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Effects of foot pronation on the lower limb sagittal plane biomechanics during gait

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ABSTRACT

Background: Increased foot pronation may compromise ankle plantarflexion moment during the stance phase of gait, which may overload knee and hip.

Research question: This study investigated the influence of increased foot pronation on lower limbs angular displacement, internal moments and power in the sagittal plane and ground reaction force and center of pressure displacement during the stance phase of gait.

Methods: Kinematic and kinetic data of 22 participants (10 women and 12 men) were collected while they walked wearing flat (control condition) and laterally wedged sandals to induce foot pronation (inclined condition). We used principal component analysis for data reduction and dependent *t*-test to compare differences between conditions with $\alpha = 0.05$.

Results: The inclined condition increased forefoot range of motion ($p < 0.001$; effect size = 0.73); increased ankle plantarflexion angle ($p < 0.001$; effect size = 0.96); reduced ankle plantarflexion moment in mid and terminal stance phases and delayed and increased ankle plantarflexion moment in late stance ($p < 0.001$; effect size = 0.72); increased range of ankle power during late stance ($p = 0.006$; effect size = 0.56); reduced knee range of moment ($p < 0.001$; effect size = 0.76); increased range of knee power in early stance and reduced knee power generation in late stance ($p = 0.005$; effect size = 0.56); reduced the anterior displacement of the center of pressure ($p < 0.001$; effect size = 0.82) and increased the ground reaction force in the anterior direction ($p = 0.003$; effect size = 0.60).

Significance: Increased foot pronation compromises lower limb mechanics in the sagittal plane during the stance phase of gait. These findings are explained by the fact that foot pronation increases foot segments flexibility and compromises foot lever arm function during the stance of gait.

1. Introduction

During the loading response phase of gait, the foot is more flexible because foot pronation places midtarsal joint in its loosed-packed position [1–3]. This increased foot flexibility improves its adaptability to different ground surfaces and also contributes to absorption of ground reaction forces (GRF) during early stance [1]. During late stance, the subtalar and midtarsal joints are locked due to foot supination, which reduces foot flexibility and consequently contributes to its lever arm function during push-off [4,5]. Different factors may cause foot pronation instead of supination during late stance, such as forefoot varus misalignment [6]. Pronation during late stance may hamper foot lever arm function and consequently reduce ankle internal plantarflexion moment during this phase. Considering that the ankle plantarflexor muscles and tendons are the primary generators of positive power during late stance [7], compromised foot lever arm function due to foot pronation may reduce ankle power generation and consequently impair early leg swing and acceleration of the body center of mass [8].

The hip flexor muscles also contribute, in a lesser amount, to the power generated by the lower limb during late stance [9]. Therefore, if the power generated at the ankle is reduced due to increased foot pronation, individuals might have to increase hip power generation to compensate for the reduced ankle power and consequently maintain early leg swing and body center of mass propulsion. In fact, there is empirical evidence in support of this contention [10–13]. Caputo and Collins [10] used a custom-built device emulator to vary the magnitude of ankle push-off power during gait. They demonstrated that, in fact, ankle power generation has an inverse relationship with hip power

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generation during late stance. Therefore, it is possible that individuals with increased foot pronation during late stance have to increase hip flexion moment and positive power to compensate for the reduced ankle plantarflexion moment and power caused by foot pronation. In the long-term, these increased hip moments and positive power may overload this joint and consequently contribute to the development of hip injuries, such as femoroacetabular impingement [14].

Increased foot pronation may also influence on lower limb kinematics in the sagittal plane. It has been demonstrated that, during terminal stance, subtalar plantarflexion is responsible for the onset of the ankle joint complex plantarflexion [15]. Foot pronation is characterized by subtalar dorsiflexion [16]. Therefore, increased foot pronation during late stance may delay or reduce ankle plantarflexion angle during this phase. In addition, the transition between foot pronation to supination and knee joint flexion to extension occur at approximately the same time during midstance. In this context, foot pronation past midstance might delay knee extension during this phase and consequently result in soft tissue stress, similar to what have been demonstrated during running [17]. Therefore, increased foot pronation may influence ankle, knee and hip biomechanics not only in the frontal [18–20] and transverse [21] but also in the sagittal plane. The main purpose of this study was to investigate the effects of increased foot pronation on the forefoot, ankle, knee and hip angular displacements, internal moments and power in the sagittal plane during the stance phase of gait. The secondary purpose was to investigate the effects of increased foot pronation on center of pressure (COP) displacement and GRF during the stance phase. The main hypothesis of this study was that increased foot pronation would reduce ankle plantarflexion moment and positive power and increase hip flexion moments and positive power during late stance.

