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Effects of medially wedged insoles on the biomechanics of the lower limbs of runners with excessive foot pronation and foot varus alignment

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A R T I C L E I N F O

Keywords:
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A B S T R A C T

Background: Excessive foot pronation during running in individuals with foot varus alignment may be reduced by medially wedged insoles.

Research question: This study investigated the effects of a medially wedged insole at the forefoot and at the rearfoot on the lower limbs angles and internal moments of runners with excessive foot pronation and foot varus alignment.

Methods: Kinematic and kinetic data of 19 runners (11 females and 8 males) were collected while they ran wearing flat (control condition) and medially wedged insoles (insole condition). Both insoles had arch support. We used principal component analysis for data reduction and dependent *t*-test to compare differences between conditions.

Results: The insole condition reduced ankle eversion ($p = 0.003$; effect size = 0.63); reduced knee range of motion in the transverse plane ($p = 0.012$; effect size = 0.55); increased knee range of motion in the frontal plane in early stance and had earlier knee adduction peak ($p = 0.018$; effect size = 0.52); reduced hip range of motion in the transverse plane ($p = 0.031$; effect size = 0.48); reduced hip adduction ($p = 0.024$; effect size = 0.50); reduced ankle inversion moment ($p = 0.012$; effect size = 0.55); and increased the difference between the knee internal rotation moment in early stance and midstance ($p = 0.012$; effect size = 0.55).

Significance: Insoles with 7° medial wedges at the forefoot and rearfoot are able to modify motion and moments patterns that are related to lower limb injuries in runners with increased foot pronation and foot varus alignment with some non-desired effects on the knee motion in the frontal plane.

1. Introduction

The prevalence of injuries in runners is high. For example, Hespanhol Junior et al. [1] found that approximately 30% of runners develop lower limb injuries in 12 weeks of follow-up. Different factors, such as excessive foot pronation, footwear characteristics and training parameters contribute to the development of injuries during running [2–4]. Excessive foot pronation during the stance phase of running increases knee internal rotation [5] and hip adduction and internal rotation [6], which may help to explain the association between excessive foot pronation and different lower limb injuries (e.g. tibial stress syndrome [7,8] and patellar tendinopathy [9]). In addition, although there is no study demonstrating the effects of increased foot pronation on the pelvis movement during running, during gait, increased foot pronation reduces the lower limb functional length and consequently increases pelvic ipsilateral drop [10]. Therefore, understanding the mechanical effects of available interventions to reduce excessive foot pronation in runners may contribute to the treatment and prevention of related injuries.

Increased foot varus alignment assessed in open kinematic chain increases foot pronation during the stance phase of running [11]. Medially wedged insole is an option to reduce excessive foot pronation during running due to increased foot varus alignment. However, the results of most previous studies about the effects of medially wedged

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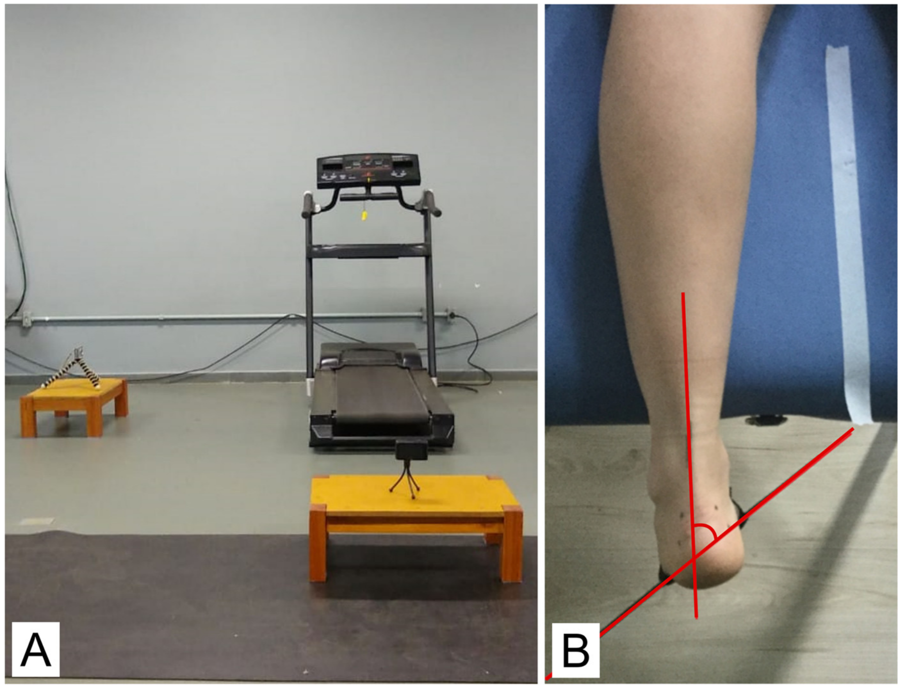

