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Original Research

Non-linear interactions among hip and foot biomechanical factors predict foot pronation during walking in women

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ABSTRACT

Background: Interactions between hip and foot biomechanical factors may result in different magnitudes of foot pronation during walking.
Objective: To investigate non-linear interactions between hip and foot biomechanical factors and their capability to predict foot pronation during walking and identify the profiles of biomechanical factors that predict greater and lower pronation.
Methods: This is a cross-sectional study. Fifty-one women were classified into greater and lower foot pronation during walking. Biomechanical factors measured: (1) foot-ankle varus alignment, (2) hip passive stiffness, (3) isokinetic eccentric strength of hip external rotators, and (4) foot abduction angle. Classification and regression trees (CART) were used to investigate non-linear interactions that predict greater and lower foot pronation.
Results: Four main profiles of biomechanical factors were identified as related to greater and lower foot pronation. Profiles for greater pronation were: (1) varus >25.83°; (2) interaction between varus ≤25.83° and hip stiffness ≤0.09 Nm/rad kg^{-1}; (3) interaction between varus ≤25.83°, hip stiffness >0.09 Nm/rad kg^{-1}, and foot abduction >19.58° The profile for lower pronation involved an interaction among varus ≤25.83°, hip stiffness >0.09 Nm/rad kg^{-1}, and foot abduction ≤19.58° The model had 61% sensitivity and 96% specificity, with the total prediction of 78%. The area under the ROC curve was 0.79 ($p = 0.001$).
Conclusion: Foot-ankle varus, hip passive stiffness, and foot abduction predicted greater and lower foot pronation. Non-linear interactions between hip and foot factors influence the magnitude of foot pronation during walking. The observed profiles help identify which combinations of biomechanical factors should be assessed in individuals with increased or reduced pronation.

Introduction

During walking, greater magnitudes of foot pronation, consisting of increased rearfoot eversion and longitudinal arch lowering, are related to musculoskeletal painful conditions and injuries at the lower limbs and trunk.[1–3] Therefore, researchers have investigated possible causes of excessive foot-ankle pronation during walking. Some proximal and distal biomechanical factors related to increased pronation during walking are large varus alignment of the foot-ankle,[4–7] low hip passive stiffness,[6,8] and increased abducted foot position.[9,10] Varus alignment of the foot-ankle is an anatomical misalignment in which the forefoot, rearfoot, and/or tibia have an inverted position in non-weight-bearing situations.[6] Low hip muscle strength is related to greater foot-ankle eversion posture[11] and greater pronation during running.[12] However, there is no evidence for walking.

While there is evidence on these isolated biomechanical factors, little is known about how their interactions may relate to the magnitude of foot pronation. Different individuals have different combinations of these biomechanical factors. For example, we can ask, “Does a person having large foot-ankle varus, but also high hip strength and stiffness, tend to have large or small pronation magnitudes?”. Souza et al.[6] observed that the linear combination of large varus alignment and low

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hip passive stiffness is related to greater foot-ankle pronation. However, non-linear relationships are expected in complex biological systems,[13,14] and non-linear interactions among those biomechanical factors may have a role in producing greater or lower pronation. Bittencourt et al.,[15] using Classification and Regression Trees (CART), investigated how biomechanical factors interact non-linearly to predict greater magnitudes of knee frontal-plane projection angle during squatting. Distinct profiles composed of biomechanical factors are identified with this approach.[15,16] This approach may help a practitioner to determine whether a client with increased foot pronation belongs to one of the profiles related to greater pronation and then intervene on the identified factors. However, to our knowledge, non-linear interactions among biomechanical factors in predicting the magnitude of foot-ankle pronation have not been investigated yet.

This study investigated whether non-linear interactions between foot-ankle varus alignment, foot abduction angle, hip passive stiffness, and hip strength predict greater and lower foot-ankle pronation during walking in women.

