



Original article

Clinical measures of hip and foot–ankle mechanics as predictors of rearfoot motion and posture



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ABSTRACT

Health professionals are frequently interested in predicting rearfoot pronation during weight-bearing activities. Previous inconsistent results regarding the ability of clinical measures to predict rearfoot kinematics may have been influenced by the neglect of possible combined effects of alignment and mobility at the foot–ankle complex and by the disregard of possible influences of hip mobility on foot kinematics. The present study tested whether using a measure that combines frontal-plane bone alignment and mobility at the foot–ankle complex and a measure of hip internal rotation mobility predicts rearfoot kinematics, in walking and upright stance. Twenty-three healthy subjects underwent assessment of forefoot–shank angle (which combines varus bone alignments at the foot–ankle complex with inversion mobility at the midfoot joints), with a goniometer, and hip internal rotation mobility, with an inclinometer. Frontal-plane kinematics of the rearfoot was assessed with a three-dimensional system, during treadmill walking and upright stance. Multivariate linear regressions tested the predictive strength of these measures to inform about rearfoot kinematics. The measures significantly predicted ($p \leq 0.041$) mean eversion–inversion position, during walking ($r^2 = 0.40$) and standing ($r^2 = 0.31$), and eversion peak in walking ($r^2 = 0.27$). Greater values of varus alignment at the foot–ankle complex combined with inversion mobility at the midfoot joints and greater hip internal rotation mobility are related to greater weight-bearing rearfoot eversion. Each measure (forefoot–shank angle and hip internal rotation mobility) alone and their combination partially predicted rearfoot kinematics. These measures may help detecting foot–ankle and hip mechanical variables possibly involved in an observed rearfoot motion or posture.

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1. Introduction

Development of painful musculoskeletal conditions has been attributed to altered pronation of the foot (commonly measured as eversion around a longitudinal axis of the foot) (Willems et al., 2007; Barton et al., 2009; Barton et al., 2010). Thus, health professionals are frequently interested in identifying clinically measurable variables that influence rearfoot pronation and that are susceptible to intervention (Hamill et al., 1989; Hunt et al., 2000; Cornwall et al., 2006). Varus/valgus bone alignment of forefoot, rearfoot and tibia–fibula have been considered as variables that affect pronation magnitude (Root et al., 1977; Michaud, 1993). However, studies that investigated these relationships produced inconsistent results (Hamill et al., 1989; McPoil & Cornwall, 1996a;

Donatelli et al., 1999; Cornwall et al., 2004). It is possible that the mobility provided by midfoot soft tissues, in the frontal plane (i.e. around the longitudinal axis of the foot), also influence rearfoot kinematics. When the forefoot is on the ground, in weight bearing, eversion of the rearfoot is accompanied by motions at the midfoot joints (Neumann, 2002). These motions permit the metatarsal heads, as a unit, to invert relative to the rearfoot and stay horizontally supported (Neumann, 2002). Thus, the soft tissues that resist this collective inversion of the midfoot joints may also resist weight-bearing rearfoot eversion (Fig. 1). The mobility of collective midfoot inversion index this resistance such that the greater the mobility, the smaller the resistance. Therefore, together with varus bone alignment, greater midfoot inversion mobility would contribute to greater weight-bearing rearfoot eversion.

Soft tissues at the hip may also influence foot kinematics by affecting lower-limb axial rotations (Fonseca et al., 2007; Snyder et al., 2009; Souza et al., 2010). According to traditional theories, shank axial rotation would be transferred to the talus and, due to

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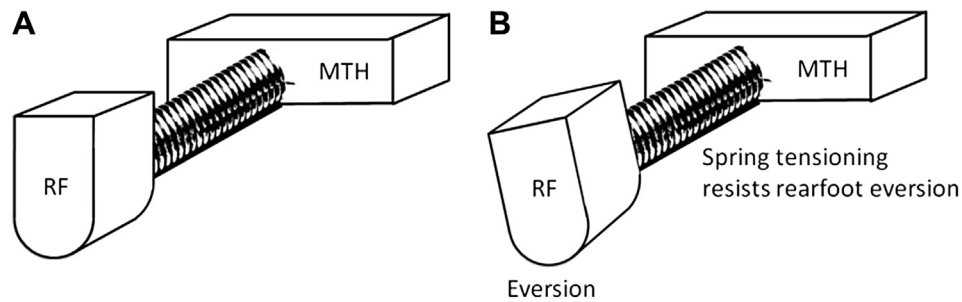