Fig. 1. A) Setup for the qualitative video analysis of foot strike pattern and forefoot and rearfoot pronation during running; B)Shank-forefoot alignment measurement.

insoles are inconclusive. For example, Almonroeder et al. [12] demonstrated that medially wedged insoles anticipated the peak of ankle inversion moment during the stance phase of running. On the other hand, Eslami et al. [13] did not find any effect of a medially wedged insole on the time to rearfoot eversion peak during running. These partially inconclusive results may be explained by the possible different magnitudes of foot pronation and foot alignments of the participants of these studies. Unfortunately, most of the previous studies did not report participants' foot alignment and magnitude of foot pronation nor used these factors as inclusion criteria [12–18]. For example, although Mündermann et al. [16] only included individuals with increased foot pronation, they did not use foot varus alignment as inclusion criterion. Therefore, it is not clear that the participants had increased foot pronation due to foot varus alignment, since foot pronation is a complex movement that might be due to other factors, such as hip muscles weakness [19], that should consequently be target by different interventions, such as hip muscles strengthening [19]. Our study aims at filling this gap.

Interventions such as the use of medially wedged insoles during running should be prescribed based on specific assessment criteria. Considering the rationale of the mechanical effects of medially wedged insoles, they would probably be more suitable for runners with excessive foot pronation and foot varus alignment. However, most of the previous studies did not select participants based on this criterion [13,14]. In addition, most of the previous studies investigated the effects of medially wedged insoles only at the rearfoot. This may also explain why some studies did not show the mechanical effects of medially wedged insoles, since the forefoot alignment, measure din a non-weight bearing position, determines the magnitude of foot pronation during running [11].

This study investigated the effects of an insole with medial wedge at the forefoot and at the rearfoot and arch support on the angular displacements and moments in the frontal and transverse planes of the lower limbs of runners with excessive foot pronation and increased foot varus alignment. Our hypotheses were that the medially wedged insole would reduce ankle eversion, knee abduction and internal rotation, hip adduction and internal rotation, and reduce ipsilateral pelvic drop, ankle inversion moment and knee and hip external rotation moment during the stance phase of running.

2. Methods

2.1. Participants

Sample size was determined using the software G*Power [20] with the following input data: repeated measures t test, desired statistical power of 80%, significance level of 0.05, and an expected medium effect size ($d = 0.7$). This resulted in an estimated minimum sample size of 19 participants. Nineteen healthy adult recreational runners (11 females and 8 males) with average age, mass and height of 35.7years (SD 7.72), 65.9 kg (SD 9.9) and 1.66 m (SD 0.08), respectively, participated in this study. The inclusion criteria were: (i) age between 18 and 50 years old; (ii) minimum of six months of experience in running and at least 10 km of training per week; (iii) no history of lower limbs or back surgery or injury and no use of foot orthoses during the past six months; (iii) present heel strike pattern and excessive foot pronation during the stance phase of running and shank-forefoot varus alignment equal or greater than 10° [21]; and (iv) no structural or functional leg length discrepancy greater than 0.5 cm [22,23]. The exclusion criterion was report of discomfort or pain during data collection. None of the participants were excluded. Each participant signed a consent form approved by the university's Ethical Research Committee (CAAE: 69277417.2.0000.5149).