Methods

Sample

A cross-sectional study was conducted with 51 participants. The sample of this study was part of a larger study database. Data collections were performed at the Movement Analysis Laboratory of Universidade Federal de Minas Gerais, in the period between 2012 and 2016. The sample, composed of women, was selected by convenience. Only women were selected to avoid sex-related variability. The inclusion criteria were: age between 18 and 35 years; body mass index less than or equal to 25 kg/m^2; absence of minor symptoms or musculoskeletal injuries in the last three months; absence of history of surgery or major musculoskeletal conditions at the spine, pelvis, and lower limbs; not being engaged in physical exercise or sports in the previous three months; a physiological range of motion of hip internal rotation (from 34° to 71°) and hip external rotation (from 25° to 56°)[17] ; a physiological range of motion of ankle dorsiflexion (above 20°).[18] There was no specific inclusion criterion for the magnitude of foot pronation and the participants were comparatively classified as individuals with lower and greater pronation. The 45th and 55th percentiles of the magnitude of foot pronation were used to dichotomize the sample in greater and lower foot-ankle pronation. These percentiles were chosen to avoid considerable loss of participants (i.e., losing only the participants between the 45th and 55th percentiles). Peak rearfoot eversion was the kinematic variable used to index foot pronation magnitude. Participants who could not maintain the hip muscles relaxed during the hip passive stiffness test and had pain during the tests or could not perform them correctly were excluded from the study.

Procedures

The participants signed an informed consent form agreeing to participate in the study. This study was approved by the Research Ethics Committee (CAAE – 0427.0.203.000–11) of Universidade Federal de Minas Gerais. The lower limb evaluated was selected randomly. The participants were blinded to the possible outcomes of the study. The examiner and participants were blinded to the post hoc classification of the participants into greater and lower pronation.

Varus alignment of the foot-ankle complex

The varus alignment of the foot-ankle was measured as the “forefoot-shank angle”, in non-weight-bearing, prone position. This measure combines midfoot inversion mobility and varus/valgus bone alignment of the foot-ankle and is performed without weight bearing (Fig. 1A, B, C).[6,19] The average of three measures was used for analysis. The intra-examiner reliability of this measure was investigated with 10 individuals, with a one-week interval between the measurements. An Intraclass Correlation Coefficient (ICC$_{3,3}$) of 0.93 was found (95% confidence interval (95% CI) of 0.73, 0.98, standard error of measurement (SEM) of 2.16).

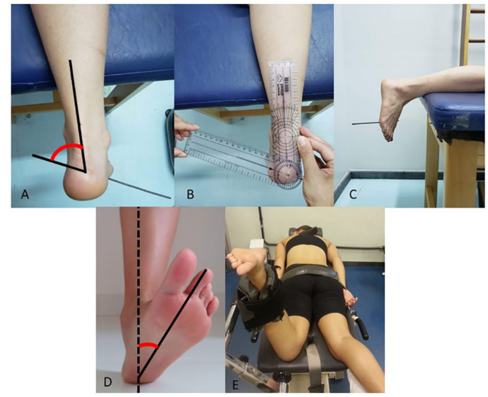

Fig. 1. Superior view (A and B) and lateral view (C) of the measurement of the foot-ankle alignment. This clinical measure provides the forefoot-shank angle, measured in open chain. The black lines in A represent the alignments of the forefoot (heads of the metatarsals) and shank, and the red arc represents the angle between the forefoot and tibia. (D): The foot abduction angle (in red) determined by the angle of the rearfoot (continuous line) relative to the laboratory (dashed line), in the transverse plane, at the initial contact of walking. (E): Participant positioning during the measurements of hip passive stiffness and eccentric strength of the hip external rotators.

Walking kinematics

The Codamotion three-dimensional system (Charnwood Dynamics, Rothley, England), with four capture units, was used for the walking kinematic analysis. Markers were placed on anatomical landmarks to create the kinematic model,[20,21] and an additional marker was placed on the fifth metatarsal head to determine toe-off.[22] Clusters of three active markers were used to track the rearfoot and shank trajectories during barefoot walking. The rearfoot cluster was placed on the calcaneus, below the insertion of the Achilles tendon,[20,21] and the shank cluster was placed at the distal third of the shank[20,21] (Supplementary material, Fig. 1). The participant was first asked to stay in a comfortable standing position for a data collection of 5 s. Subsequently, data for three additional standing trials were collected with the subtalar joint of the participant positioned in neutral by the examiner.[23,24] To do so, the examiner palpated the head of the talus anteromedially and anterolaterally with one hand (using thumb and index finger), and the subject moved the rearfoot into inversion and eversion. The subtalar neutral position was determined when the talus was felt equally between the thumb and index finger.[23] The subtalar neutral position was used to determine the 0° position of the rearfoot relative to the shank during walking, as explained later. No standing measures were taken in the study. The ICC$_{3,3}$ for the intra-examiner reliability of this procedure was 0.94 (95% CI 0.77, 0.99, SEM 1.15°). Finally, the participant walked barefoot on a ProAction G635 Explorer treadmill (BH Fitness – Vitoria-Gasteiz, Alava – Spain) at a self-selected speed. Data for 30 walking support (stance) phases were collected,[25] with a sampling frequency of 100 Hz.

