DOI: https://doi.org/10.30795/jfootankle.2026.v20.1995

Original Article

Development and reliability of a device to measure medial longitudinal arch loading in individuals with foot pronation

Alícia Correa Brant[1], Isabela Lurdes Miranda Pereira[1], Fabrício Anício de Magalhães[2], Jéssica Pinheiro dos Santos[1], Larissa Aimée Assunção Alves[1], Ana Clara D'Villa Gonçalves Barbosa[1], Douglas Novaes Bonifácio[1], Emanuel Rodrigues Pinheiro[1], Henrique Silveira Costa[1], Renato Trede[1]

1. Federal University of the Jequitinhonha and Mucuri Valleys - UFVJM, Graduate Program in Rehabilitation and Functional Performance, Physical Therapy Department. Diamantina, MG, Brazil.
2. College of Education, Health, and Human Sciences, Department of Biomechanics, University of Nebraska at Omaha, University Drive South, Omaha, NE, USA.

Abstract

Objective: To develop the pronation loading (PL) device, evaluate its intra- and inter-rater reliability in single-leg stance, and investigate correlations between PL measures and gait kinetic and kinematic variables.

Methods: This cross-sectional study was conducted in two stages: (1) assessment of intra- and inter-rater reliability of the PL device, and (2) investigation of correlations between PL measurements and kinetic and kinematic gait variables. Reliability was analyzed using the intraclass correlation coefficient (ICC), and measurement error was assessed using the standard error of measurement (SEM) and minimal detectable change (MDC). Associations were evaluated using Pearson's correlation coefficient (r).

Results: Intra-rater reliability showed ICC of 0.75 and 0.73, with SEM ranging from 0.04 to 0.05. Inter-rater reliability demonstrated an ICC of 0.77, an SEM of approximately 0.04, and an MDC between 0.11 and 0.13. Pronation loading measurements showed moderate correlations with peak ankle evertor external moment, knee internal rotation moment, hip internal rotation moment, and hip adductor moment during gait.

Conclusion: The PL test demonstrated good reliability and significant associations with selected gait-kinetic variables, suggesting potential clinical applicability for assessing pronation-related loading during orthostatic standing.

Level of evidence III; Cross-sectional (two stages).

Keywords: Foot; Gait; Flatfoot; Kinematics; Pronation.

Introduction

Foot pronation plays a fundamental role in human gait mechanics[1]. Occurring primarily during the first half of stance, it contributes to energy dissipation through the foot structures[2]. As a triplanar motion, pronation involves calcaneal eversion, talar plantarflexion, and adduction, resulting in lowering and medial displacement of the navicular bone[3].

Excessive pronation may overload musculoskeletal structures and increase injury risk[4] and is associated with plantar fasciitis[5], posterior tibial tendinitis, and medial tibial stress syndrome[6]. Due to the oblique orientation of the subtalar joint, calcaneal eversion may be transferred proximally, promoting medial rotation of the leg and thigh[7,8].

Orthotic insoles are commonly used to support the medial longitudinal arch (MLA) and limit excessive pronation[9]. However, excessive correction may restrict the physiological pronation required for energy dissipation after initial contact[10,11]. Variable-density insoles produced by additive

Study performed at the Federal University of the Jequitinhonha and Mucuri Valleys (UFVJM), Diamantina, MG, Brazil.

Correspondence: Renato Guilherme Trede Filho. MGT-367 Highway, Km 583, 5000 - Alto da Jacuba - Diamantina, MG, 39100-000, Brazil. **E-mail:** trede@ufvjm.edu.br. **Conflicts of interest:** none. **Source of funding:** none. **Date received:** March 16, 2026. **Date accepted:** April 02. 2026.

How to cite this article: Brant AL, Pereira ILM, Magalhães FA, Santos JP, Alves LAA, Barbosa ACDG, et al. Development and reliability of a device to measure medial longitudinal arch loading in individuals with foot pronation. J Foot Ankle. 2026;20(1):e1995.

manufacturing have emerged as a strategy to allow controlled pronation while maintaining biomechanical efficiency[12-14].