2.2. Instruments and procedures

Initially, the participant's height and mass were measured. Then, inclusion criteria were measured. Increased foot pronation was determined based on a qualitative video analysis of running [24]. Participants ran for 40 s on a treadmill (S62, embreex, SC, Brazil, 40 X 130 cm) at self-selected speed wearing a neutral shoe (Bouts New Fit®). Two cameras (Sony, DSC-TX5, 10.2 mega pixels) were positioned on the lateral and posterior views at standardized distances and heights (Fig. 1A). The examiner assessed the lateral view video to identify heel

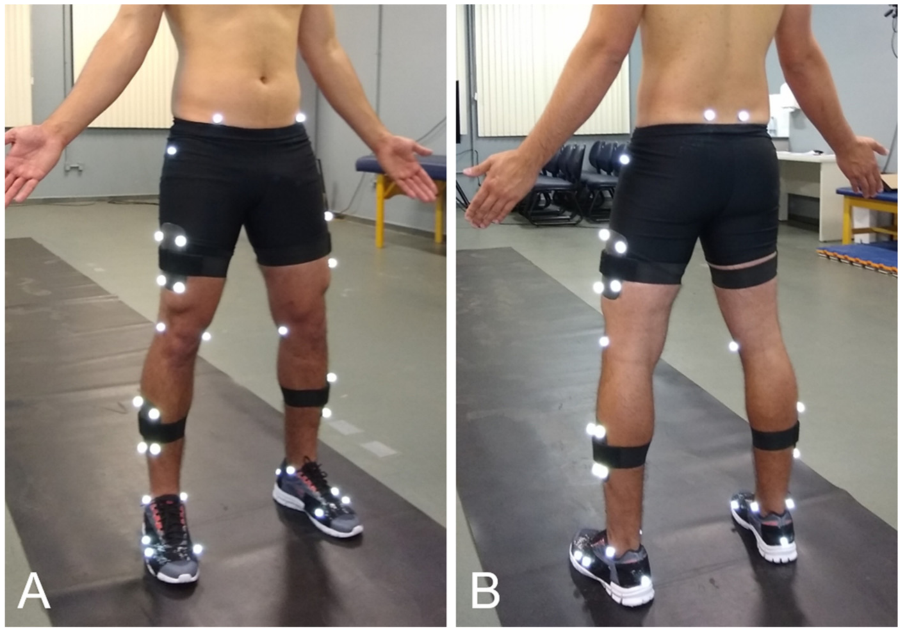

Fig. 2. Marker placement: A) anterior, and B) posterior views.

strike pattern and the posterior view to identify increased forefoot and rearfoot pronation during the midstance phase of running [24]. Next, the shank-forefoot alignment was measured following the methods described by Mendonça et al. [21] and summarized in the Supplementary Material (Fig. 1B).

Following inclusion criteria measurement, running data were recorded at 200 Hz using an 8-camera motion capture system (Oqus 3 +, Qualisys, Gothenburg, Sweden) synchronized with three force plates (Bertec FP 4060-08, Columbus, OH, EUA) positioned in series and not visually identifiable by the participants. The force plates registered ground reaction force data at 1000 Hz, which was subsequently down sampled at 200 Hz. Anatomical and clusters of tracking markers were used to determine the coordinates of the pelvis, thigh, shank and feet [25] using data obtained with the participant in a relaxed standing position (static trials) (Fig. 2). The same experienced examiner positioned the markers on all participants. Participants then ran at self-selected speed in a 15-meters walkway under two different conditions, as described:

1) control condition: wearing a pair of flat insoles at the forefoot and rearfoot and with a semi-rigid arch support (Fig. 3A);
2) insole condition: wearing a pair of insoles with a semi-rigid arch support and with medial elevation at the forefoot and at the rearfoot of 7° each (Fig. 3B).

The magnitude of the medial wedge was defined considering the inconclusive results of previous studies using smaller degrees of medial wedges (i.e. 4° or 5°). The order of data collection between each condition was randomized. Before data collection in each condition, participants ran for approximately one minute to familiarize with each set of insoles. Only the data of the lower limb with the greatest magnitude of shank-forefoot varus alignment were analyzed for the two conditions. The participants ran between 10 and 20 trials to allow at least five valid trials (i.e. with proper foot contact on the force plates) [14,18]. The insoles were made from a block of ethyl vinyl acetate with thermo moldable polymer (shore hardness of 45A) and were manufactured through an automated computer numeric control machine (CNC routers).