Hip passive stiffness

Hip passive stiffness against internal rotation was measured using a Biodex 3 Pro isokinetic dynamometer (Biodex Medical Systems, Shirley, USA) (Fig. 1e). Electromyography (ME6000, Mega Electronics Inc., Kuopio, Finland) was used to ensure that hip muscles were relaxed. Active surface electrodes were placed on the following muscles: gluteus maximus, gluteus medius, biceps femoris, tensor fascia latae, and adductor magnus.[26] The test was performed from 25° of external rotation to 25° of internal rotation.[7,20] The protocol used in the test was in passive mode, with a sampling frequency of 100 Hz and an angular speed of 5°/s.[7,20] The tibial tuberosity of the assessed limb was aligned with the rotation axis of the isokinetic dynamometer, the dynamometer's attachment was fixed at the distal half of the shank, right below the medial malleolus, and the pelvis was stabilized with a belt. Cruz et al.[7] described this measure in more detail.

Eccentric strength of the hip external rotators

To measure the eccentric strength of the hip external rotators an isokinetic dynamometer (Biodex Medical Systems, Shirley, USA) was used. The participants remained in the same position as in the stiffness test (Fig. 1e) and were instructed to actively resist the movement of hip internal rotation imposed by the dynamometer. The test was performed from 30° of external rotation to 20° of internal rotation with an angular speed of 30 °/s.[7,20] For familiarization, the participant performed one trial with submaximal strength. The examiner gave verbal encouragement for the participant to perform maximum strength. Eccentric strength was recorded for three sets of five repetitions.

Data reduction

Walking kinematics

Data were processed using Visual 3D software (C-Motion, Inc., Rockville, Maryland, EUA). Initially, a kinematic model of six degrees of freedom was created. The shank and rearfoot segments were defined as rigid bodies with their respective coordinate systems.[27] The segments' coordinate systems and their poses were determined according to the positions of anatomical markers at the femoral epicondyles and malleoli, for the shank; and malleoli, peroneal tuberosity, and sustentaculum tali, for the rearfoot.[20,21] The data were filtered with a fourth-order, *Butterworth*, low-pass filter, with a cutoff frequency of 6 Hz.[28] The walking stance phase was determined between the contact of the calcaneus on the ground and the removal of the forefoot from the ground.[22] The subtalar neutral position was used to define the rearfoot neutral position relative to the shank, which was considered the 0° position of the rearfoot angles during the stance phase of walking. Everted positions were negative. The peak (minimum angle of the angle-time curve) was calculated for the eversion of the rearfoot relative to the shank during the walking stance phase and was used to index pronation magnitude. The foot abduction angle was determined by the angle of the rearfoot relative to laboratory in the transverse plane, at the initial contact of the calcaneus in the beginning of the stance phase of walking (Fig. 1d). Peak rearfoot eversion and foot abduction angle were computed as the average values from the 30 stance phases of walking. The walking variables' reliability was verified in a pilot study with seven subjects in two assessments with a seven-day interval. The $ICC_{3,3}$ of the peak rearfoot eversion was 0.87 (95% CI 0.22, 0.98, SEM 1.14°). The $ICC_{3,3}$ for the foot abduction angle was 0.98 (95% CI 0.88, 0.99, SEM 2.05°).

Isokinetic dynamometer

The data obtained by the isokinetic dynamometer were processed with a MATLAB routine, as described by Cruz et al.[7] Hip passive stiffness to the internal rotation movement (i.e. of the external rotator tissues) was calculated as the mean slope of the torque-angle curve (first derivative) in Nm/rad. In the first 15° of hip internal rotation of each repetition, the average value of the multiple slopes obtained between every two subsequent points of the torque-angle curve (0.05° intervals) was calculated.[29] This method was used to consider the nonlinearities and irregularities of the torque-angle curves.