To optimize orthotic prescription, it is essential to quantify the force applied by the MLA to the ground in standing and determine whether this measurement relates to dynamic gait behavior. Therefore, this study describes the development of a device to assess medial longitudinal arch loading during single-leg stance, evaluates its intra- and inter-rater reliability, and investigates its association with gait kinematic and kinetic variables of the hip, knee, and ankle.

Methods

This cross-sectional study was divided into two stages. The first stage evaluated the intra- and inter-rater reliability of the device named pronation loading (PL), whereas the second stage investigated the association between PL measurements and gait kinetic and kinematic variables.

Participants were recruited from the physiotherapy teaching clinic at the Universidade Federal dos Vales do Jequitinhonha e Mucuri (UFVJM) through posters and social media. All volunteers provided written informed consent before the assessments. The research protocol was approved by the Institutional Review Board under the number CAAE 31685420.9.0000.5108.

Pronation loading

The PL consists of a digital tensile and compression dynamometer (Homis - MOD2100 - H004 - 557) with a load cell fixed inside a wooden box measuring 1 m × 1 m and 15 cm in height. At the top of the box, a piece of wood roughly the shape of the MLA is connected to the load cell by a screw, allowing the measurement of the applied direct force. The collected data, corresponding to the weight load on the MLA support, are displayed on an external digital display, with values presented in kilogram-force (kgf) (Figure 1).

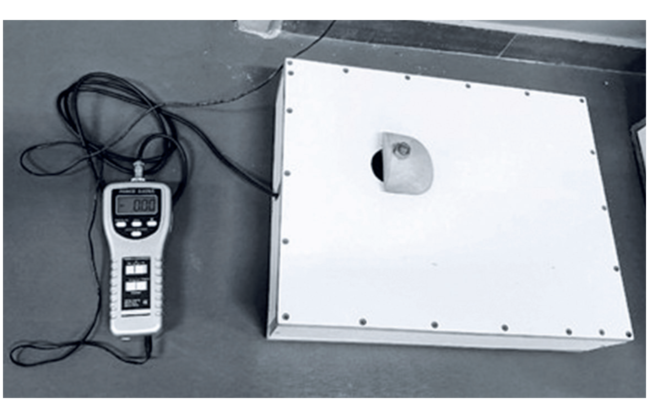

Figure 1. Pronation loading equipment.

Reliability

Participants

Individuals aged 18 to 45 years were included; they had no neurological or orthopedic diseases, had not undergone surgery on their lower limbs or spine, or used orthopedic insoles in the 12 months prior to data collection, and were able to remain in a single-leg stance for 10 minutes. Reports of pain or discomfort during data collection were considered exclusion criteria.

A sample size calculation was performed using G*Power software[15], based on Student's t-test, with statistical power set at 80% and a significance level of 0.05. The first data collection involved 39 volunteers, 10 (25.65%) men and 29 (74.35%) women. During the two days of data collection, we had a sample loss of 14 (35.90%) participants for various reasons, including travel, illness, and nonresponse. Therefore, 25 (64.10%) volunteers completed the evaluation on the second day of data collection, 5 (20%) men and 20 (80%) women, with a mean weight of 62.50 kg.

Procedures

Data collection was performed on the same day, in a laboratory, with the experimental conditions randomly assigned by lottery using an opaque envelope. The volunteers' structure and mass were measured. For the PL measurement, the volunteer was seated with the hip and knee joints flexed at 90 degrees and the ankle in a neutral position.

The foot was positioned on the wooden piece shaped like the MLA. The volunteer was asked to stand upright, extend the knee in a single-leg stance, and transfer all their body weight onto the apparatus for 5 seconds, simulating the midstance phase of gait. To avoid falls or instability, participants were allowed light fingertip support against a wall. This support was standardized across participants and limited to minimal contact to reduce its influence on load distribution. Only the dominant lower limb was assessed to standardize testing conditions and reduce variability associated with inter-limb differences. Although limb dominance may not directly reflect pronation characteristics, this approach was adopted to improve internal consistency.

Each measurement was repeated five times by two independent evaluators, and the mean value was used for analysis. After a seven-day interval, the same evaluators repeated the procedure to calculate the intraclass correlation coefficient (ICC).

