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Increased unilateral foot pronation affects lower limbs and pelvic biomechanics during walking

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ABSTRACT

Background: Increased unilateral foot pronation may cause biomechanical changes on the lower limbs during gait. We investigated the effects of increased unilateral foot pronation on the biomechanics of lower limbs and pelvis during gait.
Methods: Kinematic and kinetic data of 22 participants were collected while they walked wearing flat and laterally wedged sandals. Principal omponent analysis was used to compare differences between conditions.
Findings: Wearing the wedged sandal on the ipsilateral side increased ankle eversion moment ($p < 0.001$; effect size = 0.97); rearfoot eversion angle ($p < 0.001$; effect size = 0.76); shank internal rotation ($p = 0.009$; effect size = 0.53); increased and reduced knee internal rotation angle during early and late stance, respectively ($p < 0.001$; effect size = 0.89); increased femur internal rotation ($p = 0.005$; effect size = 0.90); reduced hip internal rotation moment during late stance ($p = 0.001$; effect size = 0.68); and increased pelvic ipsilateral drop ($p = 0.02$; effect size = 0.48) of the ipsilateral side. Wearing the wedged sandal on the contralateral side increased pelvic contralateral drop ($p = 0.001$; effect size = 0.63); hip adduction moment throughout stance ($p = 0.027$; effect size = 0.46); and increased and reduced the knee adduction moment in early and late stance, respectively ($p < 0.001$; effect size = 0.79).
Interpretation: The increased lower limb internal rotation caused by the wedged sandal reinforces the assumption that rearfoot eversion is coupled with shank internal rotation. The increased pelvic contralateral drop caused by the wedged sandal on the contralateral side may explain the increased hip and knee adduction moments on the ipsilateral side. Increased unilateral foot pronation causes biomechanical changes on both lower limbs that are associated with the occurrence of injuries.

1. Introduction

Increased foot pronation causes biomechanical changes at the lower limbs, which may result in musculoskeletal injuries at the proximal joints [1,2]. Previous study had shown that inadequate forefoot alignment at ground contact could produce large pronation torques that result in increased magnitude and duration of pronation during walking [3]. Following this rationale, Souza et al. [4] demonstrated that walking using lateral wedges under the forefoot increases rearfoot eversion and shank and hip internal rotation angles during the stance phase. Our work builds on these insights by examining the effects of increased unilateral foot pronation on knee and hip transverse plane moments and pelvic kinematics, since previous studies have demonstrated the occurrence of asymmetries in foot pronation in young [5] and elderly people [6].

The pelvic motion is dependent on the interaction of the lower limbs [7]. Therefore, it is logical to hypothesize that increased unilateral foot pronation may also influence the biomechanics of the opposite lower limb. Research had demonstrated that during

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quiet standing, foot pronation increases pelvic ipsilateral drop [8]. If that coupling mechanism remains true for walking, unilateral foot pronation may increase pelvic drop and consequently increases contralateral knee adduction moment [9], which is associated with knee ostearthritis progression [10]. In addition, unilateral foot pronation had been associated to low back pain [6], which reinforces the need to understand the biomechanical effects of unilateral foot pronation.

In order to investigate the effects of foot pronation during walking, different strategies have been implemented on shoes [11], foot orthoses [12] and sandals [13]. Specifically for studies using segmented foot models, the use of sandals seems to be more appropriate, since it was demonstrated that markers placed on shoes overestimate foot segments motion [14]. Regardless of the method chosen, it is usual to make assumptions about the effects of increased foot pronation based on a small set of biomechanical variables, such as the ipsilateral knee adduction moment [15]. However, considering that the influence of increased unilateral foot pronation on pelvic kinematics may also affect the biomechanics of the contralateral lower limb [16], more information about the effects of increased unilateral foot pronation on the mechanics of the lower limbs is necessary.