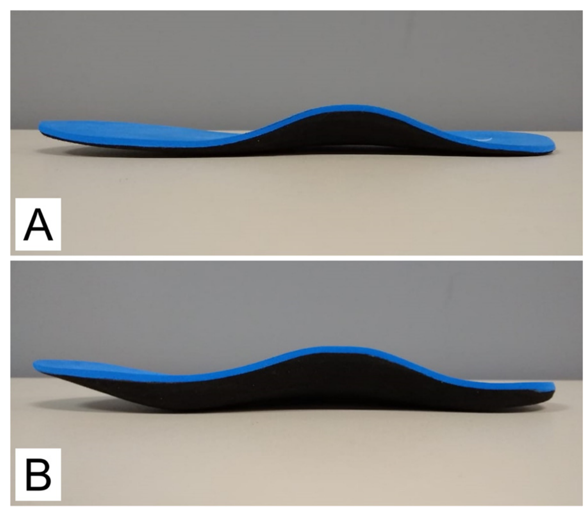

Fig. 3. Insoles used in the A) control and B) insole conditions.

2.3. Data reduction

Running data were processed using the software Visual3D (C-motion, Inc., Rockville, USA). Raw kinematic and force data were filtered using a low-pass fourth order Butterworth filter with a cut-off frequency set at 15 Hz and 25 Hz, respectively. Initial contact and toe-off were determined automatically in Visual3D using the vertical ground reaction force and a threshold of 10 N. Description of the joint centers definition and segment coordinate systems are provided in Supplementary Material. The following joint kinematics were calculated: (i) ankle inversion-eversion; (ii) knee and (iii) hip adduction-abduction and internal-external rotation; and (iv) pelvic ipsilateral-contralateral drop. Kinetic variables were ankle moment in the frontal plane and knee and hip moments in the frontal and transverse planes. Kinematic and kinetic data were computed in the joint coordinate system [26] following the mediolateral, anteroposterior and longitudinal Cardan's sequence [27]. Joint moments were calculated using the inverse dynamic procedure, normalized to body mass (kg), and reported in Nm/kg. Internal joint moments are reported throughout this study. Joints angular displacement and internal moments were normalized to 101 data points, one for each percentage of the stance phase. The coefficients of multiple correlation for these variables varied between 0.74 and 0.99, which means good to excellent reliability.

2.4. Data analysis

2.4.1. Principal component analysis (PCA)

PCA is the recommended choice as a first step for gait and running time series data reduction [28] without loss of temporal information, which generates independent principal components and scores [29] that were used to test the hypothesis of the present study [30,31]. PCA was performed, separately, on the six kinematic and five kinetic variables during the stance phase. Therefore, PCA was performed on 11 separate 38 × 101 data matrices (19 participants x 2 conditions x 101-time samples throughout the stance phase). Each row represented the time-series of each participant in each condition, and each column represented the values of the variable at each percentage of the stance phase.

The covariance matrix was calculated from the mean-centered of each original data matrix. Following this, an eigen value decomposition of this covariance matrix was performed by the PC model Z = [UtX], where U is the transformation matrix that realigned the original data into a new coordinate system. The columns of U were the eigenvectors of the covariance matrix of the original dataset and were designated PC loading vectors [32]. The PCs were obtained and ordered according to the amount of variance explained in the original data. A criterion of 90% of variance explained was used to determine the number of PCs to retain for data analysis [33,34]. The scores of each PC of a particular variable retained for analysis (PC_k) were computed as the sum of the products of the centered scores pi (i = 1–101) and its correspondent coefficient ui (i = 1–101) in the PC loading vector. Then, if more than one PC demonstrated differences for a specific variable, only the PC describing the largest amount of variance was reported.

2.4.2. Statistical analysis and interpretation of the PC-scores

The Shapiro-Wilk test demonstrated that the scores of the PCs retained for analysis demonstrated normal distribution. These scores were compared between conditions using dependent t-tests. The significance was set at α = 0.05 for the ankle and pelvis comparisons and at 0.025 for the knee and hip comparisons, since frontal and transverse planes were compared. The effect sizes (i.e. r-value) of the comparisons with statistically significant differences were also calculated as follows: $r = \sqrt{\frac{t^2}{t^2 + df}}$, where t is the t-value and df is the degree of freedom [35].

