

Central Lancashire Online Knowledge (CLoK)

Title	Clinical and biomechanical characteristics of responders and non-responders to insoles in individuals with excessive foot pronation during walking
Type	Article
URL	https://clock.uclan.ac.uk/id/eprint/51718/
DOI	https://doi.org/10.1016/j.jbiomech.2024.112182
Date	2024
Citation	Magalhães, Fabrício A., Souza, Thales R., Trede, Renato, Araújo, Vanessa L., Teixeira, João Pedro M. P., Richards, James and Fonseca, Sérgio T. (2024) Clinical and biomechanical characteristics of responders and non-responders to insoles in individuals with excessive foot pronation during walking. <i>Journal of Biomechanics</i> , 171. ISSN 0021-9290
Creators	Magalhães, Fabrício A., Souza, Thales R., Trede, Renato, Araújo, Vanessa L., Teixeira, João Pedro M. P., Richards, James and Fonseca, Sérgio T.

It is advisable to refer to the publisher's version if you intend to cite from the work.
<https://doi.org/10.1016/j.jbiomech.2024.112182>

For information about Research at UCLan please go to <http://www.uclan.ac.uk/research/>

All outputs in CLoK are protected by Intellectual Property Rights law, including Copyright law. Copyright, IPR and Moral Rights for the works on this site are retained by the individual authors and/or other copyright owners. Terms and conditions for use of this material are defined in the <http://clock.uclan.ac.uk/policies/>

Clinical and biomechanical characteristics of responders and non-responders to insoles in individuals with excessive foot pronation during walking

Fabício A. Magalhães^{1,2}, Thales R. Souza¹, Renato Trede³, Vanessa L. Araújo¹, João Pedro M. P. Teixeira¹, Jim Richards⁴, Sérgio T. Fonseca¹

¹Graduate Program in Rehabilitation Sciences, Department of Physical Therapy, Universidade Federal de Minas Gerais (UFMG), Belo Horizonte, Minas Gerais, Brazil

²College of Education, Health, and Human Sciences, Department of Biomechanics, University of Nebraska at Omaha, Omaha, NE, USA

³Graduate Program in Rehabilitation and Functional Performance, Department of Physical Therapy, Universidade Federal dos Vales do Jequitinhonha e Mucuri (UFVJM), Diamantina, Minas Gerais, Brazil.

⁴Allied Health Research Unit, University of Central Lancashire (UCLan), Preston, UK

* Corresponding author: Graduate Program in Rehabilitation Sciences, Universidade Federal de Minas Gerais, 6627 Antônio Carlos Avenue, Belo Horizonte, MG 31270-901, Brazil

E-mail address: sergioteixeirafonseca@gmail.com (S.T. Fonseca).

Statement

I confirm that all authors were fully involved in the study and preparation of the manuscript, and the material within has not been. It will not be submitted for publication elsewhere.

Abstract

This study aimed to identify the clinical and biomechanical factors of subjects with excessive foot pronation who are not responsive (i.e., “non-responders”) to medially wedged insoles to increase knee adduction external moment. Ankle dorsiflexion range of motion, forefoot-shank alignment, passive hip stiffness, and midfoot passive resistance of 25 adults with excessive bilateral pronation were measured. Also, lower-limb angles and external moments were computed during walking with the participants using control (flat surface) and intervention insoles (arch support and 6° medial heel wedge). A comparison between “responders” ($n = 34$) and “non-responders” ($n = 11$) was conducted using discrete and continuous analyses. Compared with the responders, the non-responders had smaller forefoot varus ($p = 0.014$), larger midfoot passive internal torque peak ($p = 0.005$), and stiffness measured by the torsimeter ($p = 0.022$). During walking, non-responders had lower angle peaks for forefoot eversion ($p = 0.001$), external forefoot rotation ($p = 0.037$), rearfoot eversion ($p = 0.022$), knee adduction ($p = 0.045$), and external hip rotation ($p = 0.022$) and higher hip internal rotation angle peak ($p = 0.026$). Participants with small forefoot varus alignment, large midfoot passive internal torque, stiffness, small knee valgus, hip rotated internally, and foot-toed-in during walking did not modify the external knee adduction moment (“non-responders”). Clinicians are advised to interpret these findings with caution when considering the prescription of insoles. Further investigation is warranted to fully comprehend the response to insole interventions among individuals with specific pathologies, such as patellofemoral pain and knee osteoarthritis (OA).

Keywords: foot orthoses, foot pronation, gait, knee adduction moment, clinical measurements

Introduction

Excessive foot pronation alters the kinematics and kinetics of the lower limb during weight-bearing activities, promoting changes such as excessive shank, knee and thigh internal rotation, and dynamic knee valgus (Cheung et al., 2011; Chuter and Janse de Jonge, 2012; Farahpour et al., 2018a). These altered biomechanics are related to some pathological conditions at the knee, such as patellofemoral pain (Neal et al., 2014) and osteoarthritis (Jani et al., 2012; van Tunen et al., 2018). One risk factor for knee osteoarthritis is a high external knee adduction moment (KAM) associated with excessive foot pronation (Sawada et al., 2016a). Excessive foot pronation may increase the external KAM (Farahpour et al., 2018b), thus overloading the knee's lateral compartment (Wang et al., 2021a).

External KAM has been used as a biomechanical marker to evaluate the risk of knee osteoarthritis development and progression (Andriacchi, 2013; Hinman et al., 2008). Increased or decreased external KAM may overload the medial or lateral compartments of the knee, respectively (Andriacchi, 2013). For instance, individuals with medial knee osteoarthritis demonstrate an increased likelihood of a higher external knee adduction moment (KAM) (van Tunen et al., 2018). Consequently, interventions to modify this altered KAM may benefit the patient (Wang et al., 2021b), particularly as conservative treatments designed to restrict foot motion have been shown to influence the external KAM and potentially alleviate knee pain (Mendes et al., 2020; Rodrigues et al., 2013). However, further investigation is required to fully ascertain the extent to which these interventions targeting KAM modification can effectively benefit patients.

Orthopedic insoles may be an effective clinical conservative intervention that can modify some kinetic parameters of the ankle and knee joints (Jafarnejhadgero et al., 2017; Jones et al., 2014; Sawada et al., 2016b; Ulrich et al., 2020). However, a

systematic review revealed that orthopedic insoles do not improve pain and functionality in patients with knee osteoarthritis (Yu et al., 2021). Moreover, recent studies showed that the biomechanical effects of the insoles were minimal (Jafarnejadgero et al., 2018) or deleterious in some individuals (Jones et al., 2014; Kim et al., 2018). The musculoskeletal characteristics of those subjects may partially explain this minimal or adverse effect (Bonifacio et al., 2018; Ohi et al., 2017), showing the necessity of subgroup analyses. For example, Kim et al. (2018) showed that individuals with a lower external hip rotation range of motion and higher internal/external rotation ratio of the hip rotators strength did not display the expected decreases in external KAM with a specific shoe (laterally wedged). This result indicates that the effectiveness of an intervention to produce clinically desired mechanical changes may depend on the individual's musculoskeletal characteristics (Chapman et al., 2015; Kim et al., 2018). However, the efficacy of rearfoot wedges on the external KAM has been limited to the effects of lateral wedges.