Therefore, the purpose of this study was to investigate the effects of increased unilateral foot pronation on the biomechanics of the lower limbs during the stance phase of walking. We hypothesized that increased foot pronation will increase ipsilateral lower limb internal rotation angles and ipsilateral pelvic drop and reduce internal rotation moments of the ipsilateral knee and hip. In addition, hip and knee adduction moments of the contralateral lower limb were expected to increase during early stance.

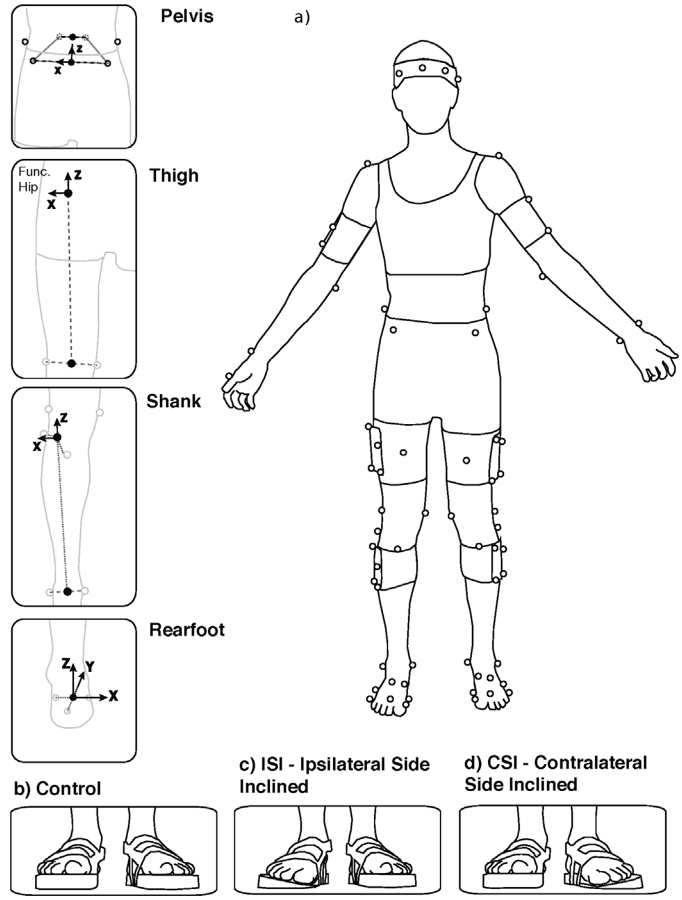

Fig. 1. Marker placement and segments coordinate systems (a); control condition (b); ipsilateral side inclined condition (c) and contralateral side inclined condition (d).

2. Methods

2.1. Participants

Sample size was determined as the number of participants necessary to reach a statistical power of 80%, with a significance level of 0.05, considering an expected moderate effect size ($d = 0.6$) [17]. Twenty-two healthy subjects (10 females, 12 males) with an average age, mass and height of 25 years (SD 4.5), 71.7 kg (SD 11.3) and 175 cm (SD 8), respectively, participated in the study. The inclusion criterion was no history of surgery or injuries to the lower limbs or to the lumbar-pelvic complex in the last year. Each participant signed a consent form approved by the university's Ethics Research Committee.

2.2. Procedures

Initially, the heights and masses of the participants were obtained. Subsequently, gait data were recorded at 200 Hz using 12-camera motion capture system (Oqus 4, Qualisys, Gothenburg, Sweden) and six tandem force platforms (Custom BP model, AMTI, Massachusetts, USA). The force platforms registered ground reaction force (GRF) data at a frequency of 1000 Hz, which was subsequently resampled to 200 Hz to match the motion capture data.

Anatomical markers and clusters of tracking markers were used to determine the coordinates of the whole body using data obtained with the participant in a standing position on flat sandals (static trial) (Fig. 1a). Kinematics and kinetic data were collected in three different conditions:

1) control condition: the participant walked wearing flat sandals (Fig. 1b);
2) ipsilateral side inclined condition: flat sandal on the left and a wedged sandal on the right foot (Fig. 1c);
3) contralateral side inclined condition: wedged sandal on the left and flat sandal on the right foot (Fig. 1d).