We used the method of single component reconstruction to interpret the differences between conditions in PC-scores [36]. First, the time-series representing the insole and control conditions pattern of variance on the specific PC were plotted in the same graph (Fig. 4). The time-series representing the insole and the control conditions correspond to a high or low value of the PC-score, depending on which condition had higher or lower scores on that specific PC. These time-series were calculated by first multiplying one standard deviation of the corresponding PC-scores by the PC loading vector, and then adding (high) or subtracting (low) the resulting product to the sample mean time-series [36]. Second, the portions of the stance phase that substantially contributed to the biomechanical feature captured by the specific PC were defined based on the portions of the stance phase that had greater PC loading vector magnitude (defined by vertical dashed lines and shaded areas in Figs. 4 and 5) [36]. Portions of the stance phase with loading vector magnitude equal or greater than half of the peak-loading vector for that specific PC were shaded. Finally, the differences between the time-series representing the insole and control conditions on the shaded areas in the graphs were analyzed in order to interpret the meaning of the differences between conditions.

3. Results

3.1. Characterization of the participants and running temporal-spatial measures

The participants' average weekly training volume corresponded to 20.9 km (SD 8.95), and the shank-forefoot alignment was 15.40° (SD 4.30) of varus. The control and insole conditions showed an average running speed of 3.29 m/s (SD 0.28) and 3.30 m/s (SD 0.37), respectively, and these differences were not statistically significant (p = 0.807).

3.2. Control condition versus insole condition

The comparisons of the PC scores between insole and control conditions demonstrated that seven PCs were statistically different between the two conditions, and these results are described in Table 1. The loading vectors of these PCs and the time-series representing the pattern of variance of the control and insole conditions, either high (+1SD) or low (-1SD) PC-scores, depending on the control and insole conditions mean scores, are shown in Figs. 4 and 5. The insole reduced ankle eversion angle, reduced ankle inversion moment, increased knee range of motion in the frontal plane during early stance and anticipated knee adduction during late stance, reduced knee range of motion in the transverse plane, increased knee range of internal rotation moment during the first half of stance, reduced hip adduction and reduced hip range of motion in the transverse plane (marginal effect).

4. Discussion

Most of our hypotheses were confirmed, except for the hip moment and pelvis movement. The insole reduced ankle eversion, reduced knee and hip range of motion in the transverse plane, increased the knee range of motion in the frontal plane and reduced hip adduction during the stance phase of running. In addition, the insole reduced the ankle inversion moment and increased the difference between the knee internal rotation moment in early stance and midstance phases. There were no effects of the insole on the hip moments in the frontal and transverse planes and on pelvis movement in the frontal plane.

During the insole condition, the participants demonstrated reduced ankle eversion and knee and hip range of motion in the transverse plane. Foot pronation is an expected and desired movement during the loading response phase of running [37]. However, excessive foot pronation overloads structures responsible for controlling this movement, such as foot [7,38], ankle [39] and hip muscles [40] and foot connective tissues [41]. In addition, excessive foot pronation may compromise foot lever arm function during late stance and consequently hamper ankle push-off [37,42]. Increased knee and hip movement in the transverse plane is also associated to lower limb injuries in runners [5,38,43]. Therefore, the medially wedged insole was able to reduce excessive foot pronation and related proximal joint movement patterns that are associated with lower limb injuries in runners.

The insole increased knee range of motion in the frontal plane and reduced hip adduction during the stance phase. This finding is in accordance with the findings of a previous study demonstrating that hip adduction is correlated with foot pronation during the stance phase of running of runners with patellofemoral syndrome [44]. Therefore, the reduction in foot pronation caused by the insole help to explain why the insole also reduced hip adduction. Increased hip adduction during running is associated with lower limb injuries, such as iliotibial band syndrome [45] and patellofemoral pain syndrome [46]. Therefore, the insole was able to modify hip motion patterns that are related to lower limb injuries. On the other hand, the insole increased knee range of motion in the frontal plane, which is non-desirable, since it isalso related to the occurrence of knee injuries [45].