Medially wedged insoles, customarily designed to decrease foot pronation, have been shown to increase the external KAM (Bonifacio et al., 2018). Foot pronation is associated with an increased knee valgus alignment (Ohi et al., 2017), overloading the lateral compartment of the knee (Felson et al., 2013). Thus, an increased KAM may be a desirable protective effect against lateral knee OA. Likewise, as observed for the laterally wedged insoles (Chapman et al., 2015; Jones et al., 2014; Kim et al., 2018), the biomechanical effects of laterally wedged insoles may vary according to the patient's clinical characteristics. As a result, different individuals may exhibit greater or less responsiveness to such insoles.

Therefore, the present study aimed to investigate (a) whether some subjects with excessive foot pronation are not responsive to using medially wedged insoles and (b) to identify clinical and biomechanical factors that may help explain this possible lack of

response. The present exploratory study investigated only young asymptomatic adults with excessively pronated feet, so no specific hypothesis is proposed a priori.

Methods

Twenty-five (17 females and 8 males) participated in the present study and signed an Informed Consent Form before data collection. The University Ethical Research Committee approved this study (CAAE: 50164515.7.0000.5149). The inclusion criteria were age between 18 and 50 years, body mass index (BMI) $< 30 \text{ Kg} \times \text{m}^{-2}$, Foot Posture Index (FPI) $\geq +6$ (bilaterally)(Redmond et al., 2006), footwear size between women's 6 and men's 11 (US standard), no neurological or orthopedic conditions, no history of lower limb or back surgery, and no use of foot orthotics during the last year. These criteria were determined to homogenize the subjects for the clinical measurements described to avoid confounding factors during the analysis. Table 1 presents participants' age, height, mass, BMI, and FPI. During data collection, reporting any pain or discomfort was an exclusion criterion. A sample size of 24 participants was calculated considering an effect size (d) of 0.6, statistical power of 80%, and significance level of 0.05 for single t-tests (difference between means and two tails) using the software G*Power version 3.1 (Faul et al., 2007).

Clinical measures were obtained to identify musculoskeletal factors that may influence the response to intervention. Because of the known associations between foot pronation and lower limb movement patterns, we measured ankle dorsiflexion range of motion (Fig. 1A), forefoot-shank alignment (Fig. 1B), passive hip stiffness (Fig. 1C), and midfoot joint complex passive resistance (Fig. 1D). Three valid trials of each test were acquired and transferred to the web-based application REDcap (Harris et al., 2019), where the mean values were computed for all clinical measures.

Detailed information about the clinical measures can be found in the Supplementary Material. The FPI score was obtained to certify inclusion criteria and performed according to Horwood and Chockalingam (2017). Only subjects with FPI > 5 in both feet participated. The weight-bearing lunge test measured the ankle dorsiflexion range of motion (ROM) (Dill et al., 2014; Kang and Oh, 2017). The higher the shank angle, the higher the ankle dorsiflexion ROM. The forefoot-shank alignment test was used to measure the angle between the forefoot and shank, referred to as the varus angle (Araujo et al., 2020; Cruz et al., 2019); the greater the angle, the greater the forefoot varus. The hip passive internal rotation mobility test assessed the passive hip stiffness (Cardoso et al., 2020; Fajardo et al., 2021), with greater internal rotation ROM indicating a lower hip stiffness. Finally, the passive torque of the midfoot joint complex (MFJC) was assessed using the Foot Torsimeter (Magalhaes et al., 2020b), and the mean passive stiffness ($\text{Nm} \times \text{deg}^{-1} \times \text{kg}^{-1}$) between 20° and 25° of inversion (MFJC inversion mean stiffness) was calculated from the average of the instantaneous slope of the internal passive torque-angle time series using the 4th-order polynomial method (Magalhaes et al., 2021; Magalhaes et al., 2020b).

A		Test: weight-bearing lunge Measure: ankle passive dorsiflexion range of motion ($^{\circ}$) measured by the shank inclination when the knee touched the wall with the foot as far as possible from the wall.
B		Test: forefoot-shank alignment Measure: forefoot varus alignment ($^{\circ}$) measured by the angle between the rod under the metatarsal heads and the shank's bisection.
C		Test: hip passive internal rotation mobility Measure: passive hip stiffness ($^{\circ}/\text{kg}$) measured by the hip internal rotation up to the first passive resistance normalized by the body mass.
D		Test: foot torsimeter Measures: • Peak torque of the midfoot joint complex (Nm/kg) during inversion, measured at 45° of inversion and normalized by body mass. • Mean stiffness of the midfoot joint complex ($\text{Nm}/^{\circ}/\text{kg}$) during inversion, measured between 20° and 25° of inversion and normalized by body mass.

Figure 1: Musculoskeletal assessment tests and measurements description

Kinematic and kinetic analysis was conducted during standing and walking. Fifty-seven spherical 12-mm retro-reflective markers were attached to the pelvis, thighs, shanks, and feet. One trial was recorded in the standing position to define the segments'

coordinate systems with the subject using the insoles before each condition (Fig. 2A and 2B). Each participant walked at a self-selected speed on a split-belt instrumented treadmill (Bertec Corp., 120Hz, USA) synchronized with a 9-camera optoelectronic system (Oqus 5+, 120Hz, Qualisys, Sweden). Participants determined their self-selected speed during the familiarization trial by indicating to the treadmill operator whether the current speed felt faster or slower than their natural pace. Speed adjustments were made until they were confident that the selected speed closely matched their natural walking pace. After five minutes of familiarization with each pair of insoles on the treadmill, several walking trials were conducted over a period of 90 seconds. All trajectories from all markers and the ground reaction forces (GRF) were identified for each trial, and the participants stepped the right and left feet on the right and left force plates, respectively.