Although we collected data from the whole body, all results presented are only from data of the right lower limb on the three different conditions.

According to the methods described by Souza et al. [4], the wedged sandals were flat at the rearfoot and 10° laterally wedged (medially depressed) under the forefoot (Fig. 1c and d) to simulate a forefoot varus deformity, which has been shown to affect the magnitude and duration of pronation during walking [3]. Two sizes of sandals for each condition with the metrics described in the supplementary Fig. S1 were available. The base of the sandals was made of high-density ethylene vinyl acetate and was attached to the participants' feet with Velcro™ straps. The participants walked at their self-selected normal speed, performing six trials per condition along a 15-m distance. The order of data collection was randomized. Before data collection in each condition, the participants walked for approximately 1 min for familiarization with the pair of sandals.

Supplementary material related to this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2014.10.025.

2.3. Data reduction

Synchronized raw kinematic and kinetic signals were processed using Visual 3D (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 6 Hz [18] and 18 Hz, respectively. Heel contact and toe-off were determined automatically in Visual 3D using the vertical GRF with a threshold of 20N. Forefoot inversion–eversion with respect to the rearfoot was calculated to ensure that the wedged sandal increased forefoot eversion. Gait kinematics included: (1) rearfoot inversion–eversion (antero-posterior axis) with respect to the shank; (2) shank internal–external rotation (longitudinal axis) represented by the motion of the shank relative to the lab (3) femur internal–external rotation, represented by the motion of the thigh relative to the lab; (4) knee internal–external rotation, represented by the motion of the shank relative to the thigh; (5) hip internal–external rotation, represented by the motion of the thigh relative to the pelvis; (4) pelvic ipsilateral–contralateral drop (antero-posterior axis) with respect to the lab [9]. Kinetic data included ankle inversion moment and knee and hip adduction and internal rotation moments. Kinematic and kinetic data were calculated based on the Cardan flexion/extension, adduction/adduction, internal/external rotation sequence [19]. Joint moments were calculated using the inverse dynamic procedures, normalized to body mass (kg), reported in Nm/kg. External moments were reported throughout the text. Gait measures were normalized to 101 data points, one for each percentage of the stance phase.

2.4. Data analysis

2.4.1. Principal component analysis (PCA)

PCA was performed to 11 gait measures arranged in 11 separate 66×101 data matrices (22 subjects $\times$ 3 conditions $\times$ 101 time points per stance phase). The principal components (PCs) summarize and capture the unique patterns of variation of each gait measure allowing comparisons between conditions without loss of temporal information [20].

Data related to each measure m were organized in an $n \times p$ matrix X_m. Each row in the matrix X_m represented a waveform m for one participant in one of the three experimental conditions. Each column represented the time samples of measure m at one particular instant for all participants in each condition.

The first step of PCA was to subtract off the mean across subjects and conditions from each time sample in matrix X_m. Subsequently, the co-variance matrix of X_m (C_{X_m}) was computed. The off diagonal terms of C_{X_m} represented the co-variance between all possible pairs of time-samples and its diagonal terms captured the variance in each time sample. The PCs are a set of orthogonal row vectors $U\{u_1 \ldots u_{101}\}$, referred to as loading vectors, which transforms C_{X_m}, so as to maximize its variance and minimize its co-variance. Mathematically, the loading vectors (or PCs) are the eigenvectors of C_{X_m}, which are ordered as a function of the amount of variance explained.

The scores of each PC of a particular measure retained for analysis (PC_k) were computed as the sum of the products of the centred scores p_i (i = 1–101) and its corresponding coefficient u_i (i = 1–101) in the PC loading vector. The scores were the sought-after summary measure that allowed comparisons between conditions. A criteria of 90% of variance explained was used to determine the number of principal components (PCs) to retain for data analysis [21,22]. When two or more PCs retained for analysis demonstrated differences for a specific gait measure, only the PC describing the largest amount of variance was reported.