The medially wedged insole reduced ankle inversion moment and

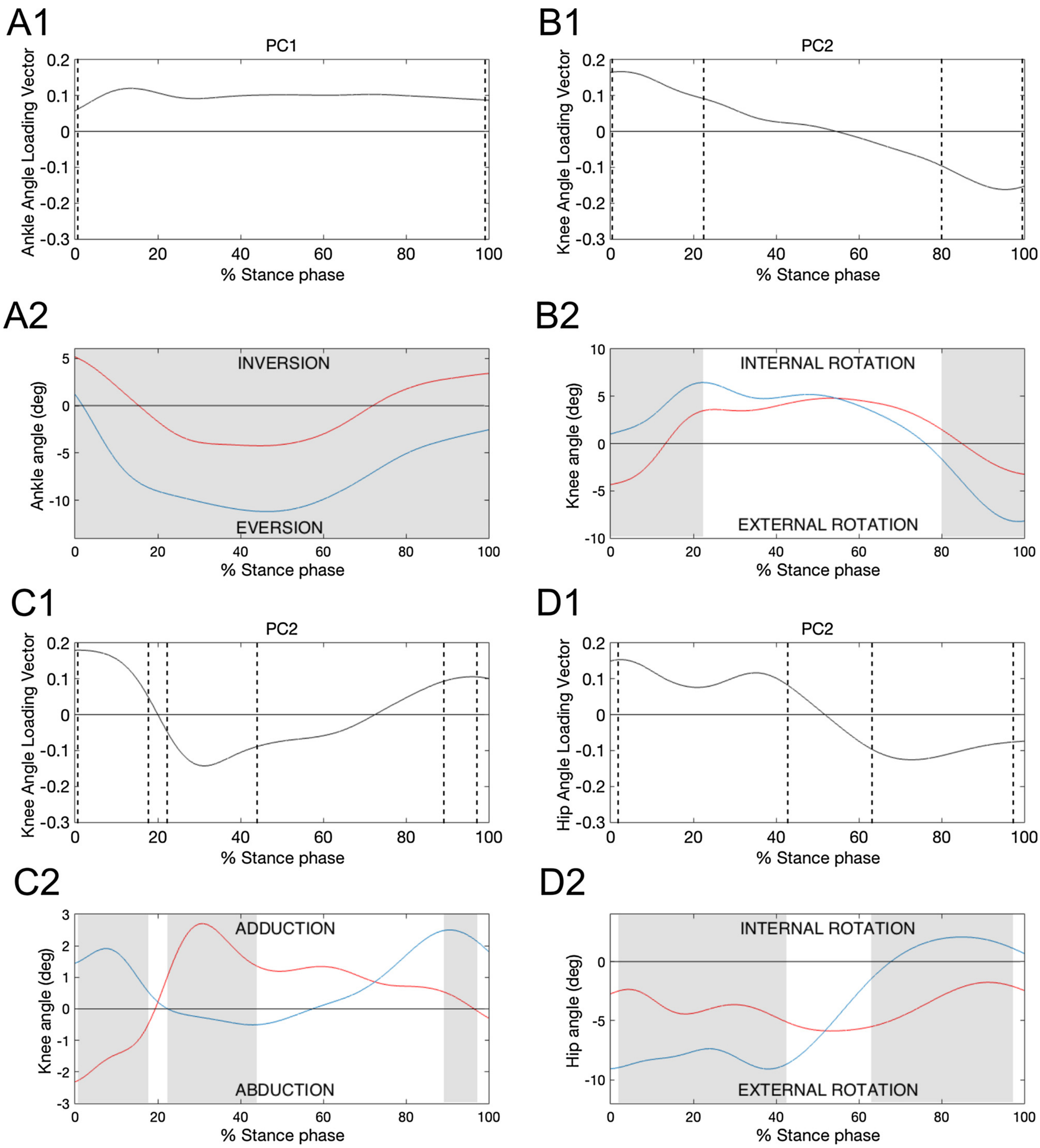

Fig. 4. Control versus insole conditions differences in the ankle, knee and hip angles demonstrated by the statistical comparisons. Shown in the figures are the time-series that represent high and low principal component (PC) scores for the indicated measure. In all cases, the time-series that represents the PC score (i.e. high or low PC score) that characterizes the insole condition is shown as a red line; the time-series that represents the PC score that characterizes the control condition is shown as a blue line. The shaded areas demonstrate the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed between conditions. The shaded areas were defined based on the magnitudes of the PC loading vector. The ankle inversion angle PC1 (A1 and A2); knee internal rotation angle PC2 (B1 and B2); knee adduction angle PC2 (C1 and C2); and hip internal rotation angle PC2 (D1 and D2) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.).