Fig. 1. (A) Model of the soft tissues of the midfoot joints as a torsion spring connecting rearfoot to forefoot, which would influence rearfoot eversion in weight-bearing activities. The figure shows a posterior-lateral view of the right foot, and the forefoot is represented as a bar corresponding to the metatarsal heads horizontally supported on the ground. (B) The torsion spring resists rearfoot eversion relative to the metatarsal heads (i.e. inversion of the metatarsal heads relative to the rearfoot). MTH: metatarsal heads; RF: rearfoot.

the oblique axis of the subtalar joint, talus adduction–abduction would lead to calcaneus eversion–inversion (Root et al., 1977). There would be no axial rotations of the talus in the talocrural joint (Michaud, 1993). Although bone-pin studies questioned these mechanisms at individual joints (Arndt et al., 2004; Nester et al., 2007; Lundgren et al., 2008), the whole ankle complex makes lower-limb internal rotation and rearfoot eversion to be relatively interdependent and simultaneous (Snyder et al., 2009; Souza et al., 2010). Therefore, hip soft tissues that resist internal rotation may also resist foot pronation indirectly. Greater values of hip internal rotation mobility would be associated with greater values of rearfoot eversion.

The aim of this study was to investigate whether a measure of hip internal rotation mobility and a measure that combines midfoot inversion mobility and varus/valgus bone alignment predict rearfoot kinematics, during walking and upright stance.

2. Methods

2.1. Subjects

Twenty-three young and healthy subjects (9 men, 14 women) participated in the study. Their mean (\pm SD) age, mass and height were 24.6 ± 4.01 years, 69.59 ± 12.22 kg, and 1.71 ± 0.09 m, respectively. They constituted a convenience sample from the university community, who met the following inclusion criteria: not having symptoms or any pathology in the lower limbs and lumbo-pelvic complex during the six months previous to the study; not having undergone orthopedic surgery; not having used any kind of foot orthoses; and having a maximum body mass index of 25 kg/m^2 . One lower limb of each subject was studied. The dominant limb was chosen for standardization and was defined as the limb that the subject would use to kick a ball. The sample size was estimated considering a moderate effect size ($r = 0.5$) for the association of foot kinematics with the combination of both clinical measures, with significance level of 0.05 and statistical power of 0.8 (Portney & Watkins, 2000). Moderate association was considered due to the multifactorial nature of foot kinematics. The participants signed a consent form and the Institution's Ethics Committee approved this study.

2.2. Procedures

2.2.1. Clinical measures

2.2.1.1. Forefoot–shank angle. This measure was developed to include varus/valgus bone alignments of the foot–ankle complex as well as midfoot inversion mobility (Holt & Hamill, 1995). The assessment was carried out with a goniometer by measuring the angle between the forefoot and a line drawn on the posterior aspect

of the shank, with the subject lying prone (Mendonça et al., 2013) (Fig. 2A). The shank line connects a proximal reference, at the midpoint between the medial and lateral extremes of the tibial plateau, and a distal reference, at the midpoint between the medial and lateral malleoli. These references were obtained with an analogic caliper rule. Initially, the examiner placed the ankle at 0° of flexion–extension and measured it with a goniometer. Then, the subject was asked to actively maintain this position while the examiner measured the forefoot–shank angle. The required muscle contraction hampers palpation of the talus. Thus, the subtalar joint was not placed in neutral position as in traditional measures (Michaud, 1993). To measure the forefoot–shank angle, the fixed arm of the goniometer was aligned with the shank line and the moving arm was visually aligned with a rod fixed with velcro® on the plantar surface of the metatarsal heads (Fig. 2A). To standardize the transverse-plane position of the assessed lower limb, the posterior aspect of the calcaneus was maintained facing upwards by positioning the contralateral lower limb with hip external rotation and knee flexion. The same examiner conducted this measurement in all subjects. Three repetitions were carried out and their mean value was registered for each subject. For this measure, greater inversion angles of forefoot–shank result from a combination of greater varus alignments and midfoot inversion mobility (Fig. 3). Positive scores represented inverted positions. The description of ankle–foot alignment and mobility components is shown below.