2.4.2. Statistical analysis of the PC-scores

The scores of the PCs retained for analysis were tested for normal distribution and homogeneity of variance using Shapiro–Wilk and Levene's test, respectively, and then compared between conditions using one-way repeated measures analysis of variance (ANOVA) with pre-planned contrasts between the control condition and the ipsilateral side inclined condition for the following gait measures: (1) ankle inversion moment; (2) rearfoot inversion angle; (3) shank and femur internal rotation angles; (4) knee and hip internal rotation moments and angles; (5) pelvic ipsilateral drop. Pre-planned contrasts were also performed between the control and the contralateral side inclined conditions for the following gait measures: (1) pelvic ipsilateral drop; (2) hip and knee adduction moments. The significance was set at $\alpha = 0.05$ except for the pre-planned comparisons of the pelvic ipsilateral drop that α was set at 0.025 following Bonferroni correction. The effect sizes (e.g. *rcontrast*) of the contrasts with statistically significant differences were also calculated as follows: $r = \sqrt{F(1, df_R)/(F(1, df_R) + df_R)}$ where F is the F-value and df is the degree of freedom [23].

2.4.3. Interpretation of PCs

In the interpretation of the biomechanical meaning of a single PC, the goal is to isolate the variance captured by the PC of interest. This can be visualized on a single figure by plotting two waveforms, $\hat{x}_H$ and $\hat{x}_L$, representing waveforms corresponding to a high and low value of the PC by adding and subtracting a scalar multiple of the loading vector, u_R, to the average waveform, $\bar{x}$. A convenient scalar multiple is one standard deviation (SD) of the corresponding PC scores, SD($\vec{z_i}$) The high and low PC waveforms differ only in the feature captured by a single PC [24]. Thus, high and low PC waveforms that include only the contribution of the Rth PC, denoted PC_R, can be computed:

$$\hat{x}_H = \bar{x} + u_R \times SD(\vec{z_i}) \tag{1}$$

$$\hat{x}_L = \bar{x} - u_R \times SD(\vec{z_i}) \tag{2}$$

where $\hat{x}_H$ is the reconstructed high waveform for PC_R; $\hat{x}_L$ the reconstructed low waveform for PC_R; $\bar{x}$ is the mean temporal waveform for all subjects; u_R the pattern of variance, or loading vector, for PC_R; SD($\vec{z_i}$) is the SD of the PC_R scores [25].

Table 1
Principal components (PCs) that demonstrated differences between ipsilateral side inclined and control conditions. Percentage of variance explained and an interpretation of each PC are also provided.

Measure	PC	Variance explained (%)	*p-value*	Effect size	Interpretation based on the effects of the ipsilateral side inclined condition
Ankle inversion–eversion moment	1	87.6	<0.001	0.97	Had greater ankle eversion moment between 15 and 90% of stance.
Rearfoot inversion–eversion angle	1	70.8	<0.001	0.76	Had greater rearfoot eversion throughout stance.
Shank internal–external rotation angle	1	78.8	0.009	0.53	Had greater shank internal rotation throughout stance.
Knee internal–external rotation moment	1	62.4	<0.001	0.87	Had smaller knee internal rotation moment in late stance.
Knee internal–external rotation angle	2	13.5	<0.001	0.89	Had greater knee internal rotation in early stance and reduced knee internal rotation in late stance.
Femur internal–external rotation	1	69.8	0.005	0.90	Had greater femur internal rotation throughout stance.
Hip internal–external rotation moment	2	23.9	0.001	0.68	Had smaller hip internal rotation moment in late stance.
Hip internal–external rotation angle	1	63.4	0.031	0.45	Had greater hip internal rotation throughout stance phase.
Pelvic ipsilateral–contralateral drop	2	27.3	0.02	0.48	Had greater ipsilateral pelvic drop in late stance.