2. Methods

2.1. Participants

Twenty-two healthy subjects (10 women and 12 men) with an average age of 25 ± 4.5 years, body mass of 71.7 ± 11.3 kg, and height of 1.75 ± 0.08 m participated in this study. Participants were eligible if the following inclusion criteria were met: not having undergone surgery on lower limbs or pelvic-lumbar complex in the last year and be of age between 20 and 40 years old. All participants signed a consent form approved by the university's Ethics Research Committee (6007495).

2.2. Procedures

We collected gait data using a 12-camera motion analysis system (Oqus 4, Qualisys, Gothenburg, Sweden) synchronized with six force platforms (Custom BP model, AMTI, Massachusetts, USA). A marker set was used to determine the coordinate systems of the pelvis, thighs and shanks [22] using data obtained with the participant wearing a pair of flat sandals in orthostatic position (Fig. 1). This marker set was also used on both feet for kinetics, but forefoot and rearfoot kinematics were computed using a multisegment foot model [23]. A trial with hip flexion/extension, adduction/abduction and circumduction was performed to calculate the hip joint center based on the algorithm developed by Schwartz and Rozumalski [24].

Using the same methods described in Resende et al. [21], gait data were collected in two different conditions: subject wore flat sandals on both feet (Control Condition) and subject wore a flat sandal on the left foot and a wedged sandal on the right foot (Inclined Condition). The wedged sandals were flat at the rearfoot and 10° laterally wedged (medially depressed) under the forefoot. This medial inclination has been shown to affect the duration and amplitude of subtalar pronation during walking [6,21]. In the present study, we used two sizes of sandals for each condition, with the specific dimensions described in a previous study [21]. The sandals' soles were made of high- density ethylene vinyl acetate and were attached to the participants' feet with Velcro®.

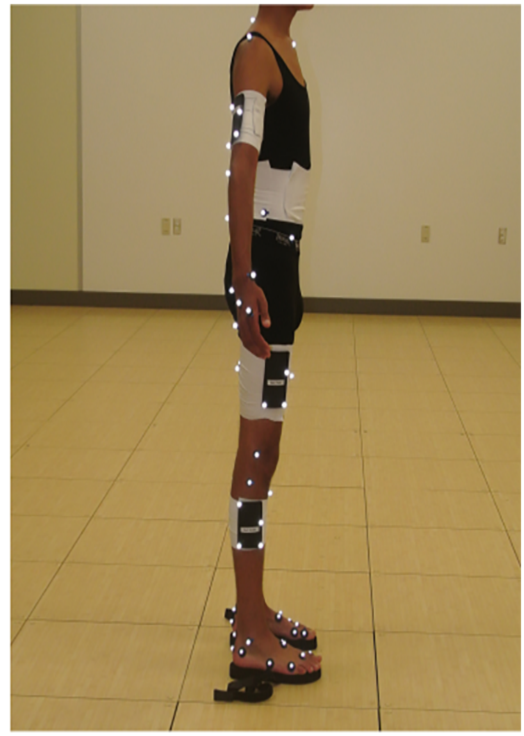

Fig. 1. Marker placement during the static data collection.

Before starting data collection, the participants walked for one minute to familiarize with the sandals. The order of data collection in both conditions was randomized. A total of six trials per condition, in which the subjects walked for 15 m at a self-selected speed, were performed and recorded.

