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Speed dependent effects of laterally wedged insoles on gait biomechanics in healthy subjects

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ABSTRACT

Laterally wedged insoles have been shown to be effective for the reduction of the knee adduction moment and other biomechanical variables that are associated with the pathogenesis of knee osteoarthritis. However, inconclusive results such as adverse effects in individual subjects or even no group-wise wedge effects have been presented in different studies and it has been suggested to identify variables that potentially confound the wedge effect. The main objective of this study was the investigation of interaction effects of lateral wedges with walking speed, as different self-selected speeds have mainly been used in previous studies.

Twenty-two healthy subjects completed gait analysis trials on an instrumented treadmill. They walked in different speed conditions (0.9, 1.1, 1.3, 1.5 m/s) with a neutral and a laterally wedged insole. Kinematics were acquired using infrared cinematography with reflective markers attached to the lower body. From the stance phase we extracted biomechanical parameters that are associated with knee joint loading and osteoarthritis severity.

No interaction effect of lateral wedges and speed was observed for most biomechanical parameters except for the ankle eversion range of motion. The main effects of wedges were reductions of the external knee adduction moment and of the knee adduction angular impulse. All biomechanical variables changed with increasing speed. Only the lateral offset of the center of pressure did not respond to wedge or to speed changes.

Our results suggest that different self-selected speeds do not confound the effect of laterally wedged insoles.

1. Introduction

Knee osteoarthritis (OA) is a progressive musculoskeletal disorder with high prevalence of 12% in adults aged 60 and older [1]. Its pathogenesis has been associated with biomechanical factors such as increased medial knee compartment load which results from the ground reaction force (GRF) passing medially to the knee joint during gait giving rise to the external knee adduction moment (EKAM) [2–5]. Laterally wedged insoles have been proposed as a non-invasive method for the treatment of early stage OA by reducing knee joint loading [4,6]. By applying lateral wedges, the frontal plane lower limb alignment is altered towards a more valgus alignment, resulting in a reduction of the knee joint moment arm, a more vertical alignment of the GRF and thus a decreased EKAM [7].

While many studies have shown evidence for biomechanical lateral wedge efficacy with EKAM reductions of up to 10% [8], single subjects showed no or adverse reactions and some studies demonstrated no or little group effects of wedges on some biomechanically relevant parameters [5,7,9–11]. Additionally, studies on the clinical benefits are still inconclusive [12]. This indicates that clinical and biomechanical responses are not universal and may depend on other variables that potentially confound the effect of insoles.

Gait speed has an effect on kinematic and kinetic variables [13] and it has been suggested that it might influence the biomechanical effect of wedged insoles when comparing subjects walking at different self-selected speeds [10].

The purpose of this study was therefore to systematically investigate the effect of gait speed on wedge efficacy for biomechanically relevant parameters such as the knee adduction moment in a population of healthy adults. We hypothesized that differences in biomechanical parameters due to speed or wedge condition would be found as previously suggested. Secondly, we investigated a potential interaction effect of speed and wedge application. The results of our biomechanical study contribute to the interpretation of speed related wedge effects.

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2. Methods

Twenty-two participants (mean ± stdev, 19 females, 3 males; age: 22.3 ± 2.2 years; mass: 62.1 ± 9.1 kg; height: 166 ± 8 cm) without any history of injuries or any other disease affecting the musculoskeletal system and no experience in wearing laterally wedged insoles for the treatment of orthopedic disorders were included in this study. Exclusion criteria were an age above 35 to limit undiagnosed degenerations to the musculoskeletal system. The local ethics committee approved the study and written informed consent was obtained from all participants before enrollment.

We collected three-dimensional kinematic and kinetic gait data using an eight camera Qualisys (Qualisys AB, Gothenburg, Sweden) motion capture system sampling at 200 Hz and an instrumented split-belt treadmill with integrated force plates (Bertec Corporation, Columbus, OH) sampling at 1000 Hz. The marker set consisted of 38 retro-reflective markers which were placed on the pelvis and the lower extremities following anatomical palpation guidelines [14]. They were placed on the anterior and posterior superior iliac spines for the definition of the pelvis and on the greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli for the definition of the hip, knee and ankle joint centers. Clusters of four markers were placed laterally on each thigh and shank for tracking. Foot markers were placed on the shoes above the first, second and fifth metatarsal heads and on the aspect of the Achilles tendon insertion of the calcaneus.

All participants wore the same shoe model (Green Silence, Brooks) with no pronation or supination support. In the first experimental condition, the foot was positioned neutrally using unwedged insoles with individualized arch support (igli insoles, medi, Bayreuth, Germany). The subtalar joint neutral position was adjusted for each subject by a professional orthopedic technician. Arch support insoles have been used in previous studies for a better fit, improved comfort, and they might better transfer mechanics to the knee [15,16]. The second experimental condition consisted of wearing laterally wedged cork insoles with a height of 4 mm over the full length of the insole that were placed below the neutral insole (Fig. 1). This amount of wedging corresponds to a wedging angle of $4-5^\circ$ which is in the range of previously used wedges [8].