The peak (maximum value) of the eccentric strength was obtained from the test of the maximum strength of the hip external rotators. For statistical analysis, the average peak from the three series was used. The peak of each series was defined as the highest peak of the five repetitions. The values of hip passive stiffness and eccentric strength were normalized by the participant's body mass.

Statistical analysis

Classification and Regression Trees (CART) were used to verify the interaction profiles related to the magnitude of foot-ankle pronation during gait. CART is a multivariate, non-parametric analysis that creates a tree-like classification model. The dependent variable is dichotomized into two classes, in this case, in greater and lower foot pronation. Independent variables (predictors) with specific cutoff values divide the data in a binary and recursive way, forming a tree with increasingly homogeneous nodes in relation to the classes of greater and lower pronation. Thus, it is possible to identify different profiles with different interactions between the predictors, establishing a non-linear relationship between the dependent variable (greater and lower pronation classes) and the independent variable (predictors). The peak rearfoot eversion was the dependent variable. The predictor variables investigated were: hip passive stiffness, eccentric strength of the hip external rotators, foot abduction angle, and foot-ankle varus alignment. The following criteria were applied to avoid over-fitting: growth criteria consisting of a minimum of eight participants for node division and a maximum of four participants to form a node; a Gini index of 0.0001; a tree depth with three levels; and pruning.[30]

The area under the ROC curve (Receiver Operating Characteristic) was analyzed with a significance of $\alpha=0.05$ to verify whether the model had a good prediction of the categories of greater and lower pronation.[31] In addition, prevalence ratios were calculated for each terminal node of the model to investigate the strength of the associations between the dependent and independent variables.

Results

Subjects characteristics

The characteristics of the sample are shown in Table 1. The 45th and 55th percentiles of the peak rearfoot eversion during walking were −10.07° and −9.51° (everted positions are negative). Therefore, participants with peak rearfoot eversion below −10.07° were categorized into greater foot pronation, and participants with peak rearfoot eversion above −9.51° were categorized into lower foot pronation. Participants with peak rearfoot eversion between −10.07° and −9.51° were excluded from the analysis. Therefore, from a total sample of 51 participants, five were excluded, 23 were categorized into lower foot pronation during walking, and 23 were categorized into greater foot pronation during walking.

As this study had four independent variables, and all participants

Table 1
Sample characteristics.

Variables	Lower foot pronation Mean (SD)	Greater foot pronation Mean (SD)
Number of participants	23	23
Age (years)	22 (3)	22 (3)
BMI (kg/m^2)	20.01 (1.86)	20.95 (2.23)
Peak rearfoot eversion (°)	−7.28 (2.10)	−13.77 (2.65)
Foot-ankle varus alignment (°)	14.96 (6.23)	17.36 (8.52)
Hip passive stiffness (Nm/rad kg^{-1})	0.13 (0.03)	0.12 (0.03)
Foot abduction angle (°)	12.89 (6.30)	14.73 (6.21)
Eccentric strength of hip external rotators (Nm/kg)	0.57 (0.13)	0.53 (0.10)

BMI, body mass index; peak rearfoot eversion during walking; foot-ankle varus alignment measured in prone; hip passive stiffness to internal rotation (i.e., stiffness of the hip external rotator tissues); foot abduction angle at the initial contact of the stance phase of walking.

were considered "cases" (i.e., greater or lower pronation), there were at least 10 participants for each independent variable.[32–34] Thus, 46 women were considered an appropriate sample size.

The prevalence ratios for each terminal node are demonstrated in TABLE 2.

The predictive foot pronation model

The predictive model correctly classified 14 out of 23 participants with greater foot pronation during walking (61% sensitivity) and 22 out of 23 participants with lower foot pronation during walking (96% specificity). The total prediction of the model was 78%. The corresponding profiles related to lower and greater foot pronation are shown in Figs. 2 and 3.

The area under the ROC curve was 0.79 (95% CI: 0.66, 0.93; standard deviation=0.34; $p = 0.001$).