2.3. Data reduction

Kinetic and kinematic data were processed using Visual 3D (C-motion, Inc., Rockville, USA). Low-pass fourth order Butterworth filters with cut-off frequency set at 6 Hz and 18 Hz were used for kinematic and kinetic data, respectively. We only analyzed the stance phase. Gait events (heel contact and toe-off) were automatically determined using the vertical GRF with a threshold of 20 N. Gait kinematics included: 1) forefoot dorsiflexion-plantarflexion with respect to the rearfoot (medio-lateral axis); 2) ankle dorsiflexion-plantar flexion, represented by the motion of the rearfoot relative to the shank (medio-lateral axis); 3) knee flexion-extension, represented by the motion of shank relative to the thigh (medio-lateral axis), 4) hip flexion-extension, represented by the motion of thigh relative to the pelvis (medio-lateral axis). Kinetic data were: dorsiflexion-plantar flexion moment of the ankle, flexion-extension moments of the knee, flexion-extension moments of the hip and ankle, knee and hip power in the sagittal plane. Kinematic and kinetic data were calculated based on the Cardan flexion/extension, abduction/adduction, internal/external rotation sequence [25]. We used three-dimensional inverse dynamic procedures to calculate joint moments [26], which were normalized by body mass and reported in NM/kg. Internal moments were reported throughout the text. Joint powers were also normalized to body mass (kg) and reported in W/kg. All of the gait variables were normalized to 101 points, one for each percentage of the stance phase. In addition, we also reported mean COP displacement (% of foot length) and peak GRF data (N) in the anterior direction during the stance phase. COP was converted into the foot coordinate system, with data normalized in the antero-posterior direction based on the participant's foot length.

2.4. Data analysis

2.4.1. Principal component analysis (PCA)

Extracting discrete variables from temporal series has the following limitations: i) loss of temporal information; ii) severe data reduction; iii) redundancy between the extracted discrete variables; and iv) difficulty to define the parameter to extract. Therefore, we chose PCA since it is the recommended choice as a first step for gait waveform data reduction [27], without loss of temporal information. The PCA generates independent principal components and scores [28] that were used for the hypothesis tests of this study.

PCA was performed in 10 separate 24 x 101 data matrices (22 subjects x 2 conditions x 101 time points per stance phase). 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 temporal series m for each participant on of each condition. Each column represented the time samples of measure m at one particular instant for all participants on both conditions. The procedures to compute the principal components (PCs) loading vectors have been described elsewhere [29,30]. The PCs were extracted in a hierarchical fashion based on the amount of variation they explained [31]. The three PCs explained the largest amount of variation in the original data were retained for data analysis [21]. When two or more PCs demonstrated differences in the PC-scores between conditions for a specific gait variable, we only reported the differences in the PC describing the largest amount of variance.

2.4.2. Statistical analysis and interpretation of the PC-scores

The scores of the PCs retained for analysis and the mean COP and peak GRF data were tested for normal distribution using Kolmogorov-Smirnov and Shapiro-Wilk tests, and then compared between conditions using dependent t-tests (all principal component scores and COP and GRF data demonstrated normal distribution). The significance was set at $\alpha = 0.05$. The effect sizes (i.e. *r-value*) of the comparisons with statistically significant differences were also calculated as follows: $r = \sqrt{\frac{t^2}{t^2 + d_f}}$, where t is the t-value and d_f is the degree of freedom [32].

The method of single component reconstruction was used to interpret the differences between conditions in PC-scores [33]. This method isolates the pattern of variance captured by the specific PC where the conditions differed, and had three steps. First, the waveforms representing the control and the inclined condition pattern of variance on the specific PC were plotted in the same graph (Fig. 2). The waveforms representing the control and the inclined 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 waveforms 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 waveform [33]. Second, the portions of the stance phase that contributed most to the biomechanical feature captured by the specific PC, and consequently to the differences between conditions, were defined based on the portions of the stance phase that had greater PC loading vector magnitude (defined by vertical dashed lines in Fig. 2) [33]. Third, the differences between the waveforms representing the control and the inclined conditions on the shaded areas in the graphs were analyzed in order to interpret the meaning of the differences between conditions in the PC-scores.

3. Results

3.1. Gait variables

The control and inclined conditions had an average gait speed of 1.44 m/s (SD 0.15) and 1.45 m/s (SD 0.16), respectively, and this difference was not statistically significant ($p = 0.69$). In addition, wearing the wedged sandal increased forefoot eversion by 5.27° (SD 0.42) during stance phase when compared to the flat sandals ($p < 0.001$), confirming that the inclined condition was able to induce foot pronation in order to test the hypothesis of this study.