2.2.1.2. Components of foot–ankle bone alignment. Since the forefoot–shank angle is a relationship between the metatarsal heads and a line on the shank, varus/valgus bone alignments of rearfoot and forefoot influence this angle (Fig. 3) (Mendonça et al., 2013). The shank line used, instead of the traditional line representing only the distal third of the shank (Tomaro, 1995), allowed including also tibio-fibular varus/valgus in the final angle obtained (Fig. 3).

2.2.1.3. Component of midfoot inversion mobility. The forefoot–shank angle is also influenced by the inversion mobility at the midfoot joints (Fig. 3D). Activity of the tibialis anterior muscle was required to maintain the ankle positioned at 0° of flexion–extension (Fig. 2B). Because the insertions of this muscle on the mid- and forefoot are medial, contraction pulls the midfoot joints into inversion. The result is inversion of the line of the metatarsal heads (Fig. 3D). The amount of inversion depends on the mobility at the midfoot joints, such that the greater the mobility, the greater the inversion produced (Holt & Hamill, 1995; Mendonça et al., 2013).

2.2.2. Mobility of hip internal rotation

The passive mobility of hip internal rotation was measured as the “position of first resistance” described by Carvalhais et al. (2011) (Fig. 4), with an analogic inclinometer. Hip internal rotation values

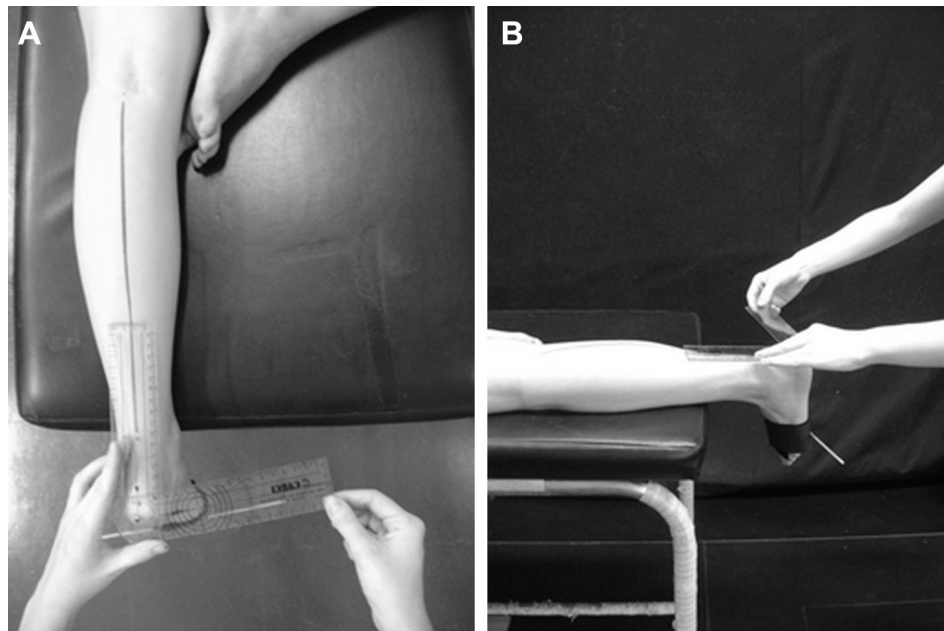


Fig. 2. Measure of forefoot–shank angle. (A) Superior (examiner) view. (B) Lateral view.

were divided by body mass, which appropriately reflects the resistance offered by hip tissues against internal rotation (Carvalhais et al., 2011). Greater angle values represented higher hip internal rotation mobility. The same examiner conducted this measurement in all subjects. Three repetitions were carried out and their mean value was registered for each subject.

