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Forefoot angle determines duration and amplitude of pronation during walking

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ABSTRACT

The biomechanical mechanisms that link foot structure to injuries of the musculoskeletal system during gait are not well understood. This study had two parts. The purpose of part one was to determine the relation between clinical rearfoot and forefoot angles and foot angles as they make contact with the ground. The purpose of part two was to determine the effects of large vs. moderate values of both forefoot and rearfoot inversion angles at foot contact on foot kinematics. Clinical foot angle, the relationship between the foot and an axis extrinsically defined relative to the ground, was calculated from digital photographs taken in a prone position. During three speeds of over-ground walking, we measured frontal plane rearfoot and forefoot angle relative to the ground at foot contact, and the following stance phase kinematic measures: amplitude of rearfoot and forefoot eversion, duration of rearfoot and forefoot eversion, and duration between heel-off and onset of rearfoot and forefoot inversion. We found that the clinical forefoot angle predicted the forefoot angle at foot contact. Individuals with a large inversion forefoot angle at contact also had greater amplitude of forefoot eversion and everted longer during stance. We discuss the possible mechanisms for the increased risk of injury to the hip reported for individuals that have a large clinical forefoot angle in non-weight bearing. Equally important is the finding that rearfoot angle at contact did not predict the motions of the rearfoot or forefoot during stance.

1. Introduction

Walking may be our preferred form of exercise as we age [1]. Although walking is generally considered a low risk for musculoskeletal injury [2], it may pose a higher risk for individuals with certain foot morphology. While foot morphology has been given significant attention in runners [3], it has been overlooked in walkers.

The question arises as to what foot abnormalities contribute to lower extremity injury while walking. The traditional clinical approach measures foot structure using an intrinsic reference frame defined relative to the proximal segment (forefoot to rearfoot, rearfoot to leg) [4] (Fig. 1, right foot). Studies using this intrinsic measure have found poor correlation between clinical rearfoot angles and rearfoot kinematics during the stance phase of gait [5]. These findings may be due to the poor inter-rater reliability of the traditional measure [6] or because it does not reflect the foot's interaction with the ground.

The latter limitation could be addressed by using a measure in which the clinical rearfoot and forefoot angles are assessed relative to a fixed extrinsic reference axis, the caudal edge of the table (Fig. 1, left foot). This edge defines an axis which is parallel to a mediolateral axis in the participant's frontal plane and is parallel to the ground when standing. Such an extrinsic clinical measure that has a spatial relation to the ground may be more predictive of foot angles at ground contact. In a biomechanical model, Holt and Hamill [7] proposed that the frontal plane foot angles at initial contact would determine how forces and torques are generated, first at the feet, and then through the body. They suggested that it is necessary to assess the effects of the forefoot and rearfoot angles at contact separately since each may have different effects on the forces and torques around the foot and more proximally. While this theory was originally applied to running, it is predicted that it will also apply to walking. Using an extrinsic clinical measure of forefoot structure, Gross et al. [8] found double the occurrence of hip osteoarthritis and a five times greater chance of total hip replacement in older adults when the clinical forefoot angle was larger than 20° of varus (i.e. inversion). Other studies have suggested little to no association between a greater fore foot varus

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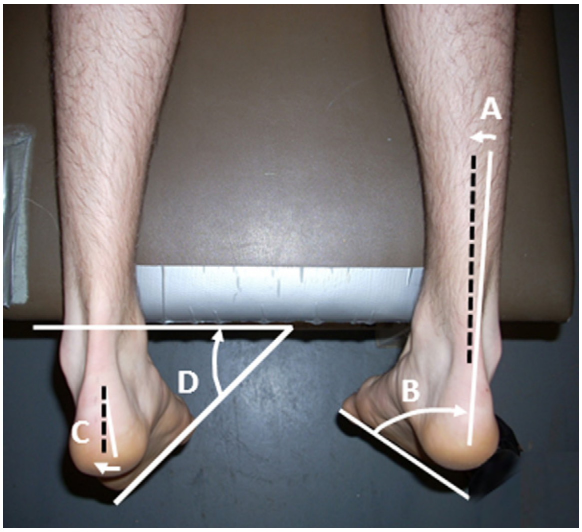

Fig. 1. Clinical foot measures. Right leg: clinical foot angle: rearfoot (A) and forefoot (B) angle to the proximal segment (intrinsic measure). The rearfoot angle is measured as the angle between the line bisecting the calcaneus (white line) and the line bisecting the lower leg (dashed black line). The forefoot angle is measured as the angle between a line (white) through the metatarsal heads and the line bisecting the calcaneus. Left leg: clinical foot angle (extrinsic measure): rearfoot (C) and forefoot (D) angle relative to, respectively, the line perpendicular to the caudal edge of the table (vertical black dashed line) and to the caudal edge of the table (horizontal white line).

angle and lower extremity running injuries [9,10]. The reason may be that these latter studies used the intrinsic measure of foot angle.