The comparisons of the PC scores between control and inclined conditions demonstrated that six PCs had PC-scores different between conditions (Table 1). There were no differences in the knee and hip kinematics and in the hip moment and power between conditions. The loading vectors of these PCs and the temporal series representing the pattern of variance of the control and inclined conditions, either high (+1SD) or low (-1SD) PC-scores, depending on the control and inclined conditions' mean score, are represented (Fig. 2). In summary, the inclined condition had increased forefoot range of motion (Fig. 2A) and ankle plantarflexion (Fig. 2B) during the stance phase, reduced ankle plantarflexion moment during mid and terminal stance phases and delayed and increased ankle plantarflexion moment in push-off (Fig. 2C), increased range of ankle power during late stance (Fig. 2D), reduced range of knee moment throughout stance phase (Fig. 2E) and increased range of knee power during early stance and reduced knee power generation in late stance (Fig. 2F).

The inclined condition had smaller anterior displacement of the COP during the stance phase of gait in comparison to the control condition (Table 2). Therefore, the inclined condition reduced foot leverage in the sagittal plane during the stance phase of gait. In addition, the inclined condition increased the magnitude of the GRF in the posterior direction (Table 2).

4. Discussion

This study showed that foot pronation increased forefoot range of motion in the sagittal plane and the ankle plantarflexion angle during the stance phase. In addition, foot pronation reduced ankle plantarflexion moment during mid and terminal stance phases, delayed and increased ankle plantarflexion moment and increased range of ankle power in late stance. Moreover, the induced foot pronation reduced knee range of moment throughout stance phase, increased knee range of power in early stance and reduced knee power generation in late stance. Finally, foot pronation reduced the anterior displacement of the COP and increased the magnitude of the GRF in the anterior direction during stance. These findings might be explained by the fact that foot pronation increases foot segments flexibility and compromises foot lever arm function during late stance, allowing increased forefoot range of motion and compromising ankle plantarflexion moment and knee range of moment during stance.

The increased forefoot range of motion might be explained by the fact that increased foot pronation increases flexibility between foot segments [34]. As the foot COP displaced anteriorly during the loading response and midstance phases, the increased foot flexibility caused by the increased foot pronation allowed the forefoot to plantarflex relative to the rearfoot, which contributed to the increased forefoot range of motion in the inclined condition. In addition, the increased forefoot plantarflexion allowed the participants to fully contact the forefoot on the ground while wearing the wedged sandals, since only the anterior part of the sandal was medially depressed. During late stance, as the rearfoot was off the ground and the load was on the forefoot, increased foot flexibility due to increased foot pronation allowed increased magnitude of forefoot dorsiflexion before the forefoot left the ground. Increased forefoot range of motion in the sagittal plane has been associated with changes in the medial longitudinal arch, which is stabilized by the foot plantar fascia [35]. Therefore, increased forefoot range of motion overstretches foot plantar fascia and consequently contributes to the development of plantar fasciitis. In fact, individuals with plantar fasciitis has demonstrated increased forefoot range of motion and foot pronation during gait [36].