Segments		Markers	
Name	Definition	Location	Type
Pelvis	Modeled as a cylinder defined by the anterior and posterior superior iliac spines	Left / Right Anterior Superior Iliac Spine	A / T
		Left / Right Posterior Superior Iliac Spine	A / T
		Central Posterior Iliac Spine	T
Thigh	Modeled as a cone defined proximally by the midpoint between the hip center and greater trochanter and distally by the lateral and medial epicondyles	Left / Right Great Trochanter	A
		Left / Right Lateral Femoral epicondyle	A
		Left / Right Medial Femoral epicondyle	A
		Left / Right Thigh Cluster of Markers	T
Shank	Modeled as a cone defined proximally by the midpoint between the lateral and medial epicondyles and distally by the lateral and medial malleoli	Left / Right Lateral Malleolus	A
		Left / Right Medial Malleolus	A
		Left / Right Shank Cluster of Markers	T
Rearfoot	Modeled as a cone defined proximally by the calcaneus apex and superior aspect and distally by the midpoint between the peroneal tuberculous and sustentaculum talus	Left / Right Peroneal Tuberculous	A / T
		Left / Right Sustentaculum Talus	A / T
		Left / Right Calcaneus (superior aspect)	A
		Left / Right Calcaneus (apex)	A / T
Forefoot	Modeled as a cylinder defined proximally by the midpoint between the 1 st and 5 th metatarsal bases and distally by the midpoint between the 1 st and 5 th metatarsal heads	Left / Right Metatarsal Heads (1 st and 5 th)	A
		Left / Right Metatarsal Heads (2 nd and 4 th)	T
		Left / Right metatarsal Bases (1 st and 5 th)	A
		Left / Right metatarsal Base (2 nd)	T



Figure 2: Top: segment definition and marker description. Bottom: anatomical and tracking markers' positions (A and B). Marker type: anatomical (A) and tracking (T).

The insoles used in the present study were modeled using the software OrthoFoot 3D (version 1.0.5912, DeepSoft, Brazil) and 3D-shaped sandals by a CNC (Computer Numerical Control) milling machine on ethylene-vinyl-acetate (EVA, shore A 40). Two full-length insoles were used: a control insole with a flat surface (Fig. 3A) and an intervention insole with a medial longitudinal arch support and a 6° medial heel wedge (Fig. 3B). Tipnis et al. (2014) and Costa et al. (2021) demonstrated that the external KAM is influenced significantly when using wedges angled up to 6°. This insole configuration was adopted because it is frequently prescribed for individuals with excessive pronation (Bonifacio et al., 2018; Braga et al., 2019; Telfer et al., 2013). Furthermore, the insoles had a 2-cm non-slip sole fixed directly to the participants' feet with elastic bands (Fig. 2A and 2B). The order of the two insole configurations was randomized among the participants during the walking trials.

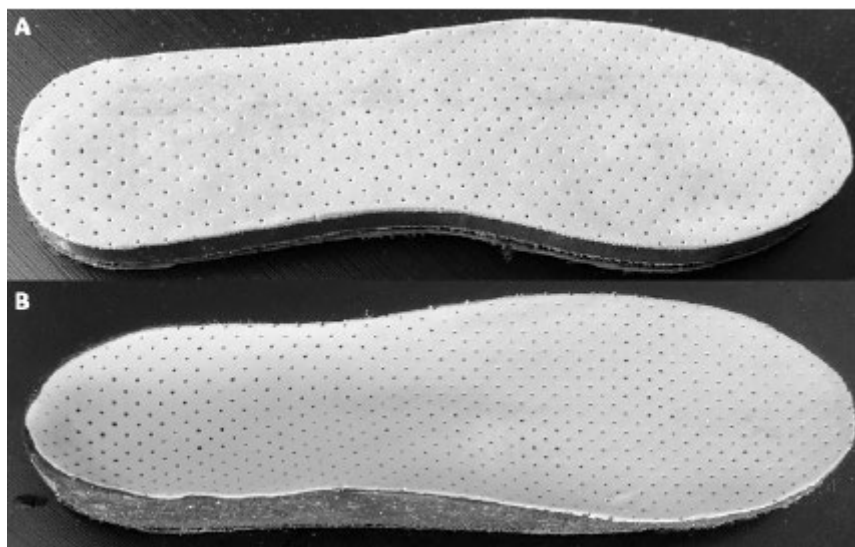


Figure 3 – A) Control insole (flat surface). B) Intervention insoles with medial longitudinal arch support (height standardized and proportional to the footwear's size) and medial heel wedge with a 6° inclination.

The kinematics and kinetics during the stance phase of gait were computed in Visual 3D (version 2021, C- Motion Inc, USA). Kinematic data were filtered with a 4th-order Butterworth low-pass filter at 6 Hz (Winter, 2005), whereas kinetic data were filtered at 15 Hz. Heel contacts and toe-offs were determined using the vertical GRF with the threshold set at 50 N (Weart et al., 2020). Midstance was defined as the phase between the forefoot contact to the ground ($\pm 11\%$ of the stance) and heel-off the ground ($\pm 63\%$ of the stance) (Araujo et al., 2019). The biomechanical multisegmented foot model was proposed by Bruening et al. (2012a) and Bruening et al. (2012b), but with the forefoot tracking markers modified by Magalhaes et al. (2020a), as presented in Fig. 2. The other segments were modeled following ISB recommendations (Wu et al., 2002).

MFJC motion was computed as the forefoot positions relative to the rearfoot, while ankle motion was the rearfoot positions relative to the shank. Knee motion was the shank position relative to the thigh, and hip motion was the thigh position relative to the pelvis. Joint net external moments were computed using the inverse dynamics method. The following variables were calculated during the midstance phase of walking for the three planes of motion: joint angles (in degrees) for the forefoot, rearfoot, MFJC, ankle, knee, and hip, as well as the ankle and knee external moments ($\text{Nm} \times \text{kg}^{-1}$). The analysis included the peaks of the following variables: forefoot eversion angle, rearfoot eversion angle, knee adduction angle, ankle eversion moment, hip external rotation angle, and hip internal rotation angle.

Forefoot, rearfoot, and knee angles, as well as the ankle moment, were calculated as the difference between control and intervention conditions to measure the effect of the intervention; positive values mean that the control condition presented higher values. Forefoot and hip transverse plane angles were calculated only for the control condition during the midstance, where negative and positive values represent the external and

internal rotation, respectively. In addition, the knee coronal plane positioning was computed while standing to document the participants' valgus alignment. The angles of MFJC, ankle, and knee were used to explore the immediate effects of the insoles, Fig. 4. Peak values of the other angles were also calculated. All data were time-normalized and extracted using the following Cardan sequence: sagittal (Y), coronal (X), and transverse (Z) planes. The average values of the five valid trials were used for analysis.

Data from all measurements were imported to Matlab® (2021a, The MathWorks, Inc., USA), where they were merged and analyzed. Participants were categorized into two groups based on the impact of the intervention insoles on the external knee adduction moment (KAM): biomechanical responders and non-responders. This assessment relied on the mean difference between the control and intervention insoles time series, as depicted in Figure 4D. To establish distinct groups, we utilized one standard deviation from this difference as the threshold ($0.022\text{Nm}\times\text{kg}^{-1}$), a value consistent with previous studies indicating its clinical significance in identifying noteworthy changes in the external KAM (Arazpour et al., 2014; Kim et al., 2018). As five participants had one foot allocated in the group “responders” and another in the group “non-responders,” their data were removed from the group with a higher sample size (i.e., “responders”), so 45 lower limbs were analyzed. Hence, 34 lower limbs were allocated to the group “responders,” while the other 11 were to the group “non-responders.” Therefore, “responders” or “non-responders” refer to the group of participants that significantly altered or not the KAM due to the use of the intervention insoles, respectively.

