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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.

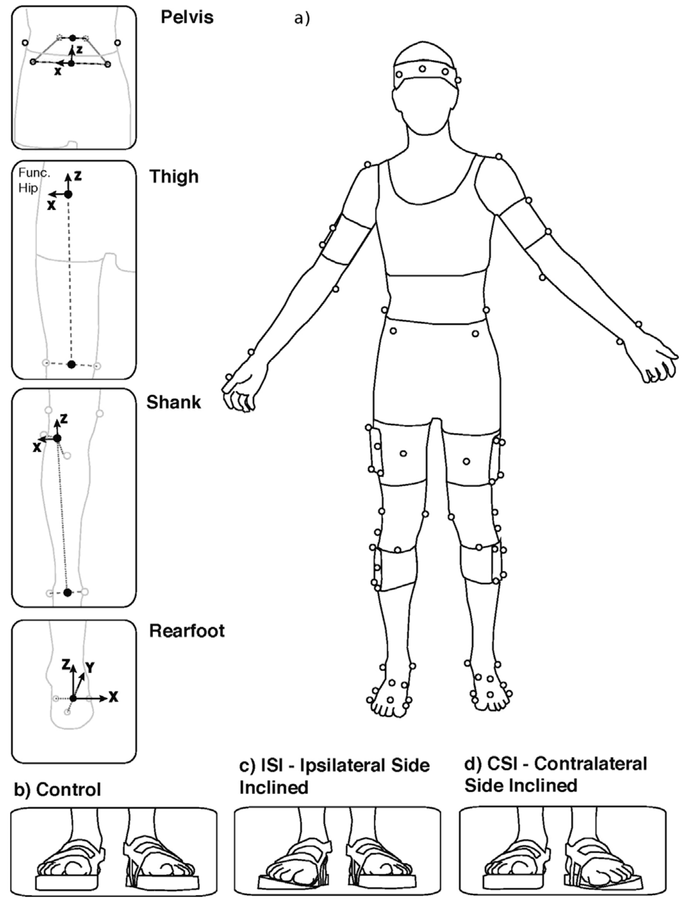

Fig. 1. Marker placement and segments coordinate systems (a); control condition (b); ipsilateral side inclined condition (c) and contralateral side inclined condition (d).

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.

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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