Correlation with biomechanical variables

Participants

The sample size calculation for the correlation stage was performed using G*Power software[15], based on Student's t-test, with statistical power set at 80%, a significance level of 0.05, and a moderate effect size derived from a previous study[16].

The sample consisted of adults aged 18 to 45 years with a body mass index (BMI) of less than 30 kg/m^2 and a foot posture index (FPI) score of ≥ +6[17]. Exclusion criteria included a history of fractures or surgical interventions in the lower limbs within the last year, as well as pain or disability during the performance of the tests[15,16].

In total, 18 women (75%) and six men (25%) completed the study. Table 1 presents the demographic characteristics of the participants, indicating a mean age of 22.50 years (SD = 4.25), a mean height of 167.22 cm (SD = 8.65), a mean body mass of 65.36 kg (SD = 10.62), and a mean BMI of 23.33 (SD = 2.76) (Table 1).

Procedures

Initially, the FPI test was performed to determine sample inclusion. Following this, PL was collected, as previously described. To obtain the kinetic and kinematic variables of the ankle, knee, and hip in the coronal and transverse planes during gait, a three-dimensional (3D) motion capture system with nine cameras (Oqus 3+, Qualisys Medical AB, Gothenburg, Sweden) operating at a frequency of 200 Hz was used, synchronized with three FP 4060-08 force platforms (Bertec, Columbus, Ohio, USA) at a frequency of 1000 Hz.

First, the cameras were positioned to cover the entire capture area, and the system was calibrated using a calibration stick to adjust the capture volume, scale, and coordinate origin. Subsequently, the force platforms were configured, leveled, and temporally synchronized with the motion capture system to ensure synchronized data acquisition. Before each session, the platforms were zeroed to eliminate deviations, and the synchronization was tested to ensure data consistency. Finally, anatomical markers were positioned on the participant's body so that the software could calculate biomechanical variables, such as joint angles and ground reaction forces, enabling a precise gait analysis.

Individual calibration involved the precise placement of anatomical markers at specific points on the body, such as joints and bone segments, to allow the capture system to accurately track 3D movement. Passive markers were attached to the lower limbs and pelvis, following the Calibrated Anatomical Systems Technique (CAST) protocol[18,19], as illustrated in Figure 2.

Table 1. Characteristics of the volunteers: correlation of pronation loading with gait variables (n = 24)

Features	Percentage	
Sex	18 women (75%)	6 men (25%)
	Mean (SD)	
Age (years)	22.50 (4.25)	
Height (cm)	167.22 (8.65)	
Mass (kg)	65.36 (10.62)	
BMI (kg/m^2)	23.33 (2.76)	

SD = standard deviation; cm = centimeters; kg = kilograms; BMI = Body Mass Index.

First, anatomical landmarks were identified, and markers were symmetrically placed, avoiding areas of excessive skin movement to minimize relative motion errors. Subsequently, a static test was performed to identify the participant's neutral posture, allowing the software to create a digital model of the body structure. This model was adjusted based on individual parameters such as joint alignment and body segment length. Finally, five strides were collected at a self-selected speed, while the 3D motion analysis system recorded the data.

Kinetic and kinematic data were processed and analyzed using Visual 3D software (C-motion, Inc., Rockville, USA). The marker trajectories and force data from the force platforms were filtered using fourth-order Butterworth low-pass filters, applying cutoff frequencies of 6 Hz for the marker trajectories and 25 Hz for the force data.