Discussion

This study aimed to investigate non-linear interactions between hip and foot-ankle biomechanical factors that predict greater and lower foot pronation during walking in women. The magnitude of foot pronation, indexed by the peak of rearfoot-shank eversion during walking, was strongly predicted by interactions between foot-ankle varus alignment (measured in prone), hip passive stiffness to the internal rotation movement, and foot abduction angle at the initial contact of the stance phase of walking. Profiles consisting of these biomechanical factors were identified. Such profiles specify how the foot-ankle and hip factors interact differently to result in greater or lower magnitudes of foot pronation during walking.

The foot-ankle varus alignment in prone was the first predictor of foot pronation magnitude during walking, and the participants with varus >25.83° (node 2) had a 108% increase in the likelihood of having greater foot pronation during walking. Previous studies that investigated the isolated influence of the foot-ankle varus (forefoot-shank angle in prone) on pronation during walking also identified that a larger varus is related to greater foot pronation during walking.[4,5,7] In contrast, one study did not find this relationship.[8] The present results demonstrated that very large varus alignment alone is related to large foot pronation during walking. Thus, people with excessive foot pronation during walking and foot-ankle varus alignment above 25.83° may benefit from interventions focused on this misalignment, such as medially posted foot orthoses.[35]

The second predictor selected by the model was hip passive stiffness to internal rotation movement. Participants with foot-ankle varus ≤25.83° and hip passive stiffness ≤0.09 Nm/rad kg^{-1} (node 3) had a 98% increase in the likelihood of having greater foot pronation during walking. This finding showed that, even without extreme values of foot-ankle varus, greater foot pronation during walking occurs in the presence of low hip stiffness. This result is consistent with those from Souza et al.,[6] who identified a linear relationship in which the combination of greater values of foot-ankle varus and lower values of hip passive stiffness predicted larger pronation during walking, with 27% of variance explanation. This relatively low explanation may have been influenced by their linear approach, which is incompatible with the influence of several factors interacting non-linearly.[13,14] In contrast, the current non-linear approach reached 78% of variance explanation and identified that both foot-ankle varus >25.83° and ≤25.83° predict greater foot pronation during stance, depending on the interaction with hip passive stiffness to the internal rotation movement. To obtain the most accurate measure, we assessed hip stiffness to the internal rotation movement with an isokinetic dynamometer. The stiffness value of 0.09 Nm/rad kg^{-1} corresponds to nearly 30° of hip internal rotation in the clinical measure of hip passive stiffness (or to nearly 0.5°/kg if a mass-normalized measure is desired).[36] Therefore, subjects with >30° of hip passive internal rotation may be prone to greater foot pronation, although further investigation using the clinical measure is needed. Muscle strengthening aimed at increasing hip passive stiffness, through hypertrophy and/or muscle length reductions,[37,38] might be prescribed to individuals with low stiffness and without extreme varus alignment to avoid greater foot pronation during stance. Cruz et al.[39] found that a hip and trunk muscle strengthening program reduced foot pronation in relaxed stance, only in women with less foot-ankle varus in prone, which is consistent with our results.

The CART selected the foot abduction angle at initial contact as the third predictor of foot pronation during walking, which interacted with varus foot alignment and hip passive stiffness. Participants with foot-ankle varus ≤25.83°, hip passive stiffness >0.09 Nm/rad kg^{-1}, and foot abduction angle >19.58° (node 6) had an 85% increase in the likelihood of having greater foot pronation during walking. A possible explanation for this interaction is that very high hip passive stiffness may increase the external rotation of the entire lower limb and, consequently, the foot abduction angle during walking.[40] This explanation agrees with studies in which voluntary increases in the foot abduction angle when walking resulted in greater foot pronation.[9,10,41] However, our study showed how naturally occurring larger foot abduction angles relate to greater foot pronation. In the corresponding profile, although the participants do not have larger foot-ankle varus and have higher hip stiffness, they tend to have greater foot pronation due to high foot abduction angles. Interventions to reduce hip passive stiffness (e.g., to increase passive internal rotation range of motion)[42] might minimize foot abduction angle and avoid greater foot pronation. However, this is speculative at this point.

The hip external rotators' eccentric strength was not a predictor in the model. However, the hip external rotators' function was assessed by their maximum torque/force production capacity. During tasks such as gait, the demands of torque production by the hip muscles do not reach their maximum capabilities.[43] Therefore, the maximum strength measures may not be related to gait kinematics. Still, the active force and torque produced during gait by the hip external rotators might be associated with the lower limb kinematics, which is suggested by electromyography.[44] However, further investigation, including estimates of the muscles' forces and torques produced during walking, is needed.