2.2.3. Rearfoot kinematics

Frontal-plane kinematics of the rearfoot relative to the shank was measured during the stance phase of walking and during bipedal relaxed standing. A three-dimensional analysis system (Qualisys ProReflex240, QUALISYS MEDICAL AB, Gothenburg, Sweden), with eight cameras and reflective markers, measured the kinematics. Anatomical markers were used for the kinematic models of the rearfoot and the shank. For the rearfoot, two

proximal markers were placed on the lateral and medial malleoli and two distal markers were fixed on the peroneal tuberosity and sustentaculum tali. For the shank, two proximal markers were placed on the lateral and medial epicondyles of the femur and two distal markers were fixed on the lateral and medial malleoli (the same proximal markers of the rearfoot). Clusters of tracking markers were used to measure position changes of the rearfoot and shank. The rearfoot cluster was comprised of a flexible metallic basis with three rigid rods to which three tracking markers were attached (Souza et al., 2010). This basis was firmly attached to the posterior aspect of the calcaneus, below the insertion of the calcaneus tendon. The shank cluster consisted of an elastic belt with a rigid plate to which three tracking markers were attached. This cluster was attached to the distal third of the shank (Manal et al., 2000). Two technical markers attached to the lateral aspect

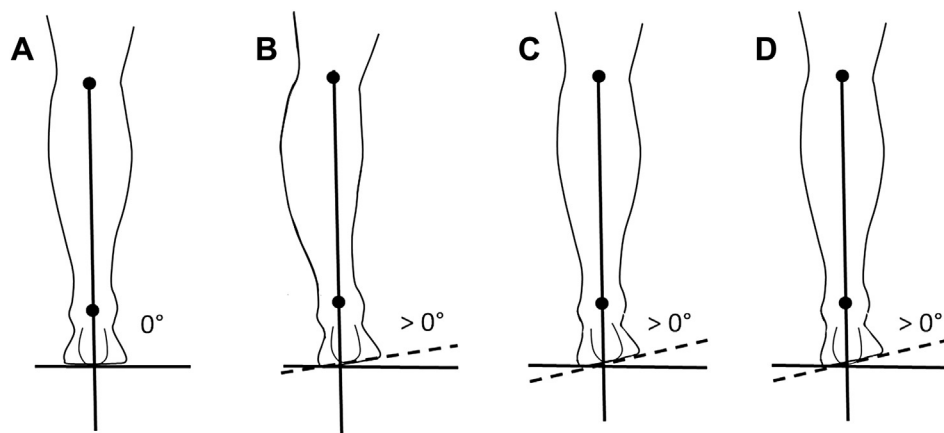


Fig. 3. Isolated influences of alignment and mobility components on the forefoot–shank angle. These components affect the frontal-plane position of the metatarsal heads (dashed line) relative to the shank line (solid line). The illustration shows the posterior view of the left lower limb. (A) The position considered as neutral (0°). (B) Influence of varus alignment of the shank on the angle measured ($>0^\circ$). Note that the forefoot line is inverted because the distal portion of the shank is inverted. Note also that using the proximal reference shown for the shank line, instead of the traditional bisection line on the distal third of the shank, makes the measured angle sensitive to variations in shank frontal-plane alignment. (C) Influence of varus alignment of the rearfoot on the angle measured ($>0^\circ$). Forefoot line is inverted because the rearfoot is inverted. (D) Influence of varus alignment of the forefoot and/or influence of midfoot mobility on the angle measured ($>0^\circ$). Forefoot line is inverted because the forefoot has a varus alignment and/or due to contraction of the tibialis anterior.



Fig. 4. Measure of hip internal rotation mobility.

of the foot, one on the lateral aspect of the calcaneus and the other on the fifth metatarsal head, were used to define the stance phase of walking. Markers' placement is shown in Fig. 5. The same examiner placed the markers, in all subjects.