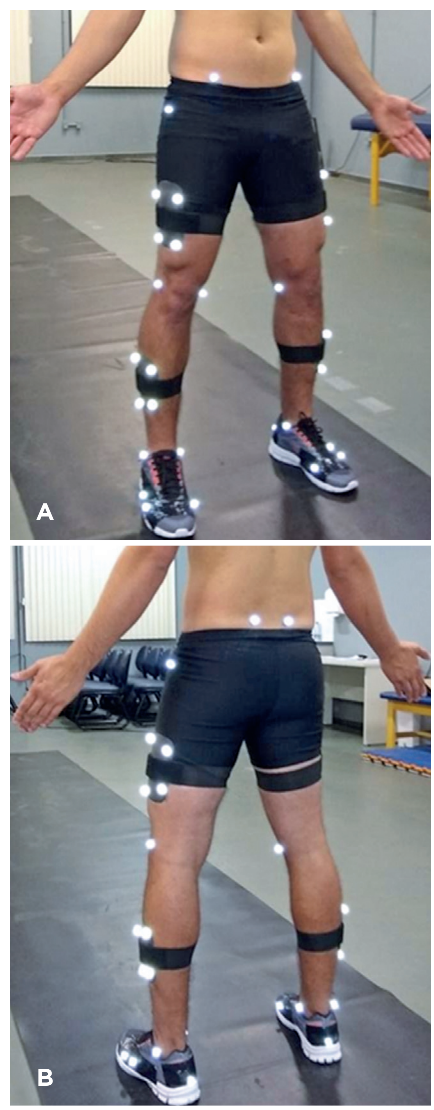

Figure 2. Passive markers of the lower limb segment.

Foot contact with the ground was automatically determined by the software, using the vertical component of the ground reaction force, with a threshold of 20 N. For each participant, the mean values across the five trials were used in the analysis, and all variables were normalized to a maximum of 101 points. The angular displacement of the hindfoot, knee, and hip was calculated in the coronal and transverse planes of the dominant lower limb.

Based on ground reaction force data, the inverse dynamics technique was applied to calculate the external force moments of the ankle, knee, and hip in the coronal and transverse planes. These moments were normalized to the volunteers' body mass and reported in units of Nm/kg to ensure comparability across participants.

Data analysis

To verify inter- and intra-examiner reliability, the ICC was applied. In addition, the standard error of measurement (SEM) was used to assess measurement accuracy, and the minimal detectable change (MDC) was used to determine the smallest detectable change.

Pearson's correlation coefficient (r) was used to assess the relationship between the PL measurements and kinematic and kinetic gait variables from the 3D motion analysis system. All statistical analyses were performed with a significance level of $\alpha = 0.05$.

Results

Pronation loading reliability

In the test-retest analysis, PL demonstrated satisfactory intra-rater reliability. The intra-rater ICC was 0.75 (95% CI: 0.60-0.84) in the first collection and 0.73 (95% CI: 0.58-0.83) in the second collection. The SEM was 0.04 (95% CI: -0.004-0.13) and 0.05 (95% CI: -0.04-0.14) on the respective days. These data are presented in Table 2.

Regarding examiner agreement, the instrument demonstrated good inter-rater reliability. The inter-rater ICC was 0.77 (95% CI: 0.63-0.85), with SEM of 0.04 (95% CI: -0.04-0.13) on the first day, and 0.77 (95% CI: 0.65-0.86) and SEM of 0.04 (95% CI: -0.04-0.12) on the second data collection. The variation in MDC was low[20], ranging from 0.11 to 0.13.

Correlation between pronation loading measurements and gait variables

The normalized mean PL value showed a weak positive correlation with peak ankle evertor external moment (r = 0.450; p = 0.027) and peak knee internal rotation moment (r = 0.427; p = 0.037). Additionally, significant correlations were observed with peak hip internal rotation (r = 0.586; p = 0.003), peak hip internal rotation moment (r = 0.583; p = 0.003), and peak hip adductor moment (r = -0.427; p = 0.039) (Table 3).

We found no correlation between PL and the variables peak ankle eversion (r = -0.209; p = 0.326); peak ankle abduction (r = -0.273; p = 0.196); peak ankle external adductor moment (r = -0.035; p = 0.872); peak knee adductor moment (r = -0.217; p = 0.309); peak hip adduction (r = -0.117) (p = 0.586); peak hip adductor moment (r = 0.050; p = 0.818) and peak hip external rotation moment (r = 0.137; p = 0.524).

Discussion

The aim of this study was to develop the PL device to quantitatively assess MLA loading, evaluate its intra- and inter-rater reliability, and investigate its association with gait variables in individuals with pronated feet.

Pronation loading reliability

Static tests, such as the PL, navicular drift test, and navicular drop test, can be predictive of MLA height during the mid-stance phase of gait. These weight-bearing tests reflect the inferior displacement of the navicular bone and MLA. In contrast, the supination resistance test (SRT), which involves elevating the navicular bone, can be performed either manually or with the supination resistance test machine (SRTM)[21]. This test is frequently used in clinical and research settings to estimate the force required to supinate the foot. All of these tests are easy to administer and low-cost, making them practical tools for clinical use.