increased the difference between knee internal rotation moment during early stance and midstance phases. The reduced ankle inversion moment indicates that the insole reduced the demand on the ankle invertor muscles, such as tibialis anteriorand tibialis posterior muscles, which act eccentrically to support the plantar arch during this phase. These muscles have small cross-sectional areas. Muscle cross-sectional area is positively correlated with the muscle's capacity to absorb energy [47]. Therefore, these muscles have reduced capacity to control increased foot pronation caused by other factors, such as foot varus alignment and hip muscles weakness. It has been demonstrated that runners with medial tibial stress syndrome present increased demand on ankle invertor muscles [8,48]. Therefore, medially wedged insoles are able to reduce demand on muscles frequently overloaded in runners with lower limb injuries in runners. However, the increased difference between the

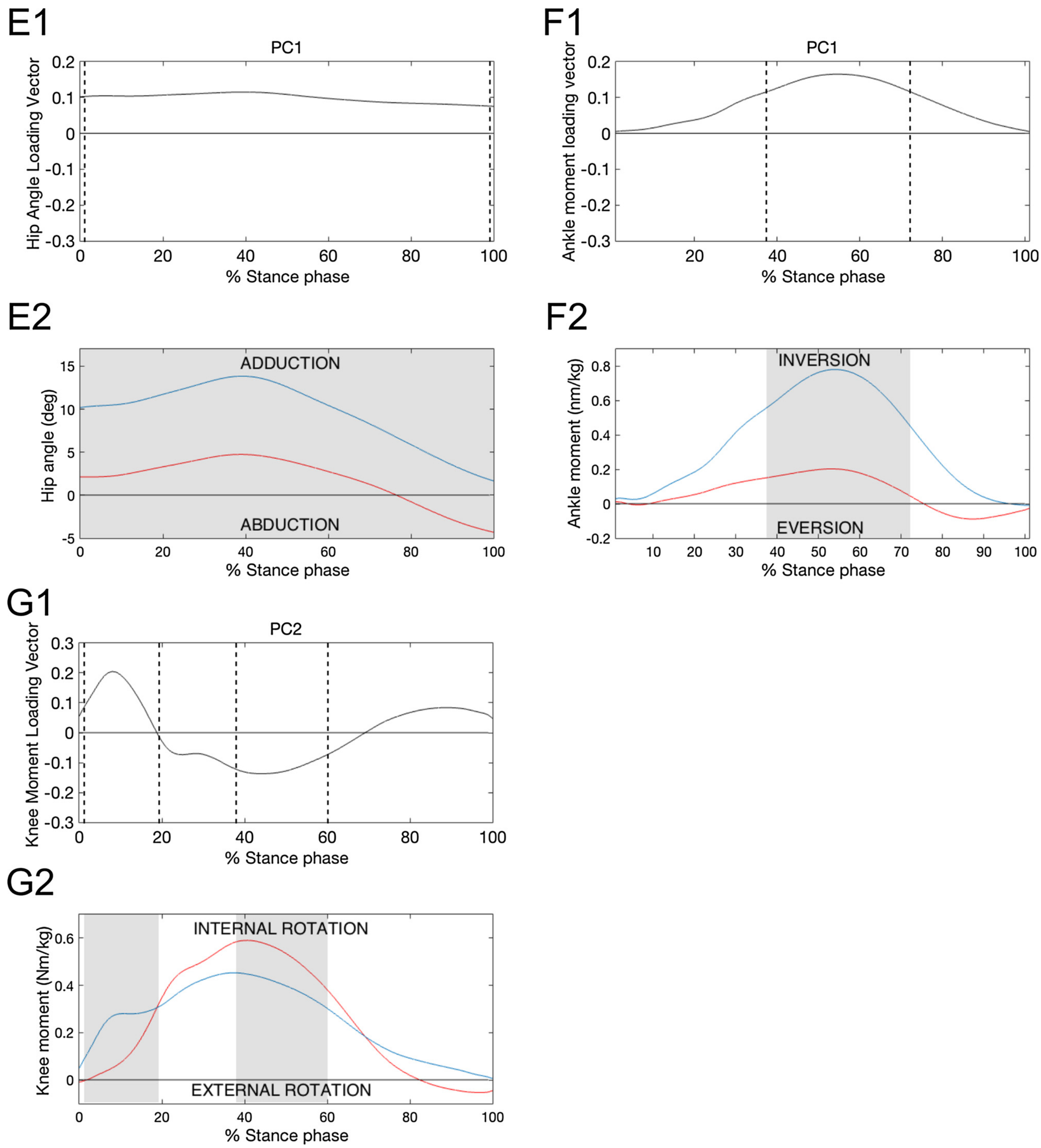

Fig. 5. Control versus insole conditions differences in the hip adduction angles and ankle and knee moments demonstrated by the statistical comparisons. Shown in the figures are the time-series that represent high and low principal component (PC) scores for the indicated measure. In all cases, the time-series that represents the PC score (i.e. high or low PC score) that characterizes the insole condition is shown as a red line; the time-series that represents the PC score that characterizes the control condition is shown as a blue line. The shaded areas demonstrate the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed between conditions. The shaded areas were defined based on the magnitudes of the PC loading vector. The hip adduction angle PC1 (E1 and E2); ankle inversion moment PC1 (F1 and F2); and knee internal rotation moment PC2 (G1 and G2) (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.).

knee internal rotation moment in early and midstance phases might be an acute non-desired effect of the use of insoles, since, during gait, it was related to knee osteoarthritis progression [49]. Among the studies that investigated the effects of medially wedged insoles on lower limb biomechanics during running [13–18,50,51], only three reported that the participants had excessive foot varus alignment or excessive foot pronation during the stance phase of running [16,50,51]. However, two of these studies [50,51] used static measures to identify excessive foot pronation, which does not guarantee that the participants had excessive foot pronation during the stance phase. Moreover, Mündermann et al (2003) [15] did not use foot varus alignment as an inclusion criterion and did not investigate the effects of the medially wedged insole on the hip kinematics in the frontal and transverse planes.

Although the participants had different degrees of forefoot and

Table 1