Although the intrinsic measure of rearfoot angle is widely used clinically to prescribe orthoses for walkers [11,12] and runners [13], there is evidence that rearfoot structure is not associated with hip pain [8] or running injuries [10]. Nonetheless, rearfoot kinematics during walking [14] and running [15] have been the focus of research investigating mechanisms of injury. In contrast, forefoot structure has been largely overlooked both in the prescription of orthoses, and in research attempting to link kinematics to injury. One study, however, showed that children with a forefoot varus as small as 4° had significantly greater rearfoot pronation and hip flexion during walking when compared to children without forefoot varus [16]. Holt and Hamill [7] proposed that forefoot varus may be particularly relevant in the occurrence of late pronation (between heel-off and toe-off) while running resulting in altered timing, disruption of the synchronous movement of the foot, knee and hip, and proximal injury. These claims have been supported by other researchers during walking [17–19] and running [17,18,20]. The importance of forefoot structure is also highlighted by findings linking forefoot varus and hip injury [8].

Therefore, the purpose of part one of this study was to determine the relation between a newly hypothesized non-weight bearing extrinsic clinical measure of rearfoot and forefoot angles and a weight-bearing measure of rearfoot and forefoot angles as they, respectively, make contact with the ground during walking. The purpose of part two was to determine the effects of large vs. moderate values of forefoot and rearfoot inversion angles at forefoot and rearfoot initial contact on foot kinematics during stance. We hypothesized that the extrinsic non-weight bearing clinical measure of forefoot angle will predict forefoot frontal plane angle at forefoot contact during walking. We also hypothesized that forefoot angle at forefoot contact will predict details of forefoot and rearfoot kinematics during stance, specifically, the amplitude and duration of eversion, and duration between heel-off and onset of inversion.[1]

2. Methods

2.1. Participants

Eight females and six males (mean age 23.6 ± 4.9 years) were recruited for this study. Participants were required to have a forefoot varus greater than 15°. We selected this range because an epidemiological study suggested that only the extremes of forefoot structural abnormality (>20°) had an impact on injury [8]. Individuals with current lower extremity pain were excluded. Each participant signed a consent form approved by the Institutional Review Boards of Boston University and University of Massachusetts, Amherst.

2.2. Procedure

Participants visited the laboratory three times. On the initial visit, they were screened to insure they met all criteria. On the second visit, the participants received a clinical evaluation as part of a larger study. Digital photographs of the feet were taken from above (Fig. 1), with the participant prone, using a six megapixel digital camera (Nikon Coolpix L11) which remained fixed to the wall throughout the study (for a full description see Gross et al. [8]). Briefly, participants were positioned with the medial malleoli aligned with the caudal edge of the table and the legs in neutral rotation. The examiner guided both ankles to 0° of dorsiflexion using light pressure over the third metatarsal head. The participant maintained this position while a digital photograph was obtained. The participant was positioned three times and three digital pictures were obtained.

Participants practiced walking along a 10 m walkway to ensure that the right foot contacted a force plate embedded in the walkway (AMTI, Watertown, MA). The participant's preferred speed was the average of three such walks measured using a photocell-triggered timer. The participant was then asked to walk at self-chosen slower and faster speeds until consistent speeds were reached. It was required that speed levels differ by 0.2 m/s. Only trials within 0.1 m/s of the designated preferred, slow and fast speeds were accepted. Thirteen retro-reflective markers (9.5 mm diameter) were placed on ankle and foot according to the Leardini et al. model [21] (Fig. 2). Once markers were securely taped, sandals were donned carefully with the help of the investigator. All participants wore Bite (LLC) running sandals with straps at the forefoot and rearfoot. As this was part of a larger study investigating effectiveness of orthoses, a neutral foot bed was needed to minimize any effects of the shoe. Although the sandals were designed to provide shock absorption, they do not control motion of the foot.