This study investigated the reliability of PL and found good intra- and inter-rater reliability. The intra-rater ICCs were 0.75 and 0.73, and the inter-rater ICCs were 0.77 on both data collection days. Similar data were reported by Kirmizi et al.[22], who reported intra-rater ICC values of 0.93 and 0.97 for the navicular drop test, 0.72 and 0.85 for the navicular drift test, and 0.85 and 0.87 for the static and dynamic arc index,

Table 2. Test-retest measures of pronation loading (n = 25)

	Mean (SD)	ICC (95%CI)	SEM (95%CI)	MDC (95%CI)
Intra-rater 1	0.17 (0.07)	0.75 (0.60-0.84)	0.04 (-0.04-0.13)	0.12 (0.00-0.48)
Intra-rater 2	0.17 (0.07)	0.73 (0.58-0.83)	0.05 (-0.04-0.14)	0.13 (0.01-0.51)
Inter-rater 1	0.17 (0.07)	0.77 (0.63-0.85)	0.04 (-0.04-0.13)	0.12 (0.00-0.49)
Inter-rater 2	0.17 (0.07)	0.77 (0.65-0.86)	0.04 (-0.04-0.12)	0.11 (0.00-0.45)

Mean: refers to the combined mean values of days 1 and 2 for intra-rater or the combined mean values of examiners 1 and 2 for inter-rater (kgf). SD: Standard Deviation. 95% CI: 95% confidence interval. ICC: Intraclass Correlation Coefficient. SEM: Standard Error of Measurement. MDC: Minimal Detectable Change in the 95% CI. Intra-rater: measurements for the same examiner on two different days. Inter-rater: measurements for two different examiners on the same day.

Table 3. Pearson correlation of the normalized mean of pronation loading with biomechanical gait variables. (n = 24)

Gait variables	Pearson correlation	Normalized mean pronation loading
Ankle eversion peak	Pearson correlation	-0.209
	sign (2 limbs)	0.326
Peak external moment of ankle evertor	Pearson correlation	**0.450***
	sign (2 limbs)	0.027
Peak ankle abduction*	Pearson correlation	-0.273
	sign (2 limbs)	0.196
Peak external moment of ankle abductor*	Pearson correlation	-0.035
	sign (2 limbs)	0.872
Knee adductor peak	Pearson correlation	-0.217
	sign (2 limbs)	0.309
Peak moment of internal rotation of the knee	Pearson correlation	**0.427***
	sign (2 limbs)	0.037
Peak hip adduction	Pearson correlation	-0.117
	sign (2 limbs)	0.586
Peak hip adductor moment	Pearson correlation	0.050
	sign (2 limbs)	0.818
Peak hip adductor moment	Pearson correlation	**-0.425***
	sign (2 limbs)	0.039
Peak internal rotation of the hip	Pearson correlation	**0.586****
	sign (2 limbs)	0.003
Peak moment of external hip rotation	Pearson correlation	0.137
	sign (2 limbs)	0.524
Peak moment of internal hip rotation	Pearson correlation	**0.583****
	sign (2 limbs)	0.003

**The correlation is significant at 0.01 level (2 extremes). *The correlation is significant at 0.05 level (2 extremes).

respectively. Furthermore, the SRTM showed an intra-rater ICC of 0.89 and an inter-rater ICC of 0.78[21].

Although the same individual may exhibit high intra-subject variability in repeated measurements[23], in this study, the SEM ranged from 0.04 to 0.05, while the MDC ranged from 0.11 to 0.13. These results indicate low variation across repeated measurements, suggesting that PL is a reliable method for assessing forces related to foot pronation. In comparison, the SRT presented SEM values ranging from 7.3 to 9.0 and MDC values between 20.4 and 24.9[24], demonstrating greater measurement variability.

Correlation of pronation loading with 3D gait analysis

In our sample of individuals with flat feet, PL was positively correlated with peak ankle evertor external moment (r = 0.480), peak knee internal rotation moment (r = 0.427), peak hip internal rotation (r = 0.586), and peak hip internal rotation moment (r = 0.583). A negative correlation was also identified with the peak hip adductor moment (r = -0.427).

