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Heel height affects lower extremity frontal plane joint moments during walking

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ABSTRACT

Wearing high heels alters walking kinematics and kinetics and can create potentially adverse effects on the body. Our purpose was to determine how heel height affects frontal plane joint moments at the hip, knee, and ankle, with a specific focus on the knee moment due to its importance in joint loading and knee osteoarthritis. 15 women completed overground walking using three different heel heights (1, 5, and 9 cm) for fixed speed (1.3 m s^{-1}) and preferred speed conditions while kinematic and force platform data were collected concurrently. For both fixed and preferred speeds, peak internal knee abduction moment increased systematically as heel height increased (fixed: 0.46, 0.48, 0.55 N m kg^{-1}; preferred: 0.47, 0.49, 0.53 N m kg^{-1}). Heel height effects on net frontal plane moments of the hip and ankle were similar to those for the knee; peak joint moments increased as heel height increased. The higher peak internal knee abduction moment with increasing heel height suggests greater medial loading at the knee. Kinetic changes at the ankle with increasing heel height may also contribute to larger medial loads at the knee. Overall, wearing high heels, particularly those with higher heel heights, may put individuals at greater risk for joint degeneration and developing medial compartment knee osteoarthritis.

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1. Introduction

Footwear design, including cushioning, stiffness, heel width and height, can alter walking mechanics. As one example, women's high-heeled shoes dramatically alter walking kinematics and kinetics. Despite a reported lack of comfort and support [1,2], the American Podiatric Medical Association reported 72% of women wear high heels; 40% of these women wear them daily [3].

Higher heel heights contribute to slower self-selected walking speeds, shorter stride lengths [4–6] and greater knee flexion, plantar flexion, anterior pelvic tilt, and trunk extension [6–9]. Higher peak vertical and anterior–posterior ground reaction forces, medial forefoot pressures, peak internal knee extensor moments, peak internal knee abduction moments (which are consistent with higher peak external adduction moments, as reported by some investigators), and lower internal plantar flexion moments have also been observed during walking in high heels [1,5,9–12]. These kinematic and kinetic changes are thought to create adverse repetitive, dynamic loading in lower extremity joints, which may contribute to joint degeneration and the development of knee osteoarthritis (OA) [11–13].

Because knee OA occurs most often in the medial tibiofemoral compartment of the knee [14,15], recent research has focused on frontal plane kinetics and possible loading asymmetries within the knee. Schipplein and Andriacchi [16] reported medial forces in the knee may be 2.5 times greater than those on the lateral side. Higher internal knee abduction moments or higher external knee adduction moments are thought to be associated with larger medial compartment loads [16,17] and the development of knee OA [16–18].

To our knowledge, the effect of heel height on internal knee abduction moments has received insufficient attention and has not been investigated completely. Kerrigan et al. [11,12] employed only two heel heights and allowed subjects to walk at self-selected speeds for each heel height condition. While preferred speed enhances real-world application, a fixed walking velocity would control for speed effects on walking kinematics and kinetics. Further, investigated heel heights have not exceeded 6 cm, which does not encompass the full range of heel heights typically worn by women. Finally, limited attempts have been made to control footwear design [11,13], which could confound the effects of heel height.

Our purpose was to determine the effect of heel height on net joint moments at the hip, knee, and ankle in the frontal plane, with particular attention placed on the knee moment. It was hypothe-

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sized that frontal plane net joint moments of the lower extremity increase systematically as heel height increases.

2. Methods

2.1. Subjects

15 women (age: 23.8 ± 4.4 years, height: 165.5 ± 7.1 cm, mass: 60.9 ± 8.7 kg), free from injury for at least 12 months prior to participation, served as subjects. All provided written informed consent prior to participation.

2.2. Data collection

Subjects completed two sessions. Session 1 was used primarily for orientation to testing and determination of preferred walking speeds for each heel condition. Session 2 served as the primary data collection session in which walking trials for each heel height condition at both preferred and fixed ($1.3\ \mathrm{m\,s^{-1}}$) speeds were completed. Because preferred speed tends to decrease as heel height increases [4–6], measuring effects of heel height at preferred speeds enhanced our ecological validity. Because walking speed is known to affect gait kinematics and kinetics, a fixed speed condition was also included. Preferred walking speeds while wearing high heels have ranged from $1.23\ \mathrm{m\,s^{-1}}$ [8] to $1.44\ \mathrm{m\,s^{-1}}$ [19]. Further, $1.3\ \mathrm{m\,s^{-1}}$ has been used in previous research [1,2].

2.2.1. Session 1

Subjects completed a short questionnaire assessing frequency, duration, and consistency of high heel use. Body weight, height, and anthropometric measurements of the right leg and foot were obtained.

Shoes with heel heights of 0.8, 5.1, and 8.9 cm (hereafter referred to as 1, 5, and 9 cm) were commercially available and selected based upon their similar construction so that the main difference amongst shoes was heel height (Fig. 1). Shoe construction included a narrow heel, narrow forefoot and toe box, and limited cushioning material.