Following a standing calibration trial, each participant performed five acceptable walking trials at each of the three speeds. On a third visit one month later, the same procedures were repeated but without the clinical evaluation.

2.3. Data reduction

2.3.1. Digital photographs

Rearfoot and forefoot clinical angles were measured from the digital photographs using commercially available software (CanvasX, ACD Systems, Vancouver, British Columbia, Canada) [8]. The angle measuring protocol is outlined in Appendix A. Rearfoot clinical angle was the angle formed by a line bisecting the calcaneus and a line perpendicular to the caudal edge of the table (the table's edge provided an extrinsically defined reference axis relative to the ground) (Fig. 1, left foot). Forefoot clinical angle was

[1] Inversion–eversion as a component of the triplanar motion are used to represent supination–pronation [11,12,14,15].

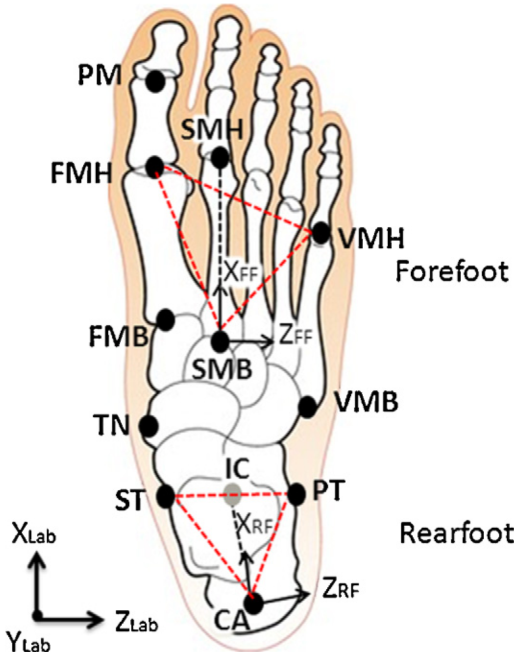

Fig. 2. Foot markers and segments. In accordance with the model by Leardini et al. [21] retro-reflective markers were placed on the bony landmarks of the foot: navicular (TN), first (FMB), second (SMB), and fifth metatarsal bases (VMB), first (FMH), second (SMH) and fifth metatarsal heads (VMH) and the most distal and dorsal point of the head of the hallux (PM). Markers on the calcaneus were located at the triceps surae insertion (CA), the most medial apex of the sustentaculum tali (ST) and the lateral apex of the peroneal tubercle (PT). IC is intermedius calcaneus, the midpoint between ST and PT. The rearfoot segment and forefoot segments are red triangles. The rearfoot segment was defined as having an origin at CA and an *x*-axis joins the origin with IC. The forefoot segment was defined as having an origin at SMB and the *x*-axis is the projection of the line SMB to SMH onto the trasverse plane passing through the origin, VMB and VMH. Image adapted from fotosearch@fotosearch. com.

the angle formed by the line connecting the first and fifth metatarsal heads and the table's caudal edge. The clinical foot angles were the averages from the three photographs. While it is possible that some ambiguity in measurement is introduced by leg position and by projecting a 3-dimensional foot posture onto a 'virtual' frontal plane, previous research [8] has shown good intra- and inter-tester reliability. Our ICC for intra-rater reliability using the same digital photographs was 0.91 for the forefoot and 0.87 for the rearfoot. Gross et al. [8] reported similar intra-rater reliability. Additionally, inter-rater reliability of 0.93 for the forefoot and 0.85 for the rearfoot were reported. The validity of this measure will be determined by its ability to predict foot angles at ground contact during walking.

2.3.2. Motion capture

Kinematic data were acquired from the right lower extremity of all subjects. During walking, data were collected at 240 Hz using eight Oqus cameras (Qualisys Medical AB, Gothenburg, Sweden) and the force plate. The measurement error of the camera system for marker position was 1.4 mm based on known marker distances. Synchronized raw kinematic and kinetic signals were processed using Visual 3D (C-motion, Inc., Germantown, MD). Raw kinematic data were low-pass filtered using a fourth order, zero-lag Butter worth filter, with a cut-off frequency of 8 Hz [22]. Time derivatives were calculated using the central difference method. Heel contact and toe-off were determined in Visual 3D using the vertical ground reaction force. Stance phase event times were normalized to percent stance. Rearfoot contact was defined as heel contact; forefoot contact was when the third derivative (jerk) of the vertical position data of the fifth metatarsal head marker crossed zero and heel-off was when the jerk of the vertical position data of the lateral heel marker crossed zero [23]. The onset of inversion occurred when the first derivative (velocity) of the frontal plane rearfoot and forefoot angle data were above zero.