The positive correlation between PL and peak ankle evertor moment suggests that increased foot pronation is associated with greater lowering of the MLA, resulting in overload on structures responsible for ankle eversion. Previous studies indicate that increased rearfoot eversion during gait is characteristic of individuals with flat feet[25,26] and is associated with reduced MLA height[27]. Changes in the elastic behavior of the plantar fascia may contribute to gait abnormalities in these individuals[25] because increased MLA flexibility reduces the foot's ability to function as a rigid lever during push-off. This change results in reduced support for the gastrocnemius and soleus muscles, compromising gait effectiveness[28].

Excessive hindfoot pronation also influences internal tibial rotation, causing passive stress and altering the direction of forces on the patellofemoral joint[29]. A positive correlation was observed between PL and peak internal knee rotation, suggesting greater compression in medial knee structures and a higher propensity for patellofemoral pain syndrome. One proposed mechanism is that excessive foot pronation leads to internal rotation of the tibia and femur, resulting in a reduced contact area at the patellofemoral joint[30]. Excessive subtalar pronation has historically been identified as a dysfunctional biomechanical factor in the lower limbs, leading to adaptations in other joints[31]. Although conflicting evidence exists regarding the relationship between patello-femoral pain syndrome and ankle and foot deformities, such biomechanical changes must be carefully analyzed[32].

Finally, the coupled movement between the hindfoot and hip joints has already been reported in a previous study[33]. In the present study, we identified a positive correlation between PL and peak internal hip rotation, and a negative correlation with peak hip adductor moment. Reduced hip adductor torque, together with increased hip and knee internal rotation[34], reinforces the association between foot pronation and proximal rotational mechanics during walking. During the stance phase, increased hip adduction can be a strategy to elevate the MLA by shifting the load to the lateral side of the foot, thereby reducing the compressive force on the MLA[35]. The hip internal rotation moment demonstrated a biphasic pattern during the stance phase, with a first peak in early stance and a second peak in late stance. This pattern is consistent with previously reported normative gait data[36].

This study had several limitations. The PL showed substantial variation over the 5-second recording period, suggesting that a longer time may be needed for dynamometer stabilization. In addition, the potential influence of external support on weight transfer during testing cannot be entirely ruled out. Another important limitation is that the wooden support arch may not fully adapt to different foot morphologies and sizes, potentially influencing the measured load and affecting construct validity. Future designs should consider adjustable or customizable interfaces to improve anatomical conformity. Finally, the sample included a greater proportion of women than men. Despite these limitations, the device appears to be a reliable, low-cost, and easy-to-use tool with potential for clinical application.

Conclusion

The pronation loading test demonstrated good intra- and inter-rater reliability for assessing pronation-related loading in standing. In addition, PL measurements were significantly associated with selected gait kinetic variables, suggesting potential clinical applicability. However, further studies are needed to investigate additional measurement properties and to refine the device for different foot morphologies.

Acknowledgments

The authors gratefully acknowledge the support provided by the Financier of Studies and Projects (FINEP) and SisAssistiva-MCTI for the development of this research.

Authors' contributions: Each author contributed individually and significantly to the development of this article: CA *(https://orcid.org/0000-0003-3670-3616) conceived and planned the activities that led to the study, data collection, participated in the review process; LI *(https://orcid.org/0009-0004-1567-8639) statistical analysis, interpreted the results of the study, participated in the review process; AF *(https://orcid.org/0000-0001-9318-8492) performed the surgeries, clinical examination, approved the final version; PJ *(https://orcid.org/0009-0008-6154-6273) data collection, survey of the medical records, participated in the review process; AL *(https://orcid.org/0009-0005-6913-3789) bibliographic review, formatting of the article, participated in the review process; DA *(https://orcid.org/0009-0006-2925-5201) conceived and planned the activities that led to the study, statistical analysis, formatting of the article, approved the final version; ND * (https://orcid.org/0009-0009-7051-7703) clinical examination, data collection, participated in the review process; PH * (https://orcid.org/0009-0003-8759-4218) interpreted the results of the study, bibliographic review, participated in the review process; SH * (https://orcid.org/0000-0002-1426-7246) survey of the medical records, data collection, approved the final version; TR * (https://orcid.org/0000-0001-6118-1181) conceived and planned the activities that led to the study, interpreted the results of the study, participated in the review process, approved the final version. All authors read and approved the final manuscript. *ORCID (Open Researcher and Contributor ID) iD.