Inferential and descriptive analyses were conducted for the participants' characteristics, FPI, and walking speed. Data normality was verified using the Kolmogorov-Smirnov test. The mean, standard deviation (SD), 95% confidence interval

(CI_{95%}), and independent t-test were used for normally distributed data. Otherwise, the median, interquartile range (IQR), percentiles 25% and 75% (P_{25-75%}), and Wilcoxon signed-rank test were used.

A time-series analysis was performed using statistical parametric mapping (SPM) to compare the knee's pronation-related angles and external moments to the control and intervention insoles before the group division. This method has been employed in recent studies (Gontijo et al., 2023; Lourenco et al., 2022). As the data were normally distributed, paired t-tests along the normalized time series were conducted to establish the presence of any significant differences using the open-source spm1d codes (www.spm1d.org). The technical details of the SPM methods have been previously reported (Pataky, 2010; Penny et al., 2011).

Independent t-tests or Wilcoxon signed-rank tests compared the musculoskeletal and walking-related measures between the responders and non-responders. Cohen's effect size and statistical power were also calculated for each comparison. Before the analyses, outliers were detected and removed using the "median absolute deviation." Therefore, the number of participants in each group could differ slightly among the comparisons. The test-retest intra-rater reliabilities for the musculoskeletal measures were verified through the Intraclass Correlation Coefficient (ICC_{3,1}). All present study analyses were performed in Matlab®, considering a significance level of 0.05.

Results

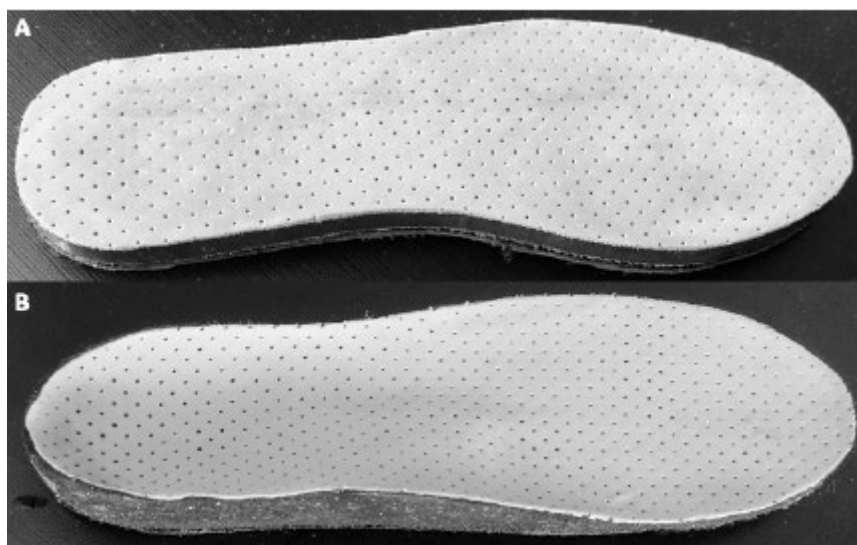
The participants' characteristics, FPI, and walking speed showed no significant differences ($p > 0.05$) between the responders and non-responders (Table 1). Across all subjects, the SPM analyses revealed that the intervention insole altered the kinematics of the MFJC in the sagittal plane ($p < 0.001$, fig. 4A) and the ankle in the sagittal and coronal planes ($p < 0.001$, fig. 4B), as well as the knee kinetics in the sagittal ($p = 0.007$,

fig. 4C) and coronal planes ($p < 0.001$, fig. 4C), compared with the control insole. Additionally, fig. 4D shows the curves depicting that the intervention and control insoles differed ($p < 0.001$) between the groups, mainly in the midstance and propulsion subphases of stance.

Table 1 Participants' characteristics, FPI, and walking speed

Variable	Descriptive	Responders (n=34)	Non-Responders (n=11)
Age (years)	Mean (SD)	24.53 (7.13)	23.82 (6.79)
	CI _{95%}	22.14 – 26.92	21.68 – 25.96
Height (m)	Median (IQR)	1.65 (0.03)	1.63 (0.13)
	P _{25-75%}	1.65 – 1.68	1.58 – 1.71
Mass (kg)	Median (IQR)	64.8 (16.6)	57.4 (17.98)
	P _{25-75%}	59.2 – 75.8	49.23 – 67.2
BMI (kg·m ⁻²)	Median (IQR)	24.4 (3.3)	22.5 (3.33)
	P _{25-75%}	22.9 – 26.2	20.93 – 24.25
FPI (score)	Mean (SD)	7.94 (1.87)	7.45 (2.42)
	CI _{95%}	7.31 – 8.57	6.02 – 8.88
Walking Speed (m·s ⁻¹)	Mean (SD)	1.04 (0.13)	1.06 (0.14)
	CI _{95%}	1.00 – 1.08	1.02 – 1.10

BMI: body mass index. FPI: foot posture index. SD: standard deviation. CI_{95%}: 95% confidence interval. IQR: interquartile range. P_{25-75%}: percentiles 25% and 75%. For variables with normal distribution, it was used the mean (SD), CI_{95%}, and two-sample t-test; otherwise, the median (IQR), P_{25-75%}, and two-sample Wilcoxon signed-rank test were used. No statistical differences existed between the groups for all variables ($p > 0.05$).



The ICC_{3,1} (CI_{95%}) was 0.98 (0.91-0.99) for FPI, 0.98 (0.94-0.99) for ankle passive dorsiflexion ROM, 0.96 (0.85-0.99) for forefoot varus, and 0.99 (0.97-0.99) for passive hip stiffness. The test-retest reliability of the torsimeter's measurements has been previously shown to be 0.98 (0.93-0.99) (Magalhaes et al., 2020b).

Table 2 shows the comparison between groups for the musculoskeletal and walking measures, including mean values for each group, IC 95%, effect size, Power, and p-value. Compared with the responders, non-responders had smaller forefoot varus (median difference = -5.51° [30.9%], $p = 0.014$), higher MFJC internal passive torque peak during inversion (mean difference = $0.02 \text{ Nm} \times \text{kg}^{-1}$ [40.0%], $p = 0.005$), and higher MFJC inversion passive stiffness during inversion (mean difference = $3.20 \times 10^{-4} \text{ Nm} \times \text{deg}^{-1} \times \text{kg}^{-1}$ [23.2%], $p = 0.022$).

During walking, Table 2 also shows that the non-responders had lower forefoot eversion angle peak (median difference = -2.75° [52.7%], $p < 0.001$), lower forefoot external rotation angle peak (median difference = -3.64° [117.8%], $p = 0.037$), and lower rearfoot eversion angle peak (median difference = -2.63° [56.3%], $p = 0.022$). Moreover, considering only the control condition, the non-responders had higher knee adduction angle peak (median difference = 1.68° [46.1%], $p = 0.045$), hip external rotation angle (median difference = 4.25° [750.6%], $p = 0.022$), and hip internal rotation angle (median difference = 3.92° [105.7%], $p = 0.026$).