After collection of a static calibration trial, the subjects started to walk on the treadmill at a velocity of 1.3 m/s for an acclimatization period of 5 min. They subsequently walked for one minute at gait speeds of 0.9 m/s, 1.1 m/s 1.3 m/s and 1.5 m/s in both insole conditions, resulting in a total of 2×4 experimental conditions. The order of speeds and wedge application was randomized to distribute any potential accommodation effect to treadmill walking over all different conditions, with changing the insole only once during the whole measurement.

Data was further processed in Visual3D (v6, C-Motion, Germantown, MD). A potential drift in the force plate data was removed by subtracting the mean of the unloaded force plate values during the airborne phases for each condition. The kinematic and kinetic data was low-pass filtered using a fourth order Butterworth filter with a cut-off frequency of 12 Hz. The gait events heel-strike and toe-off were detected for each foot separately based on a threshold of 20 N on the vertical ground reaction force component. This threshold has been shown to be high enough to avoid treadmill induced noise while maintaining a high detection sensitivity of those gait events [17]. As we told the subjects to walk in the middle of the treadmill but not to avoid the gap, we checked the data for steps spanning both belts and manually removed these steps during post-processing. Using the static standing trial and the subjects' anthropometry, a biomechanical model with seven lower body segments (feet, shank, thigh, and pelvis) was defined. Inverse kinematics and kinetics was calculated in Visual3D to determine three-dimensional joint angles and joint moments. The joint moments were normalized by each subject's mass.

The biomechanical curves during stance phase were averaged for each experimental condition, subject and leg. Parameter extraction was performed in MATLAB (R2016b, The MathWorks, Natick, MA). We extracted the following biomechanical parameters from the mean curves that have been suggested and are related to knee joint loading [7,18,19]: the peak external knee adduction moment (EKAM), the knee adduction angular impulse (KAAI), the range of motion (ROM) of knee adduction and flexion as well as the ROM of the ankle eversion during stance phase. The vertical and medio-lateral components of the ground reaction force (GRF), the knee joint moment arm (distance from the joint to the GRF vector) and the center of pressure (COP) coordinate relative to the foot coordinate system were calculated at the time of the first peak of the EKAM.

We examined each leg separately resulting in $n = 44$ cases subject to all 2 x 4 experimental conditions. To account for inflated familywise error rates and for potential correlations between the dependent variables we first conducted a repeated measures MANOVA in R 3.3.3 [20]. Follow-up two-way repeated measures ANOVAs were performed for each biomechanical parameter to identify significant changes due to wedge or speed or an interaction of both factors with a Bonferroni corrected a-priori level of $\alpha = 0.05/9 \approx 0.006$. If the assumption of sphericity had been violated for the main effects, the degrees of freedom were corrected using Greenhouse-Geisser estimates of sphericity.

3. Results

An overview over the biomechanical parameters subject to different speeds and insoles is shown in Fig. 2. Table 1 depicts the corresponding numeric values. Using Pillai's trace, significant effects of speed and wedge on the biomechanical variables were determined in the MANOVA (speed: $p < 0.001$, wedge: $p = 0.03$, interaction: $p = 0.05$). Follow-up ANOVAs revealed significant changes of the peak EKAM and the KAAI for speed and wedge condition separately but no significant interaction effect (Table 2). Averaged over all speeds, the EKAM decreased by 2.5 % and the KAAI by 3.9 % in the wedge

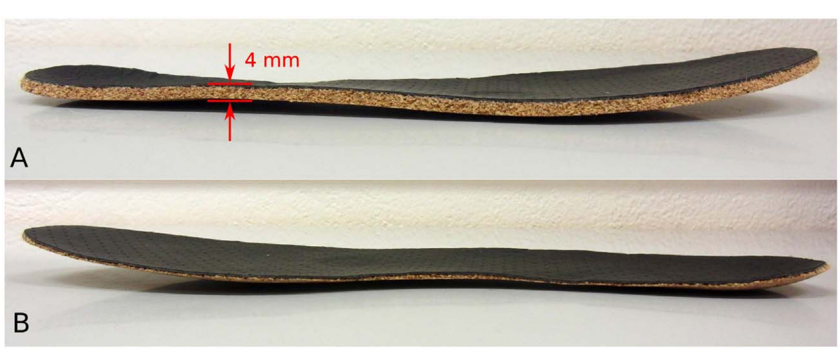

Fig. 1. Laterally wedged insoles. Lateral view (A) and medial view (B).