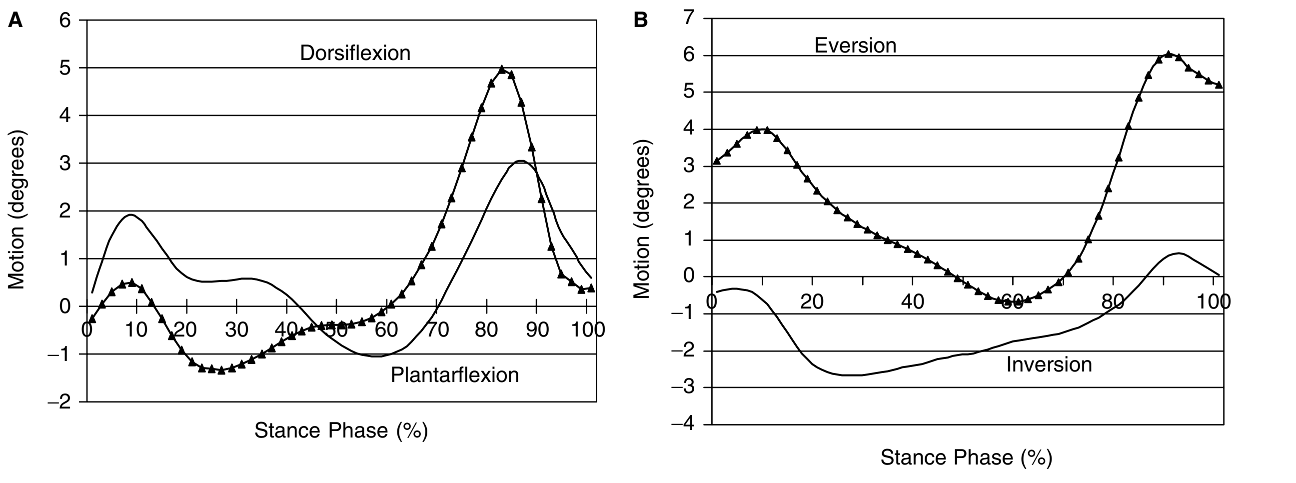

Fig. 5: Comparison between the sagittal (**A**) and frontal (**B**) plane motion between three subjects with stiff (—) and three subjects with lax (–▲–) static first ray measures.

between the dynamic and static measures of first ray motion in this study suggest caution is needed when interpreting both manual and mechanical static test results of first ray laxity in a healthy population and the affect of laxity on first ray motion during gait. This is not to say that the measures are not without merit. A significant relationship between the static mechanical measure of increased first ray dorsal mobility and hallux valgus deformity[15] has been demonstrated. Additional testing is needed to define the correlation of a static measure, whether mechanical or manual, with other symptomatic populations. Longitudinal studies of asymptomatic populations would also help to define the effect of first ray dorsal laxity on the development of mechanical or pathological impairments.

This study adds to the information base establishing the relationship of first ray static mobility to motion patterns of the foot. Additional work is currently underway to examine the relationship between the static measure of first ray dorsal mobility and dynamic motion of the first ray at peak midfoot and hindfoot motion as well as dynamic comparison of motion between foot segments.

CONCLUSION

Static mobility of the first ray was not a strong predictor of first ray dynamic peak motion, time to peak, or overall sagittal/frontal plane motion to peak. Greater static laxity of the first ray was associated with increased midfoot eversion peak motion and excursion as well as a delay in reaching peak hindfoot eversion and greater overall frontal plane motion until peak. Dynamic factors may have a strong influence on motion control of a lax segment during gait. Additional studies of larger populations with stiff and lax static first ray mobility may demonstrate relationships not seen in this study.

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