Table 2 Comparisons between responders and non-responders for the musculoskeletal and walking measures

Variable	Descriptive	Responders	Non-Responders	d	Power	<i>p</i>
Ankle Passive Dorsiflexion ROM (deg)	n Mean (SD) CI _{95%}	34 34.27 (4.8) 32.66 – 35.88	10 37.08 (3.04) 35.20 – 38.96	0.63	0.98	0.09
Forefoot-Shank Alignment (deg)	n Median (IQR) P _{25-75%}	34 17.84 (6.00) 15.33 – 21.33	11 12.33 (4.09) 11.42 – 15.50	0.9	1	0.014*
Hip Passive Stiffness (deg·kg ⁻¹)	n Mean (SD) CI _{95%}	34 0.65 (0.25) 0.57 – 0.73	11 0.77 (0.25) 0.61 – 0.93	0.49	0.89	0.164
MFJC Inversion Torque (Nm·kg ⁻¹)	n Mean (SD) CI _{95%}	34 0.05 (0.01) 0.05 – 0.05	11 0.07 (0.03) 0.05 – 0.09	1.03	1	0.005*
MFJC Inversion Stiffness (Nm·deg ⁻¹ ·kg ⁻¹)	n Mean (SD) CI _{95%}	34 1.38e-03 (3.05e-04) 1.28e-03 – 1.48e-03	11 1.70e-03 (6.05e-04) 1.34e-03 – 2.06e-03	0.83	0.99	0.022*
Knee Frontal Plane Alignment in Standing Position (deg)	n Mean (SD) CI _{95%}	33 3.01 (3.02) 2.06 – 4.14	11 1.6 (3.5) -0.48 – 3.68	0.48	0.87	0.177
Forefoot Eversion Angle Peak during stance (deg) †	n Median (IQR) P _{25-75%}	34 5.22 (3.05) 3.83 – 6.88	11 2.47 (2.34) 1.78 – 4.12	1.46	1	0.001*
Rearfoot Eversion Angle Peak during stance (deg) †	n Median (IQR) P _{25-75%}	34 4.66 (3.38) 2.58 – 5.96	11 2.03 (3.33) 0.58 – 3.91	0.79	0.99	0.022*
Knee Adduction Angle Peak during stance (deg) †	n Median (IQR) P _{25-75%}	34 3.64 (3.29) 2.05 – 5.34	10 1.96 (1.27) 1.39 – 2.66	0.74	0.99	0.045*
Ankle Eversion Moment Peak during stance (Nm·kg ⁻¹) †	n Median (IQR) P _{25-75%}	34 0.04 (0.04) 0.02 – 0.06	10 0.03 (0.04) 0.00 – 0.04	0.90	1	0.022*
Forefoot External Rotation Angle Peak during midstance (deg) ‡	n Median (IQR) P _{25-75%}	33 3.09 (3.94) 0.73 – 4.66	11 -0.55 (4.32) -1.87 – 2.45	0.83	0.99	0.037*
Hip External Rotation Angle Peak during midstance (deg) ‡	n Median (IQR) P _{25-75%}	34 -0.57 (7.78) -4.21 – 3.58	10 3.68 (8.14) 0.11 – 8.24	0.89	1	0.022*

Hip Internal Rotation	n	34	10			
Angle Peak during	Median (IQR)	3.70 (9.14)	7.62 (7.34)	0.78	0.99	0.026*
midstance (deg) ‡	P _{25-75%}	-1.42 – 7.72	5.92 – 13.26			

* $p < 0.05$. SD: standard deviation. CI95%: 95% confidence interval. IQR: interquartile range. P25-75%: percentiles 25% and 75%. For variables with normal distribution, mean (SD), C95%, and independent t-test were used; otherwise, median (IQR), P25-75%, and independent Wilcoxon signed-rank test were used. d: Cohen's effect size. Power: Statistical power. p: p-value. n: number of participants in each group. ROM: range of motion. Deg: degrees. Nm: newton-meters. Kg: kilogram. MJFC: midfoot joint complex. †: The difference between control and intervention conditions was used for the forefoot eversion angles, rearfoot eversion angles, knee adduction angles, and ankle eversion moment, where positive mean higher values for the control condition. ‡: Only the control condition during the midstance was considered for the forefoot external rotation angles and hip transverse plane angles, where negative and positive values represent the external and internal rotation, respectively.

Discussion

The present study identified clinical and biomechanical factors that influenced the effects of the pronation-controlling insoles on the external KAM. Individuals who did not increase the external KAM (i.e., non-responders) showed musculoskeletal factors related to less foot pronation. Specifically, they had lower forefoot varus and higher midfoot internal passive torque and stiffness during forefoot inversion. Also, during walking, the non-responders had foot kinematic patterns indicative of less pronation than the responders (i.e., lower forefoot external rotation and eversion, lower rearfoot eversion, and lower ankle eversion external moment), even though all participants had similar excessively pronated postures according to the FPI. Interestingly, the non-responders had a more internally rotated hip and a less adducted knee in walking, suggesting that their excessive foot pronation during walking could be due to the proximal factors in the hip (Resende et al., 2015). Thus, having a more internally rotated hip had a role in the absence of changes in the external KAM.

Although all participants were overpronators, factors contributing to pronation differ by group. While the non-responders had proximal causes of pronation (i.e., increased hip internal rotation during walking), the responders had foot-related causes of pronation

(i.e., higher forefoot varus and lower midfoot internal passive torque and stiffness). The non-responders had increased hip internal rotation and decreased external rotation during walking. A consequence of increased internal hip rotation at the knee is the reduction in external KAM due to a medial shift in the knee joint center position (Shull et al., 2013). This tendency to reduce KAM may have caused the lack of KAM increases in the non-responders.

In the present study, the non-responders had less forefoot varus (Table 2). Individuals with less forefoot varus may have the GRF positioned more medially (Cobb et al., 2004), thus producing a lower external pronation moment (Gross et al., 2007). As the external KAM calculations consider the GRF magnitude and knee-GRF force lever arm (Lewinson et al., 2015), people with less forefoot varus have a reduced knee-GRF force lever arm and, thus, lower external KAM peaks (Hinman et al., 2012). Therefore, non-responders had less forefoot varus, consequently moving the GRF medially under the foot, reducing the knee-GRF force lever arm, and minimizing the effects of the intervention insoles on the knee.

In addition, the non-responders also presented higher MFJC passive internal torque and stiffness during forefoot inversion. Paes et al. (2019) demonstrated a moderate correlation between forefoot varus and MFJC internal passive torque, showing that the lower the forefoot varus, the greater the MFJC passive internal torque. Therefore, lower forefoot varus and higher MFJC internal passive torque and stiffness seem to reduce the effect of the insoles on foot pronation and external KAM. A novel aspect of our study was the measurement of MFJC passive torque and stiffness using a Foot Torsimeter, which has yet to be included in previous studies considering foot orthoses.