The increased ankle plantarflexion angle, or reduced ankle

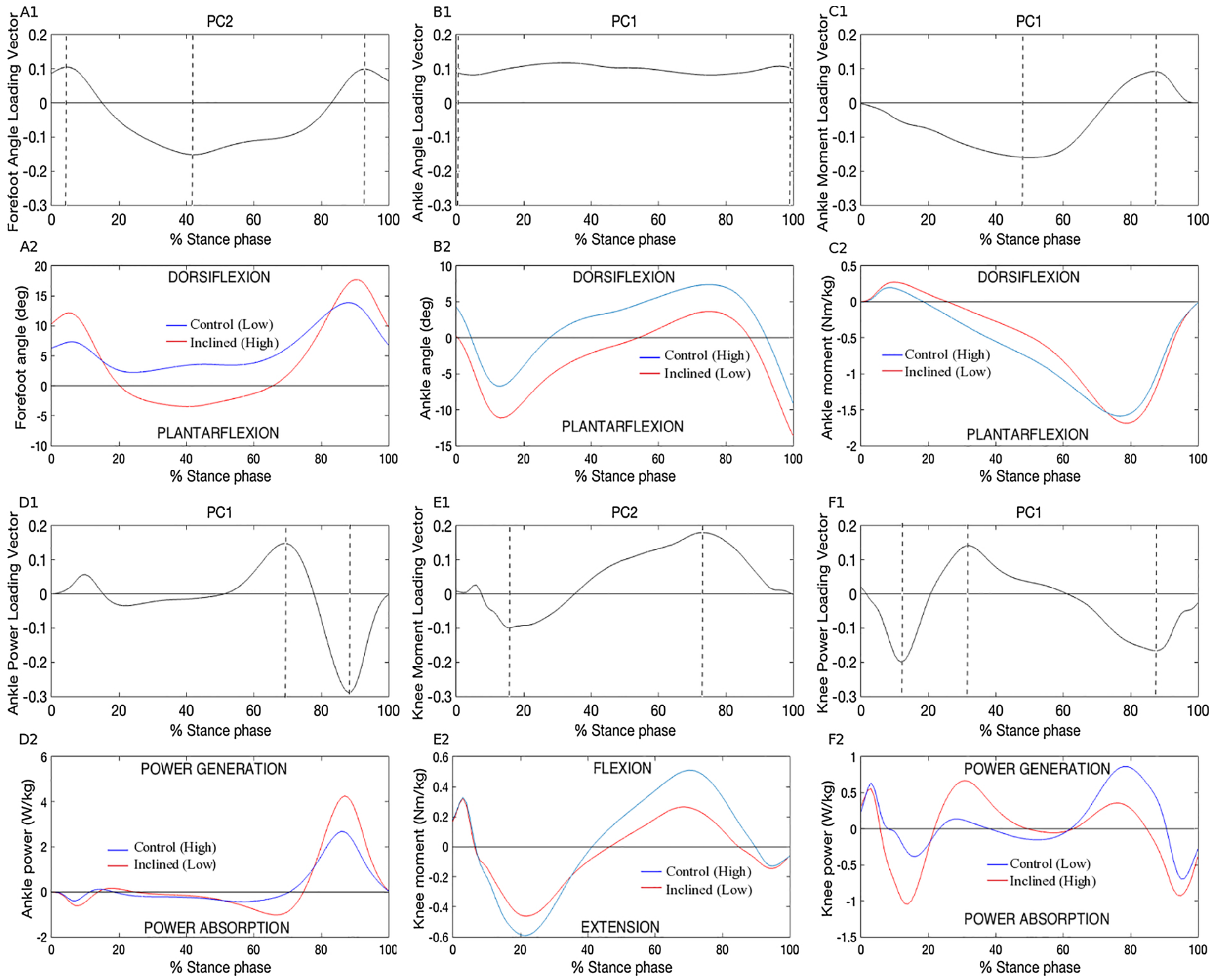

Fig. 2. Control versus inclined conditions differences demonstrated by the statistical comparisons. Shown in the figures are the waveforms that represent high and low PC scores for the indicated measure. In all cases, the waveform that represents the PC score (i.e. high or low PC score) that characterizes the control condition is shown as a blue line; the waveform that represents the PC score that characterizes the inclined condition is shown as a red line. The vertical dashed lines indicate regions when the loading vector (A1-F1) reaches a peak magnitude, which demonstrate the portions of the stance phase that most significantly contributed to the PC score and, thereby, to the differences observed between conditions. The forefoot dorsiflexion angle PC2 (A1 and A2); ankle dorsiflexion angle PC1 (B1 and B2); ankle dorsiflexion moment PC1 (C1 and C2); ankle sagittal plane power PC1 (D1 and D2); knee flexion moment PC2 (E1 and E2); and the knee sagittal plane power PC1 (F1 and F2). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

dorsiflexion angle, allowed contact of the entire foot on the ground during the inclined condition. In addition, talus plantarflexion is a component of foot pronation in closed kinetic chain, which also helps to explain the increased ankle plantarflexion angle during the inclined condition. Because the ankle was less dorsiflexed during mid and terminal stance phases, reduced ankle internal plantarflexion moments (e.g. reduced soleus and gastrocnemius eccentric activation) were required to control tibia anterior rolling, which help to explain the reduced ankle moments caused by the induced foot pronation during these phases. On the other hand, individuals demonstrated delayed and increased ankle plantarflexion moment in late stance. Increased foot pronation displaced the COP posteriorly and consequently compromised foot lever arm function during the push-off phase [34], which delayed ankle plantarflexion moment during the inclined condition. Therefore, as a compensatory mechanism to maintain gait speed, individuals increased ankle plantarflexion moment and GRF in the anterior direction during late stance, which consequently contributed to the increased range of ankle power during this phase.