Three dimensional angles were calculated using an *X* (flexion/extension), *Y* (eversion/inversion), and *Z* (axial rotation) Cardan rotation sequence [24]. While foot motion was measured in the frontal plane as eversion and inversion, these measures are proxies for pronation and supination.

The rearfoot segment included the calcaneus and the forefoot segment included the metatarsals (Fig. 2) [21]. The origin of the rearfoot segment was at insertion of the triceps surae. The origin of the forefoot segment was at the second metatarsal base. Frontal plane angles of the rearfoot and forefoot were reported relative to the laboratory coordinate system. This approach assesses the foot motion relative to the ground.

Custom Matlab programs extracted rearfoot angle at rearfoot contact and forefoot angle at forefoot contact, amplitude of rearfoot and forefoot eversion, duration (% stance) of rearfoot and forefoot eversion, and duration (% stance) between heel-off and onset of rearfoot and forefoot inversion. Amplitude of rearfoot eversion was the difference between the rearfoot angle at contact and the maximum rearfoot eversion angle. Amplitude of forefoot eversion was the difference between the forefoot angle at contact and the maximum forefoot eversion angle. Duration of rearfoot eversion was the percent of stance between contact and onset of rearfoot inversion. Duration of forefoot eversion was the percent of stance between contact and onset of forefoot inversion. Duration between heel-off and onset of rearfoot or forefoot inversion was measured in percentage of stance.

2.3.3. Participant categorization

Participants in this study were grouped twice, once based on the forefoot and once based on the rearfoot (Table 1). Based on forefoot angle at forefoot contact, participants were divided into two equal groups: those with large inversion angle and those with moderate inversion angle. Based on rearfoot angle at rearfoot contact, participants were again divided into two equal groups (large and moderate). This dual categorization allowed us to investigate the influence of contact angle on the kinematic parameters separately for forefoot and rearfoot [8].

2.4. Statistical analysis

A general estimating equation analysis (GEE, Linear) was used to test if mean forefoot and rearfoot angles measured clinically

Table 1
Contact group mean characteristics.

Forefoot grouping	Moderate ($n=7$)	Large ($n=7$)
Forefoot angle at contact	2.6 ± 1.1° (0.7–3.5)	5.9 ± 1.6° (4.1–8.2)
Rearfoot angle at contact	1.3 ± 1.2° (0.2–3.3)	2.4 ± 1.8° (0.3–5.1)
Age	24.9 ± 6.5 years (19–37)	22.8 ± 2.7 years (19–27)
Weight	65.7 ± 14.3 kg (50–90)	66.4 ± 10.5 kg (57–78)
Height	1.7 ± .16 m (1.5–2.0)	1.7 ± .06 m (1.6–1.8)
Clinical forefoot orientation	18.5 ± 2.6° (15.8–21.3)	28.9 ± 2.9° (25.5–31.5)
Clinical rearfoot orientation	1.2 ± 6.7° (−8.2–12)	7.1 ± 7.9° (−2.8–20.1)
Rearfoot grouping	**Moderate ($n=5$)**	**Large ($n=5$)**
Rearfoot angle at contact	0.7 ± 0.5° (0.2–1.3)	3.1 ± 1.3° (1.6–5.1)
Forefoot angle at contact	3.0 ± 1.7° (0.7–5.1)	5.8 ± 2.2° (2.9–8.2)
Age	21 ± 1.4 years (1–23)	22.4 ± 2.4 years (19–24)
Weight	68.7 ± 16.5 kg (50–90)	62.8 ± 10.5 kg (50–68)
Height	1.7 ± .14 m (1.6–2.0)	1.7 ± .13 m (1.5–1.8)
Clinical forefoot orientation	22.2 ± 9.3° (17.0–26.2)	25.4 ± 4.5° (18.7–32.6)
Clinical rearfoot orientation	4.6 ± 11.9° (−8.2–20.1)	4.4 ± 4.4° (−2.8–9.2)

(independent variable) could predict mean forefoot and rearfoot angles at initial contact during walking (dependent variable). The within subject variables were speed (preferred, slow, and fast) and visit (second and third).