The intervention insoles produced the desired effect of increasing KAM only in subjects with pronation causes related to the foot-ankle complex, as the responders had

higher forefoot varus and lower midfoot internal passive torque and stiffness during inversion. Therefore, the efficacy of insoles may vary based on individual musculoskeletal characteristics, suggesting a need for careful consideration before prescription. This study is an exploratory investigation, indicating the necessity for further research to substantiate this assertion. The intervention insoles, as expected, could control foot pronation during stance (Fig. 4A and Fig. 4B) in all participants (Bonifacio et al., 2018; Braga et al., 2019). Moreover, the insoles also promoted an increased external KAM in most participants (Fig. 4C), which was also expected (Bonifacio et al., 2018; Telfer et al., 2013). Thus, overall, the orthotics influenced foot pronation and KAM, as assumed, but that effect was more significant in people with foot-related pronation causes.

Another factor that can influence external KAM is knee malalignment. A standard measure of knee alignment is the angle between the thigh and shank in an anteroposterior radiograph. In the present study, no specific measurement for knee alignment was performed. However, the angle between the thigh and knee was calculated in the coronal plane standing position. The foot and hip positions were not standardized during this standing position once the participants were asked to assume their relaxed position. In addition, altering the loading in either medial or lateral knee compartments by using insoles may benefit patients with knee OA. A recent review and meta-analysis (99) shows that the prevalence rate for single-compartment knee OA is 50%. Thus, balancing the loading between the knee compartments is desirable to equalize this risk factor. Future studies should investigate this further.

Some limitations to this study should be considered. A clinical measurement for the knee alignment is absent. Five subjects were categorized as having one lower limb as a "responder" and the other as a "non-responder." Given this condition, the lower limb

labeled as a "responder" was excluded from the analysis. Consequently, three participants had both lower limbs, and five had one lower limb allocated to the "non-responder" group. Future studies should ensure consistency by selecting subjects with both feet allocated to the same group. No adjustments were made to the *p*-values to allow for multiple comparisons. Still, by calculating the percent error rate recommended by Ottenbacher (1998), we found that 92% of the comparisons (twelve out of thirteen) are likely due to non-chance factors. Finally, the methods used in this study should be extended to investigate further the response to insoles used by elders, people with pathologies, such as patellofemoral pain and knee OA, as well as during other daily activities, such as running, squatting, and stair ascent and descent.

Our results have shown that most people with excessive foot pronation had an external KAM increase due to the arch-supported insoles with a 6° medial wedge. This effect may influence the overload on the lateral compartment of the knee (Bonifacio et al., 2018; Felson et al., 2013), which may prevent the onset or progression of lateral knee osteoarthritis. However, participants with low forefoot varus angles, large MFJC internal passive torque, greater stiffness during forefoot inversion, low knee valgus angles, hip internal rotation, and toed-in feet during walking did not significantly change the external KAM. During assessments, clinicians should consider these measurements when prescribing medially wedged insoles in people with excessive pronation.

Acknowledgments

This work was supported by the Brazilian funding agencies "Coordenação de Aperfeiçoamento de Pessoal de Nível Superior" (CAPES, Finance Code 001), "Fundação de Amparo à Pesquisa do Estado de Minas Gerais" (FAPEMIG), and "Conselho Nacional de Desenvolvimento Científico e Tecnológico" (CNPq).

References

- Andriacchi, T.P., 2013. Valgus alignment and lateral compartment knee osteoarthritis: a biomechanical paradox or new insight into knee osteoarthritis? *Arthritis Rheum* 65, 310-313.
- Araujo, V.L., Santos, T.R.T., Khuu, A., Lewis, C.L., Souza, T.R., Holt, K.G., Fonseca, S.T., 2020. The effects of small and large varus alignment of the foot-ankle complex on lower limb kinematics and kinetics during walking: A cross-sectional study. *Musculoskelet Sci Pract* 47, 102149.
- Araujo, V.L., Souza, T.R., Magalhaes, F.A., Santos, T.R.T., Holt, K.G., Fonseca, S.T., 2019. Effects of a foot orthosis inspired by the concept of a twisted osteoligamentous plate on the kinematics of foot-ankle complex during walking: A proof of concept. *J Biomech* 93, 118-125.
- Arazpour, M., Hutchins, S.W., Bani, M.A., Curran, S., Aksenov, A., 2014. The influence of a bespoke unloader knee brace on gait in medial compartment osteoarthritis: a pilot study. *Prosthet Orthot Int* 38, 379-386.
- Bonifacio, D., Richards, J., Selfe, J., Curran, S., Trede, R., 2018. Influence and benefits of foot orthoses on kinematics, kinetics and muscle activation during step descent task. *Gait Posture* 65, 106-111.
- Braga, U.M., Mendonca, L.D., Mascarenhas, R.O., Alves, C.O.A., Filho, R.G.T., Resende, R.A., 2019. Effects of medially wedged insoles on the biomechanics of the lower limbs of runners with excessive foot pronation and foot varus alignment. *Gait Posture* 74, 242-249.
- Bruening, D.A., Cooney, K.M., Buczek, F.L., 2012a. Analysis of a kinetic multi-segment foot model part II: kinetics and clinical implications. *Gait Posture* 35, 535-540.
- Bruening, D.A., Cooney, K.M., Buczek, F.L., 2012b. Analysis of a kinetic multi-segment foot model. Part I: Model repeatability and kinematic validity. *Gait Posture* 35, 529-534.
- Cardoso, T.B., Ocarino, J.M., Fajardo, C.C., Paes, B.D.C., Souza, T.R., Fonseca, S.T., Resende, R.A., 2020. Hip external rotation stiffness and midfoot passive mechanical resistance are associated with lower limb movement in the frontal and transverse planes during gait. *Gait Posture* 76, 305-310.
- Chapman, G.J., Parkes, M.J., Forsythe, L., Felson, D., Jones, R., 2015. Ankle motion influences the external knee adduction moment and may predict who will respond to lateral wedge insoles?: an ancillary analysis from the SILK trial. *Osteoarthritis and cartilage* 23, 1316-1322.
- Cheung, R.T., Chung, R.C., Ng, G.Y., 2011. Efficacies of different external controls for excessive foot pronation: a meta-analysis. *British journal of sports medicine* 45, 743-751.
- Chuter, V.H., Janse de Jonge, X.A., 2012. Proximal and distal contributions to lower extremity injury: a review of the literature. *Gait Posture* 36, 7-15.
- Cobb, S.C., Tis, L.L., Johnson, B.F., Higbie, E.J., 2004. The effect of forefoot varus on postural stability. *The Journal of orthopaedic and sports physical therapy* 34, 79-85.
- Costa, B.L., Magalhaes, F.A., Araujo, V.L., Richards, J., Vieira, F.M., Souza, T.R., Trede, R., 2021. Is there a dose-response of medial wedge insoles on lower limb biomechanics in people with pronated feet during walking and running? *Gait Posture* 90, 190-196.