A static trial was recorded with the subject in standing and wearing all the markers, to subsequently create the kinematic model and define neutral rearfoot position. In this trial, the same examiner placed the foot in subtalar neutral position, which was defined as the position in which the examiner felt the talus head equally between his thumb and index finger (Sell et al., 1994). The subject was further asked to stay in upright stance, with the foot in

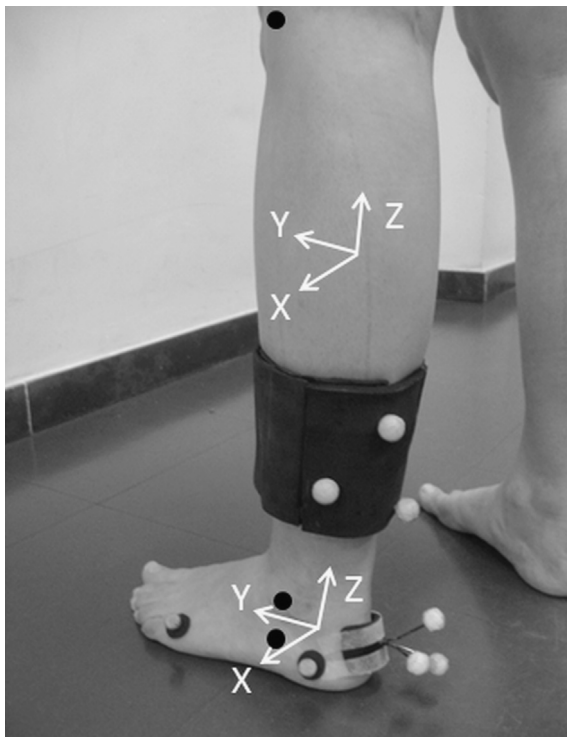


Fig. 5. The clusters of tracking markers and the technical markers mounted on the lower limb. Black circles represent the locations of the lateral anatomical markers. Medial anatomical markers are not shown. White arrows represent the coordinate systems (X, Y, Z) of the shank and rearfoot.

its natural position, with only the tracking and technical markers. Three standing trials, of three seconds each, were collected. Afterward, the subject was asked to walk barefoot on an electronic treadmill ProAction G635 Explorer (BH Fitness, Vitoria-Gasteiz, Spain), in his/her self-selected preferred speed. After a thirty-seconds walk in the selected speed for familiarization, one trial with at least ten steps was registered for each subject. All data collection was recorded with a frequency of 120 Hz.

2.3. Data reduction

Coordinate systems were created for the rearfoot and for the shank, using the anatomical markers of each segment. Each coordinate system comprised a longitudinal axial axis (Z-axis), an anterior–posterior axis (Y-axis) and a medial–lateral axis (X-axis) (Fig. 5) (Souza et al., in press). Rearfoot kinematics was calculated as the motion of the calcaneus relative to the shank, in the frontal plane (around the Y-axis). Angles with negative signs corresponded to everted positions. Neutral position (0°) of the rearfoot was defined as the neutral subtalar position registered in the initial static trial. These data were lowpass-filtered with a cut-off frequency of 6 Hz (Winter, 2005). All processing was done using the Visual 3D software (C-Motion, Inc, Rockville, Maryland, USA).

Mean rearfoot eversion–inversion position during upright stance was calculated. This variable was calculated for three standing trials, and its mean value was considered for analysis. For the walking trials, the stance phase was defined as the period between initial contact and toe-off. These events were visually determined by two examiners, using the linear motion (Y-axis) of the technical markers (Ghoussayni et al., 2004). Mean rearfoot eversion–inversion position (i.e. average rearfoot position throughout the stance phase) and rearfoot eversion peak during stance were calculated. These variables were calculated for ten stance phases, and their mean values were used for analyses.