Compared to the participants' profiles highlighted in FIG. 3, a minor part of the participants had a foot pronation pattern opposite to the one predicted by the profile in which they were included. For example, 17% of the participants ($n = 1$) in node 6 of the model, with foot-ankle varus ≤25.83°, hip passive stiffness >0.09 Nm/rad kg^{-1}, and foot abduction angle >19.58° had lower foot pronation. Those exceptions may be due to interactions with other factors not investigated in this study. Some not-investigated factors were the activity and strength of intrinsic foot[45] and ankle muscles,[46] the mobility of the first ray,[47] and the passive mechanical resistance of tissues of the shank[48] and midfoot.[8,49] These factors could be included in future investigations.

Study limitations can be pointed out. The 45th and 55th percentiles, used to categorize pronation magnitudes, were not based on clinical criteria. Still, the average values of foot pronation during walking in the participants with lower and greater values were consistent with those reported in previous studies that classified patterns of rearfoot motion[50] or categorized individuals as low and high pronators based on the Foot Posture Index.[51] A clinical limitation of the findings was the isokinetic

Table 2
Prevalence Ratio of Each Terminal Node of the CART Model.

CART Model	Terminal Node	PR (95% CI)
Foot Pronation	2	2.08 (1.37, 3.16)*
	3	1.98 (1.28, 3.08)*
	5	0.31 (0.18, 0.55)*
	6	1.85 (1.13, 3.04)*

CART, Classification and Regression Tree; PR, Prevalence Ratios; CI, Confidence Interval. * Indicates significant PR.

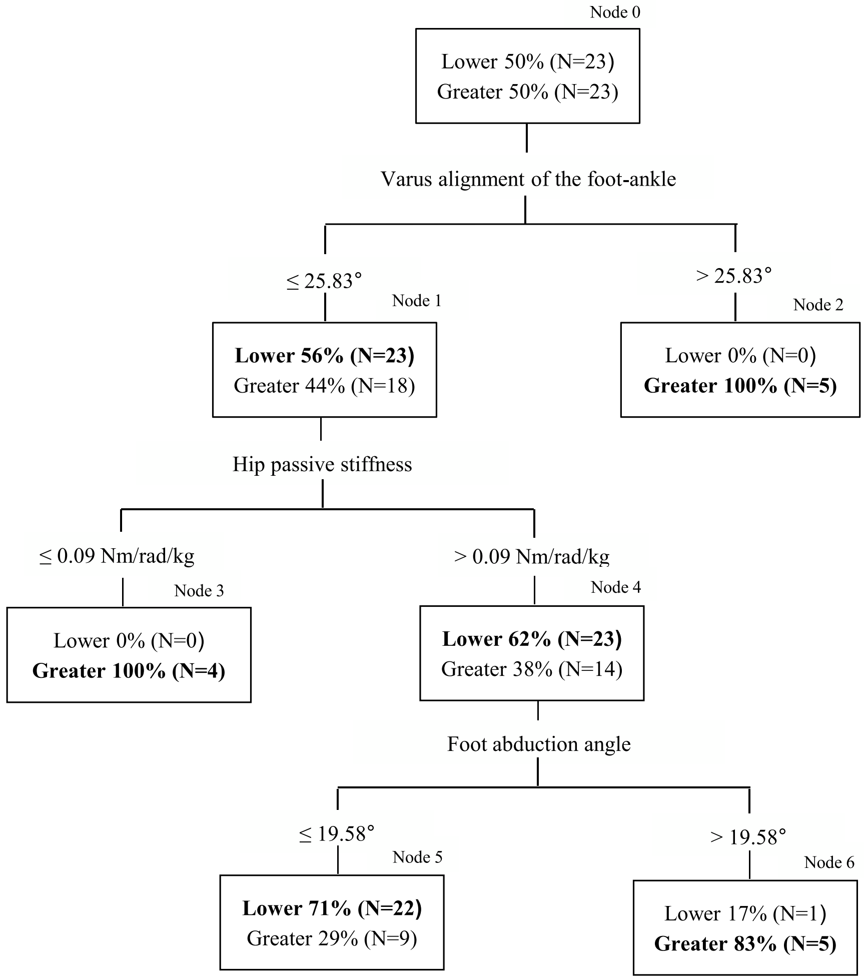

Fig. 2. Classification tree of the lower and greater foot pronation during walking. At each node, there is the prediction and the number of individuals for each category, lower and greater foot pronation during walking. The cutoff values for each predictor are shown between the nodes. The predictor variables are the varus alignment of the foot ankle measured in prone, hip passive stiffness to internal rotation, and foot abduction angle at the initial contact of walking.