Cruz, A.C., Fonseca, S.T., Araújo, V.L., Carvalho, D.S., Barsante, L.D., Pinto, V.A., Souza, T.R., 2019. Pelvic drop changes due to proximal muscle strengthening depend on foot-ankle varus alignment. *Appl Bionics Biomech* 2019.

Dill, K.E., Begalle, R.L., Frank, B.S., Zinder, S.M., Padua, D.A., 2014. Altered knee and ankle kinematics during squatting in those with limited weight-bearing–lunge ankle-dorsiflexion range of motion. *J Athl Train* 49, 723-732.

Fajardo, C.C., Cardoso, T.B., Gontijo, B.A., Magalhães, F.A., Souza, T.R., Fonseca, S.T., Ocarino, J.M., Resende, R.A., 2021. Hip passive stiffness is associated with midfoot passive stiffness. *Brazilian journal of physical therapy*.

Farahpour, N., Jafarnezhadgero, A., Allard, P., Majlesi, M., 2018a. Muscle activity and kinetics of lower limbs during walking in pronated feet individuals with and without low back pain. *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology* 39, 35-41.

Farahpour, N., Jafarnezhadgero, A., Allard, P., Majlesi, M., 2018b. Muscle activity and kinetics of lower limbs during walking in pronated feet individuals with and without low back pain. *Journal of Electromyography and Kinesiology* 39, 35-41.

Faul, F., Erdfelder, E., Lang, A.G., Buchner, A., 2007. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behav Res Methods* 39, 175-191.

Felson, D.T., Niu, J., Gross, K.D., Englund, M., Sharma, L., Cooke, T.D., Guermazi, A., Roemer, F.W., Segal, N., Goggins, J.M., Lewis, C.E., Eaton, C., Nevitt, M.C., 2013. Valgus malalignment is a risk factor for lateral knee osteoarthritis incidence and progression: findings from the Multicenter Osteoarthritis Study and the Osteoarthritis Initiative. *Arthritis Rheum* 65, 355-362.

Gontijo, B.A., Fonseca, S.T., Araujo, P.A., Magalhaes, F.A., Trede, R.G., Faria, H.P., Resende, R.A., Souza, T.R., 2023. A new marker cluster anchored to the iliotibial band improves tracking of hip and thigh axial rotations. *J Biomech* 147, 111452.

Gross, K.D., Niu, J., Zhang, Y.Q., Felson, D.T., McLennan, C., Hannan, M.T., Holt, K.G., Hunter, D.J., 2007. Varus foot alignment and hip conditions in older adults. *Arthritis Rheum* 56, 2993-2998.

Harris, P.A., Taylor, R., Minor, B.L., Elliott, V., Fernandez, M., O'Neal, L., McLeod, L., Delacqua, G., Delacqua, F., Kirby, J., Duda, S.N., Consortium, R.E., 2019. The REDCap consortium: Building an international community of software platform partners. *J Biomed Inform* 95, 103208.

Hinman, R.S., Bowles, K.A., Metcalf, B.B., Wrigley, T.V., Bennell, K.L., 2012. Lateral wedge insoles for medial knee osteoarthritis: effects on lower limb frontal plane biomechanics. *Clin Biomech (Bristol, Avon)* 27, 27-33.

Hinman, R.S., Bowles, K.A., Payne, C., Bennell, K.L., 2008. Effect of length on laterally-wedged insoles in knee osteoarthritis. *Arthritis Rheum* 59, 144-147.

Horwood, A.M., Chockalingam, N., 2017. Defining excessive, over, or hyper-pronation: A quandary. *Foot (Edinb)* 31, 49-55.

Jafarnezhadgero, A.A., Oliveira, A.S., Mousavi, S.H., Madadi-Shad, M., 2018. Combining valgus knee brace and lateral foot wedges reduces external forces and moments in osteoarthritis patients. *Gait & posture* 59, 104-110.

Jafarnezhadgero, A.A., Shad, M.M., Majlesi, M., 2017. Effect of foot orthoses on the medial longitudinal arch in children with flexible flatfoot deformity: A three-dimensional moment analysis. *Gait & Posture* 55, 75-80.

Jani, A., Dabholkar, A., Yardi, S., 2012. A Comparative Study of Ankle and Foot Characteristics in Knee Osteoarthritis Patients and Normals. *Indian J Physiother Occup Ther* 6.

Jones, R.K., Chapman, G.J., Forsythe, L., Parkes, M.J., Felson, D.T., 2014. The relationship between reductions in knee loading and immediate pain response whilst wearing lateral wedged insoles in knee osteoarthritis. *Journal of Orthopaedic Research* 32, 1147-1154.

Kang, M.H., Oh, J.S., 2017. Relationship between weightbearing ankle dorsiflexion passive range of motion and ankle kinematics during gait. *J Am Podiatr Med Assoc* 107, 39-45.

Kean, C.O., Bennell, K.L., Wrigley, T.V., Hinman, R.S., 2015. Relationship between hip abductor strength and external hip and knee adduction moments in medial knee osteoarthritis. *Clin Biomech (Bristol, Avon)* 30, 226-230.

Kim, Y., Richards, J., Lidtke, R.H., Trede, R., 2018. Characteristics of clinical measurements between biomechanical responders and non-responders to a shoe designed for knee osteoarthritis. *Gait Posture* 59, 23-27.

Lewinson, R.T., Worobets, J.T., Stefanyshyn, D.J., 2014. The relationship between maximal hip abductor strength and resultant loading at the knee during walking. *Proc Inst Mech Eng H* 228, 1258-1263.

Lewinson, R.T., Worobets, J.T., Stefanyshyn, D.J., 2015. Calculation of external knee adduction moments: a comparison of an inverse dynamics approach and a simplified lever-arm approach. *Knee* 22, 292-297.

Leys, C., Ley, C., Klein, O., Bernard, P., Licata, L., 2013. Detecting outliers: Do not use standard deviation around the mean, use absolute deviation around the median. *J Exp Soc Psychol* 49, 764-766.

Lourenco, B.M., Magalhaes, F.A., Vieira, F.M., Reis, C.K., Costa, H.S., Araujo, V.L., Richards, J., Trede, R., 2022. An exploration of the effects of prefabricated and customized insoles on lower limb kinetics and kinematics during walking, stepping up and down tasks: A time series analysis. *Gait Posture* 98, 297-304.

Magalhaes, F.A., Fonseca, S.T., Araujo, V.L., Trede, R.G., Oliveira, L.M., Castor, C., Pinto, R.Z., Souza, T.R., 2021. Midfoot passive stiffness affects foot and ankle kinematics and kinetics during the propulsive phase of walking. *J Biomech* 119, 110328.