2.4. Reliability

A pilot study with ten subjects was conducted to investigate the reliability of the measures and procedures of the study. This pilot study had two evaluation sessions separated by a one-week interval. Intraclass correlation coefficients (ICC(3)) were calculated. The intra-examiner reliability of the procedure for placing the subtalar in neutral position was 0.90 and the intra- and inter-examiner reliabilities of gait events determination were 0.99 (for both the initial contact and toe-off). The ICCs of the clinical and kinematic measures were considered appropriate (Portney & Watkins, 2000) and are presented in Table 1.

2.5. Statistical analyses

Multivariate linear regression analyses were carried out. Forefoot–shank angle and hip internal rotation mobility were entered as independent variables in the regression models. Three regression models were performed, one for each dependent variable (kinematic): mean eversion–inversion position during walking stance, eversion peak during walking stance, and mean eversion–inversion position during upright stance. Determination coefficients (r^2) represented the percentage of variance of the dependent variable that was explained by the independent variables, and standardized beta coefficients represented the explanatory weight of each independent variable for the variance of the dependent variable (Portney & Watkins, 2000). A correlation analysis checked for collinearity between the independent variables (clinical measures) prior to the regression models. The significance level was set at 0.05 for all analyses.

Table 1
Description and reliability of the clinical measures and rearfoot kinematic variables.

Variable	Mean (SD)	Range	ICC	SEM
Forefoot–shank angle	14.63° (6.11)	4.33° to 29.33°	0.91 ^a	2.13
Hip internal rotation mobility	0.48°/kg (0.31)	0.09°/kg to 1.44°/kg	0.99 ^{a,c}	0.031
Mean eversion–inversion in walking	−4.59° (2.11)	−7.98° to −1.29°	0.79 ^b	0.96
Eversion peak in walking	−9.29° (2.34)	−14.73° to −6.79°	0.79 ^b	1.07
Mean eversion–inversion in standing	−4.91° (3.01)	−10.01° to 1.27°	0.98 ^b	0.42

SD: Standard deviation, ICC: intraclass correlation coefficient, SEM: standard error of measurement.

^a Intra-examiner reliability.

^b Test–retest reliability.

^c Recently tested by [Carvalho et al. \(2011\)](#) for the same examiner and co-author that conducted the hip measures of the present study.

3. Results

There was no collinearity between the clinical measures ($r = 0.053$; $p = 0.81$). Forty percent of the variation of mean eversion–inversion in walking was explained by the forefoot–shank angle and hip mobility ($r^2 = 0.40$, $p = 0.006$). Twenty seven percent of the variation of eversion peak, during walking, was explained by both clinical measures ($r^2 = 0.27$, $p = 0.041$). Forefoot–shank angle and hip mobility measures explained thirty one percent of the variation of mean eversion–inversion, during upright stance ($r^2 = 0.31$, $p = 0.023$). [Table 2](#) shows the regression coefficients and significance of the models (regression equations). Each clinical measure contributed significantly for the prediction of mean eversion–inversion during walking and standing. Only the measure of hip internal rotation mobility contributed to the prediction of eversion peak during walking. [Fig. 6](#) illustrates the relationships between the measured kinematic values and the values predicted by the regression equations.

4. Discussion

The results demonstrated that the measures of forefoot–shank angle and hip internal rotation mobility are significantly related to frontal-plane kinematics of the rearfoot, during walking stance or relaxed bipedal standing. Greater rearfoot eversion, during walking and standing, was related to greater values of hip internal rotation

Table 2
Coefficients and significance of each clinical measure in predicting the kinematic variables.

Clinical measures	β	Standardized β	p	r^2 measure ^a	r^2 combination ^b
Model for mean eversion–inversion in walking					
Forefoot–shank angle	−0.153	−0.447	0.019*	0.20	0.40*
Hip internal rotation mobility	−3.197	−0.479	0.012*	0.23	
Model for eversion peak in walking					
Forefoot–shank angle	−0.056	−0.146	0.452	0.02	0.27*
Hip internal rotation mobility	−3.752	−0.509	0.015*	0.26	
Model for mean eversion–inversion in standing					
Forefoot–shank angle	−0.196	−0.397	0.045*	0.16	0.31*
Hip internal rotation mobility	−3.957	−0.417	0.036*	0.17	

β : beta coefficient.