References

1. Mousavi SH, Khorramroo F, Jafarnezhadgero A. Gait retraining targeting foot pronation: A systematic review and meta-analysis. PLoS One. 2024;19(3):e0298646.
2. McClay I, Manal K. A comparison of three-dimensional lower extremity kinematics during running between excessive pronators and normals. Clin Biomech (Bristol). 1998;13(3):195-203.
3. Yang S, Canton SP, Hogan MV, Anderst W. Healthy ankle and hindfoot kinematics during gait: Sex differences, asymmetry and coupled motion revealed through dynamic biplane radiography. J Biomech. 2021;116:110220.
4. Kubo T, Uritani D, Ogaya S, Kita S, Fukumoto T, Fujii T, et al. Association between foot posture and tibiofemoral contact forces during barefoot walking in patients with knee osteoarthritis. BMC Musculoskelet Disord. 2022;23(1):660.
5. Hamstra-Wright KL, Huxel Bliven KC, Bay RC, Aydemir B. Risk Factors for Plantar Fasciitis in Physically Active Individuals: A Systematic Review and Meta-analysis. Sports Health. 2021; 13(3):296-303.
6. Menéndez C, Batalla L, Prieto A, Rodríguez MÁ, Crespo I, Olmedillas H. Medial Tibial Stress Syndrome in Novice and Recreational Runners: A Systematic Review. Int J Environ Res Public Health. 2020;17(20):7457.
7. Hintermann B, Nigg BM. Pronation in runners. Implications for injuries. Sports Med. 1998;26(3):169-76.
8. Buldt AK, Murley GS, Butterworth P, Levinger P, Menz HB, Landorf KB. The relationship between foot posture and lower limb kinematics during walking: A systematic review. Gait Posture. 2013;38(3):363-72.
9. Chen H, Sun D, Fang Y, Gao S, Zhang Q, Bíró I, et al. Effect of orthopedic insoles on lower limb motion kinematics and kinetics in adults with flat foot: a systematic review. Front Bioeng Biotechnol. 2024;12:1435554.
10. Maharaj JN, Cresswell AG, Lichtwark GA. Subtalar Joint Pronation and Energy Absorption Requirements During Walking are Related to Tibialis Posterior Tendinous Tissue Strain. Sci Rep. 2017;7:17958.
11. Costa BL, Magalhães FA, Araújo VL, Richards J, Vieira FM, Souza TR, et al. Is there a dose-response of medial wedge insoles on lower limb biomechanics in people with pronated feet during walking and running? Gait Posture. 2021;90:190-6.
12. Kelly JL, Valier AR. The Use of Orthotic Insoles to Prevent Lower Limb Overuse Injuries: A Critically Appraised Topic. J Sport Rehabil. 2018;27(6):591-5.
13. Choo YJ, Boudier-Revéret M, Chang MC. 3D printing technology applied to orthosis manufacturing: narrative review. Ann Palliat Med. 2020;9(6):4262-70.
14. Hu CW, Nguyen CT, Hölbling D, Pang TY, Baca A, Dabnichki P. A novel 3D printed personalised insole for improvement of flat foot arch compression and recoil – preliminary study. Proc Inst Mech Eng H. 2022; 237(2):329-42.
15. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behav Res Methods. 2007;39(2):175-91.
16. Terwee CB, Bot SD, de Boer MR, van der Windt DA, Knol DL, Dekker J, et al. Quality criteria were proposed for measurement properties of health status questionnaires. J Clin Epidemiol. 2007;60(1):34-42.
17. Redmond AC, Crane YZ, Menz HB: Normative values for the Foot Posture Index. J Foot Ankle Res. 2008;1:6.
18. Leardini A, Benedetti MG, Berti L, Bettinelli D, Nativo R, Giannini S. Rear-foot, mid-foot and fore-foot motion during the stance phase of gait. Gait Posture. 2007;25(3):453-62.
19. Schache AG, Baker R, Lamoreux LW. Influence of thigh cluster configuration on the estimation of hip axial rotation. Gait Posture. 2008;27(1):60-9.