Magalhaes, F.A., Souza, T.R., Araujo, V.L., Oliveira, L.M., Silveira, L.P., Ocarino, J.M., Fonseca, S.T., 2020a. Comparison of the rigidity and forefoot - Rearfoot kinematics from three forefoot tracking marker clusters during walking and weight-bearing foot pronation-supination. *J Biomech* 98, 109381.

Magalhaes, F.A., Souza, T.R., Araujo, V.L., Resende, R.A., Pertence, A.E.M., Chagas, M.H., Fonseca, S.T., 2020b. Reliability and sensitivity of an instrument for measuring the midfoot passive mechanical properties. *J Biomech* 104, 109735.

Mendes, A.A.M.T., Silva, H.J.A., Costa, A.R.A., Pinheiro, Y.T., Lins, C.A.A., Souza, M.C., 2020. Main types of insoles described in the literature and their applicability for musculoskeletal disorders of the lower limbs: A systematic review of clinical studies. *J Bodyw Mov Ther*.

- Neal, B.S., Griffiths, I.B., Dowling, G.J., Murley, G.S., Munteanu, S.E., Franettovich Smith, M.M., Collins, N.J., Barton, C.J., 2014. Foot posture as a risk factor for lower limb overuse injury: a systematic review and meta-analysis. *J Foot Ankle Res* 7, 1-13.
- Ohi, H., Iijima, H., Aoyama, T., Kaneda, E., Ohi, K., Abe, K., 2017. Association of frontal plane knee alignment with foot posture in patients with medial knee osteoarthritis. *BMC Musculoskelet Disord* 18, 246.
- Ottenbacher, K.J., 1998. Quantitative evaluation of multiplicity in epidemiology and public health research. *American journal of epidemiology* 147, 615-619.
- Paes, B., Resende, R.A., Gomes, R.B., Gontijo, B.A., Magalhaes, F.A., Ocarino, J.M., Fonseca, S.T., Souza, T.R., 2019. The clinical measure of forefoot-shank alignment partially reflects mechanical properties of the midfoot joint complex. *Musculoskelet Sci Pract* 42, 98-103.
- Pataky, T.C., 2010. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. *J Biomech* 43, 1976-1982.
- Penny, W.D., Friston, K.J., Ashburner, J.T., Kiebel, S.J., Nichols, T.E., 2011. *Statistical parametric mapping: the analysis of functional brain images*. Elsevier.
- Redmond, A.C., Crosbie, J., Ouvrier, R.A., 2006. Development and validation of a novel rating system for scoring standing foot posture: the Foot Posture Index. *Clin Biomech (Bristol, Avon)* 21, 89-98.
- Resende, R.A., Deluzio, K.J., Kirkwood, R.N., Hassan, E.A., Fonseca, S.T., 2015. Increased unilateral foot pronation affects lower limbs and pelvic biomechanics during walking. *Gait Posture* 41, 395-401.
- Rodrigues, P., Chang, R., TenBroek, T., Hamill, J., 2013. Medially posted insoles consistently influence foot pronation in runners with and without anterior knee pain. *Gait posture* 37, 526-531.
- Sawada, T., Tokuda, K., Tanimoto, K., Iwamoto, Y., Ogata, Y., 2016a. Gait & Posture Foot alignments influence the effect of knee adduction moment with lateral wedge insoles during gait.
- Sawada, T., Tokuda, K., Tanimoto, K., Iwamoto, Y., Ogata, Y., Anan, M., Takahashi, M., Kito, N., Shinkoda, K., 2016b. Foot alignments influence the effect of knee adduction moment with lateral wedge insoles during gait. *Gait Posture* 49, 451-456.
- Shull, P.B., Shultz, R., Silder, A., Dragoo, J.L., Besier, T.F., Cutkosky, M.R., Delp, S.L., 2013. Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis. *J Biomech* 46, 122-128.
- Stoddart, J.C., Dandridge, O., Garner, A., Cobb, J., van Arkel, R.J., 2021. The compartmental distribution of knee osteoarthritis - a systematic review and meta-analysis. *Osteoarthritis Cartilage* 29, 445-455.
- Telfer, S., Abbott, M., Steultjens, M., Rafferty, D., Woodburn, J., 2013. Dose-response effects of customised foot orthoses on lower limb muscle activity and plantar pressures in pronated foot type. *Gait Posture* 38, 443-449.
- Tipnis, R.A., Anloague, P.A., Laubach, L.L., Barrios, J.A., 2014. The dose-response relationship between lateral foot wedging and the reduction of knee adduction moment. *Clin Biomech* 29, 984-989.

Ulrich, B., Hoffmann, L., Jolles, B.M., Favre, J., 2020. Changes in ambulatory knee adduction moment with lateral wedge insoles differ with respect to the natural foot progression angle. *J Biomech* 103, 109655.

van Tunen, J.A.C., Dell'Isola, A., Juhl, C., Dekker, J., Steultjens, M., Thorlund, J.B., Lund, H., 2018. Association of malalignment, muscular dysfunction, proprioception, laxity and abnormal joint loading with tibiofemoral knee osteoarthritis - a systematic review and meta-analysis. *BMC Musculoskelet Disord* 19, 273.

Wang, S., Chan, P.P., Lam, B.M., Chan, Z.Y., Zhang, J.H., Wang, C., Lam, W.K., Ho, K.K.W., Chan, R.H., Cheung, R.T., 2021a. Sensor-based gait retraining lowers knee adduction moment and improves symptoms in patients with knee osteoarthritis: a randomized controlled trial. *Sensors* 21, 5596.

Wang, S., Chan, P.P.K., Lam, B.M.F., Chan, Z.Y.S., Zhang, J.H.W., Wang, C., Lam, W.K., Ho, K.K.W., Chan, R.H.M., Cheung, R.T.H., 2021b. Sensor-Based Gait Retraining Lowers Knee Adduction Moment and Improves Symptoms in Patients with Knee Osteoarthritis: A Randomized Controlled Trial. *Sensors (Basel)* 21, 5596.

Weart, A.N., Miller, E.M., Freisinger, G.M., Johnson, M.R., Goss, D.L., 2020. Agreement Between the OptoGait and Instrumented Treadmill System for the Quantification of Spatiotemporal Treadmill Running Parameters. *Front sports act living* 2: 571385.

Winter, D., 2005. *Biomechanics and Motor Control of Human Movement*, Willey & Sons. INC, Waterloo.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D D'Lima, D., Cristofolini, L., Witte, H., 2002. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. *J Biomech* 35, 543-548.

Yu, L., Wang, Y., Yang, J., Wang, J., Zhang, Y., 2021. Effects of orthopedic insoles on patients with knee osteoarthritis: A meta-analysis and systematic review. *J Rehabil Med* 53, jrm00191.