Preferred walking speed was measured for each heel height in a randomly assigned order for each subject. One reflective marker was placed on the low back at the L5/S1 level. After at least a 5-min walking accommodation period in each shoe, subjects walked overground through the measurement zone of a Vicon motion system (Centennial, CO, USA) at a comfortable pace as the position of the L5/S1 marker was tracked at 200 Hz. The average horizontal velocity of the marker for 10 trials served as the measure of preferred walking speed. To conclude the first session, subjects practiced walking at $1.3\ \mathrm{m\,s^{-1}}$ with each shoe.

2.2.2. Session 2

15 reflective markers were placed over anatomical landmarks of the subject's lower trunk, pelvis, and right lower extremity. Subjects completed heel height conditions in the same order as in session 1. 10 acceptable trials for each heel height were completed at each subject's preferred speeds followed by 10 acceptable trials for each heel height at $1.3\ \mathrm{m\,s^{-1}}$. Marker position and ground reaction force (GRF) data were collected synchronously at 200 Hz and 1000 Hz, respectively, using the Vicon system and an AMTI force platform (Watertown, MA, USA). Acceptable trials were those in which average horizontal velocity of the L5/S1 marker during the stance phase on the force platform fell within 3% of the target velocity, the right foot was completely on the force platform, and there was no visible indication the participant altered step characteristics to impact the force platform.

Fig. 1. Shoes of heel height 1, 5, and 9 cm.

2.3. Data analysis

Raw marker position and force platform data were processed using Matlab (The Mathworks, Natick, MA, USA). A 4th order, zero-lag, low pass (6 Hz) Butterworth filter was used to smooth raw marker position [20] and force platform data [21]. The right leg stance phase was analyzed for all kinematic and kinetic variables with the exception of stride length and rate. A vertical ground reaction force threshold of 20 N was used for the identification of heel strike and toe off events. Stride length was calculated as the difference in right foot heel marker position for consecutive heel strikes throughout the data collection period. Stride rate was computed as the inverse of stride time, which was determined from the time between consecutive heel strikes of the right foot. Marker position data were used to calculate three-dimensional segment angles for the pelvis, thigh, shank, and foot. Segment and joint angles were calculated using marker clusters to create rotation matrices for the pelvis, thigh, shank, and foot using Cardan angles with a flexion/extension, abduction/adduction, internal/external rotation sequence.

Anthropometric measurements were used to estimate segment masses, center of mass locations, and moments of inertia needed for joint moment calculations [22]. Segment angular velocities and accelerations and center of mass linear velocities and accelerations were calculated using finite difference equations. Net internal three-dimensional joint moments were calculated using inverse dynamics with rigid body assumptions [22]. Moments were normalized to body mass ($\mathrm{N\,m\,kg^{-1}}$). Peak hip and knee moments during early and late stance and peak ankle moment during late stance were extracted for statistical analysis.

2.4. Statistical analysis

One-way analysis of variance (ANOVA) with repeated measures ($\alpha = 0.05$) was used to assess the effect of heel height on lower extremity frontal plane net joint moments and walking kinematics for both preferred and fixed speed conditions. The means of 10 trials for each subject for each heel height were used in statistical analyses. In the event of a significant heel height effect, a Bonferroni adjustment ($\alpha = 0.05/3 = 0.017$) was used for post hoc comparisons.

3. Results

Subjects reported wearing high heels 2.1 days per week (SD = 1.6, range: once every six months to five days per week). Time spent wearing high heels averaged 5.1 (SD = 1.6) hours per episode. Self-reported typical heel height worn by subjects averaged 7.6 cm (SD = 1.5, range: 5.1–10.2 cm). Mean experience with high heels was 7.6 years (SD = 4.4, range: 7 months to 20 years).

Subjects closely matched the $1.3\ \mathrm{m\,s^{-1}}$ fixed speed (Table 1). For preferred speed trials, subjects walked significantly slower for the 9 cm heel compared to the 1 and 5 cm conditions ($F_{2,13} = 7.34$, $p = 0.003$) (Table 1). Preferred speeds were similar to those previously reported ($1.23\ \mathrm{m\,s^{-1}}$ [8] to $1.44\ \mathrm{m\,s^{-1}}$ [19]). For fixed speed trials, stride length (SL) systematically decreased ($F_{2,13} = 13.71$, $p < 0.001$) and stride rate (SR) increased ($F_{2,13} = 7.18$, $p = 0.003$) as heel height increased. For preferred speed trials, SL was significantly shorter for the 9 cm condition compared to the 1 and 5 cm heights ($F_{2,13} = 13.61$, $p < 0.001$), whereas SR was not affected by heel height ($F_{2,13} = 0.09$, $p = 0.91$).

Net frontal plane hip moments were abduction moments for nearly all of the stance phase (Fig. 2, panels A and B). The hip moment profiles and amplitudes were consistent with those reported by Esenyel et al. [5]. Although similar heel height effects were observed for both fixed and preferred speeds, only the peak hip abduction moment during early stance for the fixed speed condition reached statistical significance ($F_{2,13} = 17.02$, $p < 0.001$). The peak hip abduction moment for the 9 cm heel was approximately 10% higher than those for 1 and 5 cm heels (Table 1). Heel height did not significantly affect the second peak hip abduction moment that occurs in late stance (fixed speed: $F_{2,13} = 2.34$, $p = 0.12$; preferred speed: $F_{2,13} = 0.19$, $p = 0.83$).