Two separate three factor ANOVAs were performed for forefoot and rearfoot. The three factors were group (large and moderate), speed (preferred, slow, and fast) and visit (second and third). The dependent variables were: (1) amplitude of rearfoot eversion, (2) amplitude of forefoot eversion, (3) duration (% stance) of rearfoot eversion, (4) duration (% stance) of forefoot eversion, (5) duration (% stance) between heel-off and the onset of rearfoot inversion, and (6) duration (% stance) between heel-off and onset of forefoot inversion. Statistical significance was set at an alpha level of 0.05.

3. Results

3.1. Measurement error

The mean standard deviation of the three clinical foot angle measurements for all participants was 3° for the forefoot and 3.4° for the rearfoot.

3.2. Effect of visit

There was no main effect or interaction effect of visit. Therefore, the data from the two visits were pooled for the remaining analyses.

3.3. Predicting initial contact angles from clinical angles

There were no walking speed effects for forefoot or rearfoot angle at contact. Clinical forefoot angle predicted the contact forefoot angle during walking ($B = 0.2$, $W(1,11) = 18.7$, $p < 0.001$). The clinical rearfoot angle did not predict the rearfoot angle at contact ($B = 0$, $W(1,7) = .00$, $p = 1.0$).

3.4. Group comparisons

Individuals in the large group based on forefoot contact angle also had large clinical angles (range: 25.5–31.5°, Table 1). A large clinical rearfoot angle was not present in the large group based on rearfoot contact angle (range: −2.8–9.2°).

3.4.1. Amplitude of forefoot and rearfoot eversion

There were no walking speed effects on amplitude of forefoot or rearfoot eversion. The group with a large forefoot angle at forefoot contact had significantly greater amplitude of forefoot eversion during stance than the group with moderate forefoot angle at forefoot contact (Table 2). There was no difference between forefoot groups in amplitude of rearfoot eversion. There was no difference between groups based on rearfoot angle at rearfoot contact in amplitude of rearfoot or forefoot eversion.

3.4.2. Duration of eversion

There was a significant effect of speed on the duration of eversion. As the walking speed increased, the duration in percentage of stance that the forefoot was pronating decreased. The group with a large forefoot angle at forefoot contact had a longer duration in forefoot eversion than the group with moderate forefoot angle at forefoot contact (Table 2). There was no difference between groups based on forefoot angle at forefoot contact in the duration of rearfoot eversion. There was no difference between groups based on rearfoot angle at rearfoot contact in duration of rearfoot or forefoot eversion.

3.4.3. Duration of stance between heel-off and onset of inversion

There were no effects of speed on the duration in percentage of stance between heel-off and onset of inversion. There were no differences between groups based on rearfoot angle or the forefoot angle (Table 2). Individuals with a larger forefoot angle at forefoot contact did have a longer duration of stance between heel-off and onset of forefoot inversion than the group with moderate forefoot angle, but this did not reach significance ($p = .06$).

4. Discussion

There were two parts of this study. The first was to determine if a clinical measure of foot angle (extrinsically defined relative to the ground plane) would predict the frontal plane foot angle at ground contact during walking. In part, this aim was motivated by the finding that people with a large clinical forefoot varus angle are more likely to be injured during gait [8]. Consistent with these findings, our study showed that forefoot clinical angle did predict forefoot angle at forefoot contact. Interestingly, the clinical rearfoot angle neither predicted the incidence of hip osteoarthritis [8], nor rearfoot angle at rearfoot contact. These findings call into

Table 2
Differences in forefoot group means: *F* ratios and *p* values.

Variable	FF group effect		Speed effect	
	F(1,11)	*p*	*F*(2)	*p*
Amplitude of forefoot eversion	14.5	.003**	0.44	0.65
Amplitude of rearfoot eversion	2.7	0.14	0	1
Duration the forefoot is everting	4.9	.05*	6.2	.01*
Duration the rearfoot is everting	1.3	0.29	1	0.39
Duration between heel off and onset of forefoot inversion	4.4	0.06	2.8	0.08
Duration between heel off and onset of rearfoot inversion	2.7	0.13	2.2	0.14

Differences in rearfoot group means: *F* ratios and *p* values

Variable	RF group effect		Speed effect	
	F(1,8)	*p*	*F*(2)	*p*
Amplitude of forefoot eversion	0.92	0.37	0.51	0.61
Amplitude of rearfoot eversion	3.4	0.1	0	1
Duration the forefoot is eversion	0.07	0.78	3.4	0.06
Duration the rearfoot is eversion	0.11	0.75	1	0.37
Duration between heel off and onset of forefoot inversion	0.59	0.47	2.4	0.13
Duration between heel off and onset of rearfoot inversion	0.06	0.82	1.9	0.19

* $p \leq .05$.
** $p \leq .01$.