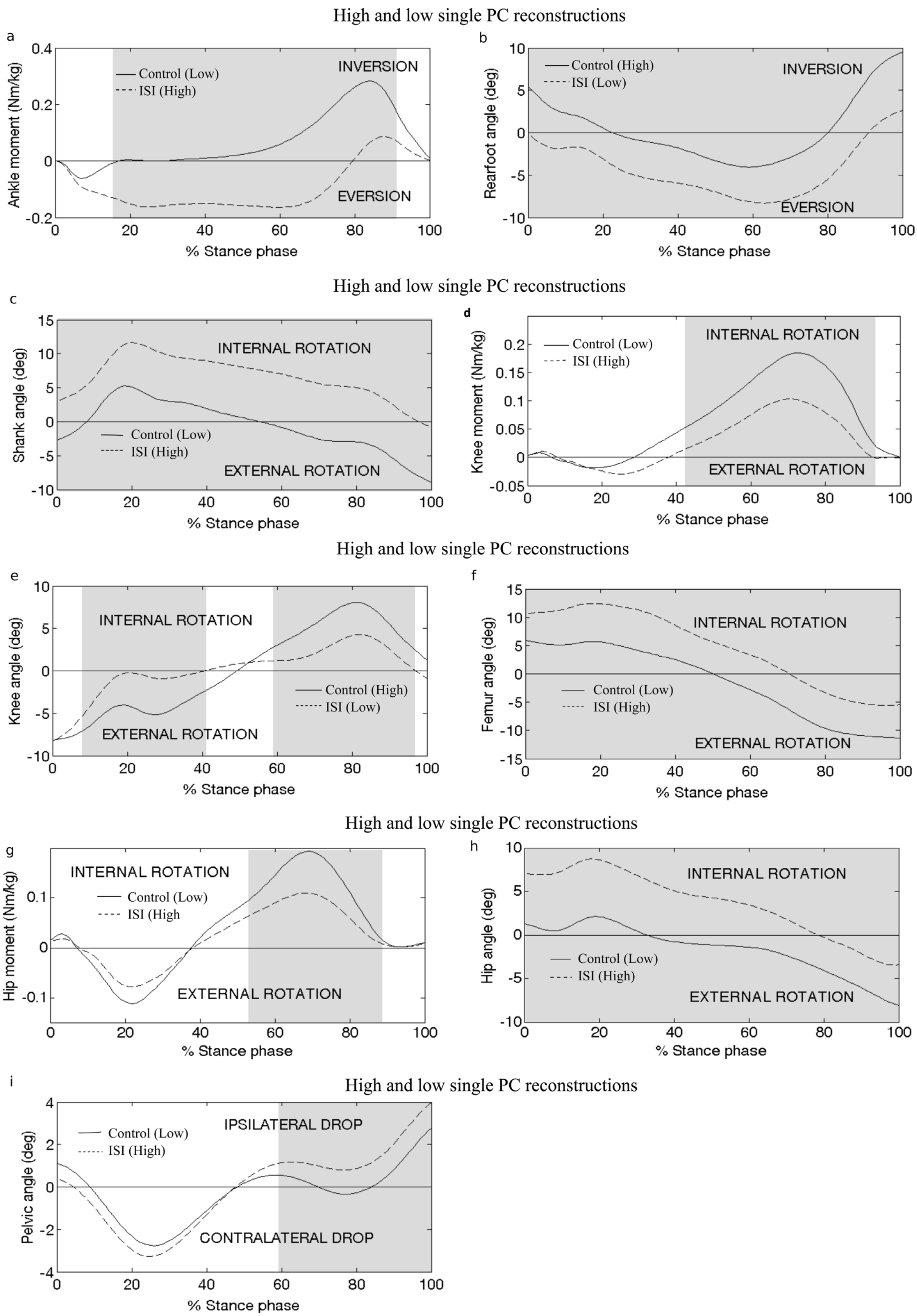

Fig. 2. Control and ipsilateral side inclined conditions differences demonstrated by the ANOVA. Shown in the figures are the waveforms that represent high and low principal component (PC) scores for the indicated measure and PC. In all cases, the waveform that represents the PC score (i.e. high or low PC score) that characterizes the ipsilateral side inclined condition is shown as a dashed line; the solid line is the waveform that represents the control condition PC score. The shaded areas demonstrate the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed. The ankle inversion moment PC1 (a); rearfoot inversion angle PC1 (b); shank internal rotation PC1 (c); knee internal rotation moment PC1 (d); knee internal rotation angle PC2 (e); femur internal rotation PC1 (f); hip internal rotation moment PC2 (g); hip internal rotation angle PC1 (h); pelvic ipsilateral drop PC2 (i). ISI: ipsilateral side inclined condition; CSI: contralateral side inclined condition.

Table 2
Principal components (PCs) that demonstrated differences between contralateral side inclined and control conditions. Percentage of variance explained and an interpretation of each PC are also provided.

Measure	PC	Variance explained (%)	*p*-value	Effect size	Interpretation based on the effects of the contralateral side inclined condition
Pelvic ipsilateral–contralateral drop	1	64.3	0.001	0.63	Had greater pelvic contralateral drop during early stance.
Hip adduction–abduction moment	1	74.4	0.003	0.61	Had greater hip adduction moment throughout stance.
Knee adduction–abduction moment	3	8.0	0.002	0.63	Had greater knee adduction moment in early stance.

3. Results

3.1. Gait speed and forefoot eversion

The control, ipsilateral side inclined and contralateral side inclined conditions showed an average gait speed of 1.44 m/s (SD 0.15), 1.45 m/s (SD 0.16) and 1.45 m/s (SD 0.16), respectively, and these differences were not statistically significant ($p = 0.69$). Wearing the wedged sandal increased forefoot eversion by 5.27° (SD 0.42) throughout stance phase ($p < 0.001$) when compared to the flat sandal.

3.2. Gait variables