20. Sedaghat AR. Understanding the Minimal Clinically Important Difference (MCID) of Patient-Reported Outcome Measures. Otolaryngol Head Neck Surg. 2019;161(4):551-60.
21. Griffiths IB, McEwan IM. Reliability of a new supination resistance measurement device and validation of the manual supination resistance test. J Am Podiatr Med Assoc. 2012;102(4):278-89.
22. Kirmizi M, Cakiroglu MA, Elvan A, Simsek IE, Angin S. Reliability of Different Clinical Techniques for Assessing Foot Posture. J Manipulative Physiol Ther. 2020;43(9):901-8.
23. Samson M, Crowe A. Intra-subject inconsistencies in quantitative assessments of body sway. Gait Posture. 1996;4(3):252-7.
24. Moisan G, McBride S, Isabelle PL, Chicoine D. The Keystone device as a clinical tool for measuring the supination resistance of the foot: A reliability study. Musculoskeletal Care. 2022;20(3):570-6.
25. Jiao X, Hu T, Li Y, Wang B, Acquah MEE, Wang Z, et al. Association between Elastic Modulus of Foot Soft Tissues and Gait Characteristics in Young Individuals with Flatfoot. Bioengineering (Basel). 2024;11(7):728.
26. Vijittrakarnrung C, Mongkolpichayaruk A, Limroongreungrat W, Chuckpaiwong B. Comparison of Foot Kinematics Between Normal Arch and Flexible Flatfoot Using the Oxford Foot Model: A Matched Case-Control Study. Foot Ankle Orthop. 2024;9(1):24730114241231245.
27. Levinger P, Murley GS, Barton CJ, Cotchett MP, McSweeney SR, Menz HB. A comparison of foot kinematics in people with normal- and flat-arched feet using the Oxford Foot Model. Gait Posture. 2010;32(4):519-23.
28. Wang Y, Qi Y, Ma B, Wu H, Wang Y, Wei B, et al. Three-dimensional gait analysis of orthopaedic common foot and ankle joint diseases. Front Bioeng Biotechnol. 2024;12:1303035.
29. Tan JM, Crossley KM, Munteanu SE, Collins NJ, Hart HF, Donnar JW, et al. Associations of foot and ankle characteristics with knee symptoms and function in individuals with patellofemoral osteoarthritis. J Foot Ankle Res. 2020;13(1):57.
30. Powers CM, Witvrouw E, Davis IS, Crossley KM. Evidence-based framework for a pathomechanical model of patellofemoral pain: 2017 patellofemoral pain consensus statement from the 4th International Patellofemoral Pain Research Retreat, Manchester, UK: part 3. Br J Sports Med. 2017;51(24):1713-23.
31. Tiberio D. The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. J Orthop Sports Phys Ther. 1987;9(4):160-5.
32. Martinelli N, Bergamini AN, Burssens A, Toschi F, Kerkhoffs GMMJ, Victor J, et al. Does the Foot and Ankle Alignment Impact the Patellofemoral Pain Syndrome? A Systematic Review and Meta-Analysis. J Clin Med. 2022;11(8):2245.
33. Koshino Y, Yamanaka M, Ezawa Y, Okunuki T, Ishida T, Samukawa M, et al. Coupling motion between rearfoot and hip and knee joints during walking and single-leg landing. J Electromyogr Kinesiol. 2017;37:75-83.
34. Butowicz CM, Krupenevich RL, Acasio JC, Dearth CL, Hendershot BD. Relationships between mediolateral trunk-pelvic motion, hip strength, and knee joint moments during gait among persons with lower limb amputation. Clin Biomech (Bristol). 2020;71:160-6.
35. Okamura K, Nagamune N, Fukuda K, Kanai S. Classification of the foot kinematics during gait and the characteristics of the knee and hip kinematics in individuals with pronated foot. J Biomech. 2024;173:112258.
36. Perry J, Burnfield JM. Gait Analysis: Normal and Pathological Function. 2nd ed. Thorofare (NJ): SLACK Incorporated; 2010.