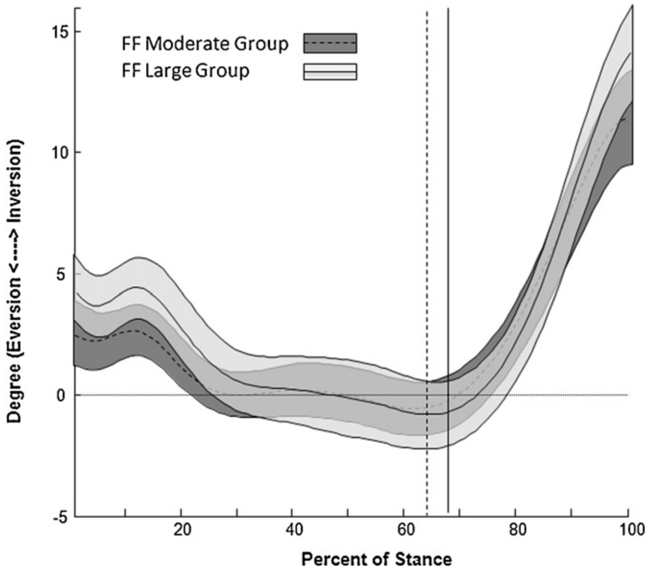

Fig. 3. Mean trajectories and standard deviations of groups indicating amplitude and duration of forefoot eversion. Vertical lines indicate onset of inversion for the moderate group (dotted) and large (solid) group based on forefoot contact angle.

question the common clinical practice of both measuring abnormality and fabricating foot orthoses based on the rearfoot [11–13,25], and the perception of the rearfoot as the implicit cause of musculoskeletal injury [14].

According to Holt and Hamill's model [7] and the findings of Lafortune et al. [26], the angle of the forefoot at ground contact can produce large pronatory torques that result in increased pronation and the continuation of pronation later into the gait cycle. This is supported by our findings that individuals with a large inversion forefoot angle at forefoot contact subsequently had greater amplitude of forefoot eversion and a longer duration of forefoot eversion than individuals with a moderate forefoot angle (Fig. 3).

The study's ability to precisely estimate the relationship of angle at contact to amplitude and duration of eversion and duration between heel-off and onset of inversion were limited by the small number of subjects in each group. It is notable that clinical forefoot angle predicted forefoot angle at contact and that differences were found in stance kinematics when grouped on forefoot angle at contact despite the small number. The rearfoot, however, had small effect sizes for both the duration of rearfoot eversion (0.01) and the duration between heel-off and onset of inversion (0.00), and would require high numbers to achieve significance (>200).

The relationship between the occurrence of greater amplitude and duration of forefoot eversion and the incidence of hip osteoarthritis may be explained by the functional/mechanical relationship between the foot and hip. Mechanically, eversion or more specifically pronation of the foot is accompanied by internal rotation of the tibia due to the arthrokinematic relationship between the subtalar and talocrural joints. If the knee ligaments are stable, this tibial internal rotation facilitates femoral internal rotation [26,27]. The greater amplitude and duration of foot pronation would hypothetically result in increased amplitude and prolonged femoral internal rotation. Hip external rotator muscles would be active to control the internal rotation and therefore increase the compressive forces on the femoral head resulting in altered hip joint loading patterns [28,29]. Over time, the repetitive stresses may contribute to the development of hip osteoarthritis [30]. Future research on the relationship between pronation and transverse plane rotational torques at the knee and hip are currently under way.

5. Conclusion

The results of this study indicate that a non-weight bearing extrinsic clinical measure of forefoot angle estimates forefoot angle at ground contact. This clinical measure accurately and reliably reflects biomechanics that occur following ground contact, and supports the measure proposed by Holt and Hamill [7]. In evaluating and treating foot dysfunction it is critical to focus on the forefoot as a potential cause of injury.

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Conflict of interest statement

There are no conflicts of interest.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2012.10.003.

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