The results of ANOVA demonstrated 11 PCs that were significantly different between the three conditions: ankle inversion moment ($F = 356.7$; $p < 0.001$); rearfoot inversion angle ($F = 17.7$; $p < 0.001$); shank internal rotation angle ($F = 5.8$; $p = 0.006$); knee adduction moment ($F = 18.6$; $p < 0.001$); knee internal rotation moment ($F = 24.2$; $p < 0.001$); knee internal rotation angle ($F = 26.6$; $p < 0.001$); femur internal rotation angle ($F = 7.7$; $p = 0.005$); hip adduction moment ($F = 4.2$; $p = 0.022$); hip internal rotation moment ($F = 9.6$; $p < 0.001$); hip internal rotation angle ($F = 3.7$; $p = 0.032$); and pelvic ipsilateral drop ($F = 7.5$; $p = 0.002$).

3.3. Ipsilateral side inclined condition versus control condition

The pre-planned comparisons between ipsilateral side inclined and control condition identified nine PCs that were statistically different (Table 1). Gait waveforms represented by high (+1 SD) and low (−1 SD) PC scores for each significant PC are shown in Fig. 2. On the graphs, the shaded areas identify the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed between conditions. This shaded area was defined based on the portions of the stance phase that presented greater PC loading vector magnitudes.

3.4. Contralateral side inclined condition versus control condition

The pre-planned comparisons between contralateral side inclined condition and control conditions identified 3 PCs statistically different (Table 2). Waveforms that represent high and low PC scores for each PC are shown in Fig. 3. The wedged sandal increased pelvic contralateral drop ($F = 13.7$; $p = 0.001$; Fig. 3a); increased hip adduction moment throughout stance phase ($F = 5.7$; $p = 0.027$; Fig. 3b); increased knee adduction moment during early stance and reduced knee adduction moment during late stance ($F = 35.9$; $p < 0.001$; Fig. 3c).

4. Discussion

This study demonstrated that unilateral increased foot pronation affects the biomechanics of the lower limbs during walking. The increased shank and femur internal rotation angles and decreased knee and hip internal rotation moments caused by the wedged sandal supports the existence of the coupling mechanism between rearfoot eversion and lower limb internal rotation [4,13]. In addition, the increased pelvic ipsilateral drop supports the hypothesis that increased foot pronation dynamically shortens the lower limb [1]. Wearing the wedged sandal increased hip and knee adduction moments of the contralateral lower limb, which may be explained by the pelvic contralateral drop [9].