Principal components (PC) that demonstrated significant differences between control and insole conditions. Interpretation of the differences in the PC-scores is also provided.

Measure	PC	*p-value*	Effect size	Interpretation based on the pattern of the insole condition
Ankle inversion-eversion angle	1	0.003	0.63	Reduced eversion magnitude throughout stance phase
Knee internal-external rotation angle	2	0.012	0.55	Reduced knee range of motion in the transverse plan during stance phase
Knee abduction-adduction angle	2	0.018	0.52	Increased knee range of motion in early stance and had earlier knee adduction peak
Hip internal-external rotation angle*	2	0.031	0.48	Reduced hip range of motion in the transverse plane during stance phase
Hip adduction-abduction angle	1	0.024	0.50	Reduced hip adduction throughout stance phase
Ankle moment in the frontal plane	1	0.012	0.55	Reduced ankle inversion moment throughout stance phase
Knee moment in the transverse plan	2	0.012	0.55	Increased difference between the knee internal rotation moment in the early stance and midstance

* marginal effect considering alpha of 0.025.

rearfoot varus alignment, we used the same degree of wedge for all participants. Therefore, the effects of medially wedged insoles may be enhanced by using customized insoles. In addition, the insole used in the control condition had also a semi-rigid arch support, which influences kinematic and kinetic parameters evaluated in the study, such as ankle eversion [52]. Thus, the effects of the insole condition were due to the addition of the medialwedge at the forefoot and rearfoot, since both insoles had arch support. Moreover, the effects of soft tissue artifact on transverse plane kinematics and kinetics during running, and the effects of using markers on the shoe on foot kinematics can be substantial [53]. Finally, the findings of this study might not reflect long-term effects of medially wedged insoles and do not allow conclusions regarding the effects on pain and performance levels of runners with injuries.

5. Conclusions

This study demonstrated that an insole with 7° of medial wedge at the forefoot and rearfoot reduced ankle eversion, ankle inversion moment, knee and hip range of motion in the transverse plane and hip adduction, but increased knee range of motion in the frontal plane and increased the difference between knee internal rotation moment in early stance and midstance phases of runners with increased shank-forefoot varus alignment and excessive foot pronation. Therefore, insoles with 7° medial wedges at the forefoot and rearfoot are able to modify motion and moments patterns that are related to lower limb injuries in runners with increased foot pronation and shank-forefoot varus alignment with some non-desired effects on the knee motion in the frontal plane.

Declaration of Competing Interest

None.

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Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.09.023.

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