The reduced knee range of moment during stance and reduced knee power generation in late stance during the inclined condition might be related to the fact that the gastrocnemius also crosses the knee joint. More specifically, foot pronation and ankle plantarflexion angle compromised gastrocnemius function and consequently reduced knee flexion moment and positive power [37]. In addition, during the loading response phase, the reduced ankle dorsiflexion angle contributed to reduce the GRF lever arm relative to the knee in the sagittal plane, reducing the knee external flexion moment and consequently reducing the knee internal extension moment. The knee extension and flexion moments are responsible for absorbing and transferring the GRF throughout the kinetic chain [38]. Therefore, the reduced knee range of moment during stance and reduced knee power generation in late stance suggest that increased foot pronation compromises part of knee functions during gait.

The findings of this study might not reflect long-term and bilateral effects of foot pronation. We evaluated the effects of unilateral foot pronation following evidence about the occurrence of asymmetrical foot pronation [39,40] and its relationship to health conditions, such as low back pain [40]. In addition, we induced increased foot pronation by specific sandals, which might not represent the effects of natural increased foot pronation due to different structure and function factors, such as foot misalignments and reduced foot and hip strength. However, we chose to artificially induce increased foot pronation in order to see the isolated effects of this motion dysfunction, which was possible by using a repeated measures design and would not have been possible

Table 1
Principal components (PCs) that demonstrated differences between control and inclined 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 pattern of the inclined condition
Forefoot dorsiflexion-plantarflexion angle	2	15.0	< 0.001	0.73	Increased range of motion during the stance phase
Ankle dorsiflexion-plantarflexion angle	1	68.0	< 0.001	0.96	Increased ankle plantarflexion angle throughout stance phase
Ankle dorsiflexion-plantarflexion moment	1	52.9	< 0.001	0.72	Reduced ankle plantarflexion moment during mid and terminal stance phases and delayed and increased ankle plantarflexion moment in push-off
Ankle sagittal plane power	1	65.7	0.006	0.56	Increased range of ankle power during late stance
Knee flexion-extension moment	2	15.3	< 0.001	0.76	Reduced range of knee moment during the stance phase
Knee sagittal plane power	1	48.2	0.005	0.56	Increased knee range of power during early stance and reduced knee power generation in late stance

Table 2
Mean and standard deviation of the peak ground reaction force (GRF) and mean center of pressure (COP) values during the control and inclined conditions with the results of statistical comparisons. Positive values represent COP displacement and GRF in the anterior directions.

Variable	Control condition	Inclined condition	*p*-value	*Effect size*
GRF in the anterior direction (N)	0.21 (0.03)	0.22 (0.03)	0.003	0.60
COP displacement in the anterior direction[a]	88.8 (7.6)	84.4 (6.9)	< 0.001	0.82

N: newtons.
[a] Reported in % of foot length.

by comparing two groups of different individuals. Finally, we did not evaluate midfoot motion, which has been recently shown to demonstrate greater mobility [41] and consequently a more important [42] and different [43] role in foot function than traditionally assumed. However, tracking the midfoot was not possible due to the sandals straps position.

5. Conclusions

Increased foot pronation increased forefoot range of motion in the sagittal plane and the ankle plantarflexion angle during the stance phase. In addition, increased foot pronation reduced ankle plantarflexion moment during mid and terminal stance phases, delayed and increased ankle plantarflexion moment during late stance, increased range of ankle power during late stance, reduced knee range of moment during stance phase, increased range of knee power during early stance and reduced knee power generation during late stance. Foot pronation also reduced the anterior displacement of the COP and increased the magnitude of the GRF in the anterior direction during the stance phase of gait.

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