The increased ankle eversion moment and rearfoot eversion angle caused by the ipsilateral wedged sandal were accompanied by increased shank and femur internal rotation angle throughout stance. The orientation of the subtalar joint axis links rearfoot eversion with lower limb internal rotation [26], as demonstrated by previous studies [4,13]. The wedged sandal increased knee internal rotation angle during early stance, but reduced knee internal rotation angle during late stance. Total knee range of motion is smaller than the total hip range of motion in the transverse plane. Therefore, it is possible that during late stance the knee was maximally internally rotated affecting the hip joint. This rationale is supported by the increased hip internal rotation throughout stance and by the difference in effect size between the increased femoral and shank internal rotation. This rationale is supported by the findings of Lafortune et al. [27] indicating rotation transference between tibia and femur.

Wearing the wedged sandal reduced the knee and hip internal rotation moments of the ipsilateral lower limb during late stance. During terminal stance, the GRF is antero-lateral and antero-medial to the hip and knee joint centres, respectively. In the transverse plane, at terminal stance, the GRF points postero-medially relative to both joint centres [28]. Because walking with lateral wedges laterally shifts and vertically aligns the GRF [29], shortening of the knee and hip GRF lever arms occurs [30]. This mechanism contributed to the smaller knee and hip internal rotation moments observed in the present study. Alternatively, it is possible that the increased hip internal rotation angle observed during the ipsilateral inclined condition may have affected the action of the hip external rotators muscles, through lengthening of the muscle fibres [31]. Therefore, smaller hip external rotation moments would be generated by the participants and consequently reduce the hip internal rotation moments computed based on the GRF, but this is only speculative at this point, since we did not evaluate muscle fibres length changes on this study.

In the present study, increased foot pronation dynamically shortened the ipsilateral lower limb during gait leading to increased ipsilateral pelvic drop [30]. Similar finding has been reported elsewhere [1]. Increased pelvic drop may contribute to the development of lower back pain [6], scoliosis and other pathological conditions in the lumbar spine [30].

The contralateral side inclined condition demonstrated increased hip and knee adduction moments. Contralateral pelvic drop causes a shift of the centre of mass towards the non-stance limb, increasing the hip and knee lever arm and consequently, the hip and the adduction moments [9]. The hypothesis that pelvic contralateral drop influences hip and knee adduction moment is supported by the findings of Lin et al. [32], which demonstrated that pelvic motion in the frontal plane is the gait measure that most significantly contribute to the medio-lateral displacement of the body centre of mass during gait. Increased knee and hip adduction moments are related to the development and progression of lower limb injuries [33]. Therefore, increased foot pronation of the opposite lower limb may help to explain increased knee and hip adduction moments in individuals with lower limb injuries, such as knee osteoarthritis [33].

Forefoot angle influences the duration and amplitude of foot pronation during gait [3]. Because the wedged sandal on the forefoot simulates the effects of forefoot varus alignment [4], individuals with different forefoot alignment may have been an additional source of variability in the present study. However, because we used a repeated measures design, possible differences in individuals' forefoot alignment affected all conditions, and therefore did not influence the differences observed among conditions. In addition, the results of the present study were