Standardized β : standardized beta coefficient, which represents the explanation weight of each independent variable (clinical measures).

p : significance for each clinical measure.

* Statistically significant ($p \leq 0.05$).

^a determination coefficient (r^2) of each clinical measure.

^b determination coefficient (r^2) of the combination of clinical measures.

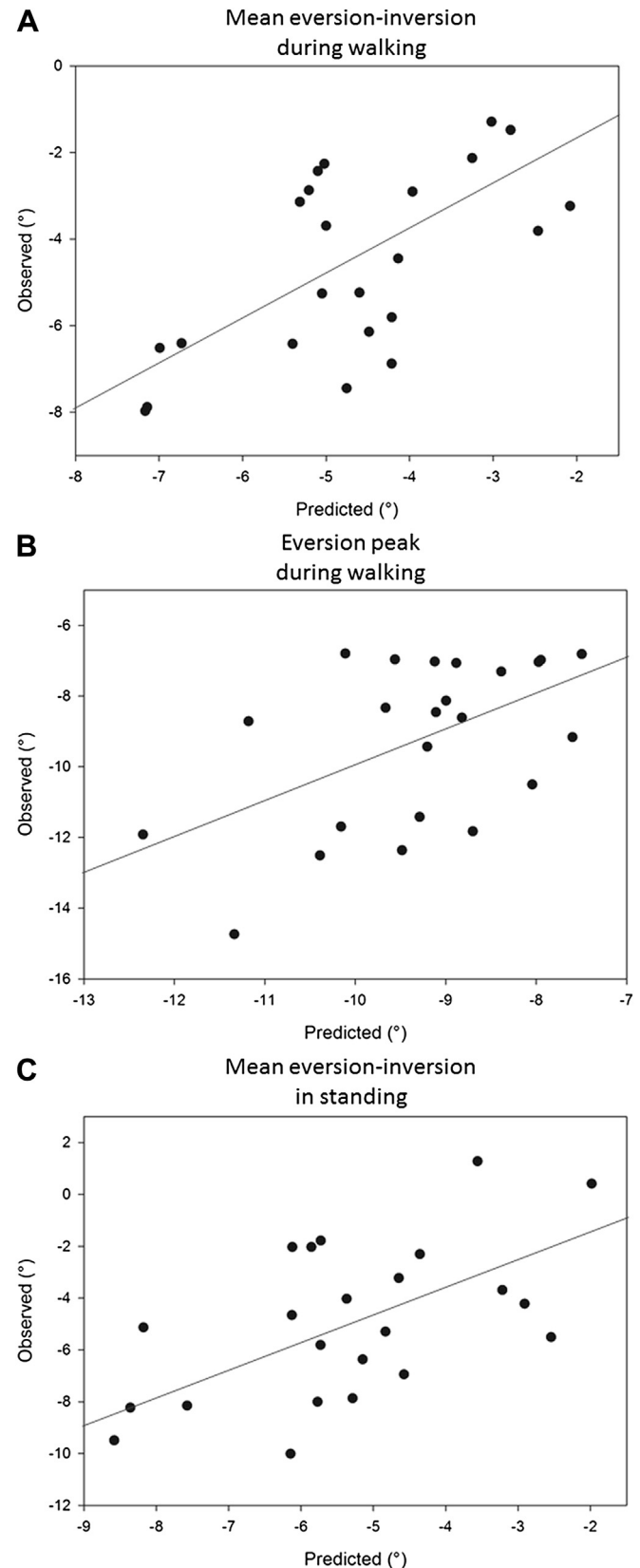


Fig. 6. Relationships between the kinematic variables observed (measured) and the kinematic variables predicted by the regression equations. (A) Mean rearfoot eversion–inversion, in walking. (B) Peak rearfoot eversion, in walking. (C) Mean rearfoot eversion–inversion, in standing. The slope lines represent theoretical perfect predictions.