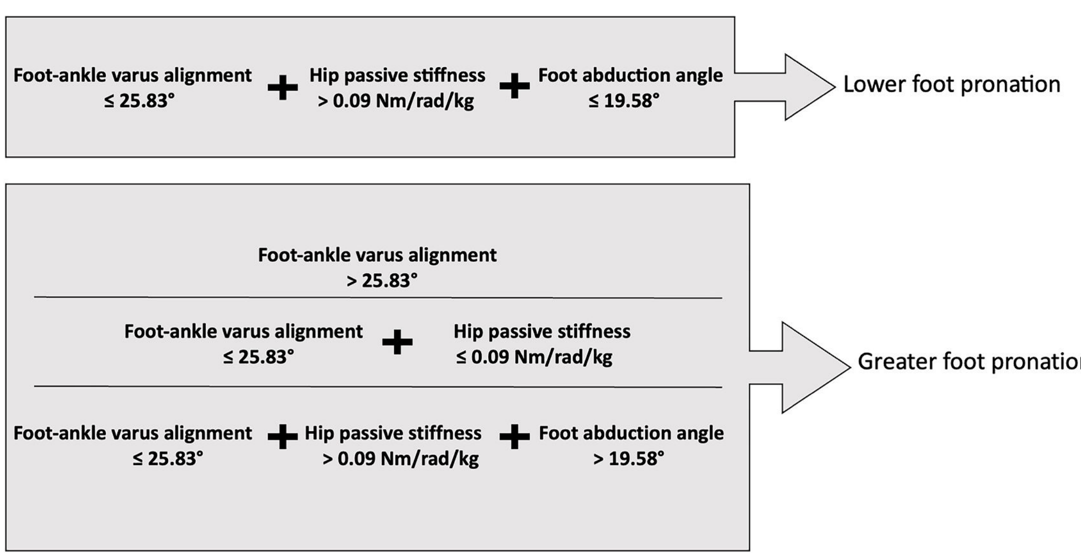

Fig. 3. Profiles of greater and lower foot pronation during walking. The predictor variables are the varus alignment of the foot ankle measured in prone, hip passive stiffness to the internal rotation, and foot abduction angle at the initial contact of walking.

measurement of the hip variables, which limits using the indicated stiffness cutoff point in patients' assessments (when isokinetic dynamometry is unavailable). Still, a corresponding value in the clinical measure of hip stiffness could be estimated in the discussion above, based on a previous study.[36] To our knowledge, a gold-standard evaluation was necessary to include only accurate measures in the first CART model investigating foot pronation. The present results motivate the development of similar studies investigating cutoff points for clinical measurements of hip stiffness and strength.[36,52] Another limitation is that only healthy women participated in this study, which limits the generalization of the results for men. Additionally, patients with musculoskeletal conditions and pain may have different interactions and kinematics. Antalgic movement patterns and tissue adaptation to the presence of pain and altered movement may occur. These adaptations may also limit the generalization of the profiles identified. The results apply to preventive approaches or advanced rehabilitation phases when pain is not present and are a necessary step for future investigations with patients with painful conditions.

In clinical practice, the profiles identified can help indicate variables that should be evaluated in individuals with greater foot pronation during walking. A clinician may verify whether an individual's hip and foot factors are consistent with one of the profiles identified and plan an intervention based on that profile.

Conclusion

Greater and lower magnitudes of foot pronation during walking are related to non-linear interactions between hip and foot biomechanical factors. The profiles revealed the factors related to greater or lower foot pronation during walking. In clinical settings, these profiles help to identify which combinations of biomechanical factors should be searched in individuals with excessive or decreased foot pronation.

Conflicts of interest

The authors declares no conflicts of interest.

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Supplementary materials

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.bjpt.2024.101136.

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