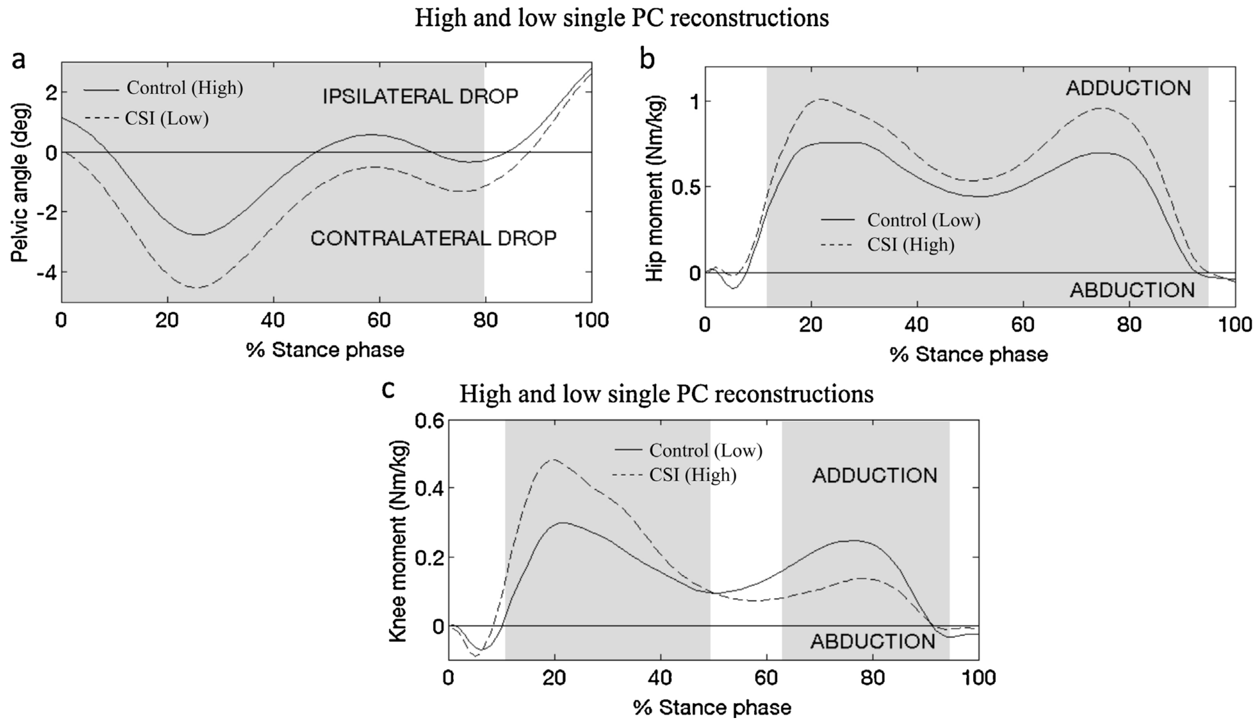

Fig. 3. Control and contralateral side inclined conditions differences demonstrated by the ANOVA. Shown in the figures are the waveforms that represent high and low principal component (PC) scores for the indicated measure and PC. In all cases, the waveform that represents the PC score (i.e. high or low PC) that characterizes the contralateral side inclined condition is shown as a dashed line; the solid line is the waveform that represents the control condition PC score. The shaded areas identify the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed. The pelvic ipsilateral drop PC1 (a); hip adduction moment PC1 (b); knee adduction moment PC2 (c).

focused on the immediate effects of unilateral increased foot pronation. However, these effects may depend on the individual's strategies to cope with increased unilateral foot pronation, which is affected by many different variables, such as knee joint laxity [34] and hip abductors and external rotators strength [35].

The findings of this study demonstrated that increased unilateral foot pronation caused by the use of wedged sandals affects the biomechanics of lower limbs during gait. In summary, increased foot pronation increased the ipsilateral lower limb internal rotation angles and reduced knee and hip internal rotation moments with increased pelvic ipsilateral drop. Increased foot pronation of the left side increased the hip and knee adduction moments of the right lower limb, which may be explained by the increased contralateral pelvic drop. These results should be considered when examining individuals demonstrating increased unilateral foot pronation. In addition, strategies to manipulate foot motion, such as the use of lateral wedges [15], should be cautiously implemented in light of the possible deleterious effects on the biomechanics of the lower limbs.

Conflicts of interest

We wish to confirm that there are no known conflicts of interest associated with this publication and there has been no significant financial support for this work that could have influenced its outcome.

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