mobility and greater values of the forefoot–shank angle (i.e. combination of varus bone alignments at the foot–ankle complex with inversion mobility at the midfoot joints). The clinical measures predicted together 27% to 40% of the variance of rearfoot kinematics (Table 2 – last column). Both measures contributed significantly to the explanation of the mean rearfoot eversion–inversion, during walking (40%) and upright stance (31%). The standardized beta coefficients of the clinical measures were similar (Table 2 – third column), which showed that both measures had similar explanatory weights for rearfoot kinematics. Further, peaks of rearfoot eversion, in walking, were also related to the combination of clinical measures (27% of explanation). However, this kinematic variable was significantly related only to hip internal rotation mobility alone (26% of explanation). The contribution of the forefoot–shank angle to eversion peak prediction was negligible. It is possible that, near the instants of eversion peaks (between 50% and 60% of stance), the contribution of hip mechanics to rearfoot kinematics is dominant over the influence of foot alignment and mobility.

The forefoot–shank angle alone predicted 16% to 20% of mean rearfoot eversion–inversion, in walking and standing (Table 2 – fourth and fifth columns), which is consistent with the results of a recent study that used a similar measure (Monaghan et al., 2013). Hip internal rotation mobility alone predicted 17% to 26% of mean rearfoot eversion–inversion, in walking and standing, and rearfoot eversion peak during walking (Table 2 – fourth and fifth columns). The passive resistance to internal rotation produced by hip soft tissues, which determine the mobility measured, possibly participate in the production of the net hip moment (in the transverse plane) in activities as standing and walking.

The limited explanation for the variance of rearfoot kinematics (27%–40%) shows that the studied clinical measures are not indicated to precisely predict altered foot kinematics. The significant relationships found imply that the use of these simple clinical measures may help to detect, in a patient, mechanical variables possibly involved in an observed rearfoot motion and posture. Weight-bearing measures of navicular height and compensated rearfoot valgus also predict rearfoot eversion partially, during walking (McPoil & Cornwall, 1996a,b; Hunt et al., 2000). Such measures constitute postural measures of the weight-bearing pronation and not measures of pronation causes that could be subjected to intervention. For example, decreased navicular height indicates excessive pronation but not whether pronation is resultant from hip and/or foot–ankle mechanics. Differently, the measures of hip mobility and forefoot–shank angle are non weight-bearing and indicate possible causes for rearfoot kinematics, which can be changed through clinical interventions (Vicenzino, 2004; Ocarino et al., 2008; Nakamura et al., 2012) in attempt to prevent or treat painful conditions related to altered foot pronation.

Some study limitations should be pointed out. Although the present and other studies observed significant associations of the forefoot–shank angle with lower-limb kinematics (Bittencourt et al., 2012; Monaghan et al., 2013) the validity of the quantification of midfoot inversion mobility, in the measurement, is still unknown. Further, in the same measure, the active ankle positioning does not enable a precise standardization of the calcaneus position, which does not guarantee that rearfoot bone alignment was appropriately addressed. It is also noteworthy that this measure is limited to patients with no difficulties to contract the tibialis anterior muscle. One potential limitation of the study is that only dominant limbs were measured and it is not known whether dominance would influence the results.

Although the measures of hip internal rotation mobility and forefoot–shank angle are associated to the magnitude of rearfoot eversion in walking and standing, the major part of the foot kinematics variability remained unexplained. Rearfoot kinematics, in

weight-bearing activities, seems to be multifactorial (Fonseca et al., 2007) and may be also influenced by other mechanical variables, such as the axial rotation of the foot at gait initial contact (Wright et al., 1964) or lower limb active muscle function (Fiolkowski et al., 2003; Preece et al., 2008). Therefore, including clinical measures for these other variables, in the assessment of a patient, may improve the explanation of an observed altered foot pronation. Further research is needed to investigate this possibility.

5. Conclusion

The clinical measures of forefoot–shank angle and hip internal rotation mobility partially predict rearfoot kinematics during walking and upright stance. These measures may help detecting mechanical variables possibly involved in an observed rearfoot motion or posture.

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