Gait & Posture 39 (2014) 1045–1050

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Effects of long-term wearing of high-heeled shoes on the control of the body's center of mass motion in relation to the center of pressure during walking

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ARTICLE INFO

Article history:
Received 28 June 2013
Received in revised form 2 December 2013
Accepted 12 January 2014

Keywords:
Gait
High heels
Balance
Habitual
Inclination angle

ABSTRACT

High-heeled shoes are associated with instability and falling, leading to injuries such as fracture and ankle sprain. This study investigated the effects of habitual wearing of high-heeled shoes on the body's center of mass (COM) motion relative to the center of pressure (COP) during gait. Fifteen female experienced wearers and 15 matched controls walked with high-heeled shoes (7.3 cm) while kinematic and ground reaction force data were measured and used to calculate temporal-distance parameters, joint moments, COM–COP inclination angles (IA) and the rate of IA changes (RCIA). Compared with inexperienced wearers, experienced subjects showed significantly reduced frontal IA with increased ankle pronator moments during single-limb support ($p < 0.05$). During double-limb support (DLS), they showed significantly increased magnitudes of the frontal RCIA at toe-off and contralateral heel-strike, and reduced DLS time ($p < 0.05$) but unaltered mean RCIA over DLS. In the sagittal plane experienced wearers showed significantly increased mean RCIA ($p < 0.05$) and significant differences in the RCIA at toe-off and contralateral heel-strike ($p < 0.05$). Significantly increased hip flexor moments and knee extensor moments at toe-off ($p < 0.05$) were needed for forward motion of the trailing limb. The current results identified the change in the balance control in females after long-term use of high-heeled shoes, providing a basis for future design of strategies to minimize the risk of falling during high-heeled gait.

1. Introduction

In modern society, many women wear high-heeled shoes in both professional and social settings [1], with up to 39% of American women [2,3] and 78% of British women wearing them on a daily basis [4,5]. High-heeled shoes often result in a reduced supporting base compared with barefoot, increasing the difficulty of maintaining balance, and the risk of falling [6,7]. Compared to barefoot gait, experienced wearers of high-heeled shoes adopted a conservative strategy for controlling the motion of the body's center of mass (COM) in relation to the center of pressure (COP) during high-heeled gait [8]. Whether long-term wearing of high-heeled shoes would help improve the balance control during high-heeled gait remained unclear.

Habitual wearers of high-heeled shoes were reported to experience chronic adaptations in muscle-tendon architecture, including shortening of the gastrocnemius medialis fascicles [9,10] and increased Achilles tendon stiffness [9]. These adaptations could shift the stretch distribution of a muscle-tendon unit away from the tendinous tissues toward the muscle fascicles during gait, altering neural activation patterns and reducing muscle-tendon unit efficiency [11]. Reduced endurance of the gastrocnemius lateralis and peroneus longus muscles was also reported [12].

Habitual and inexperienced wearers were found to accommodate differently to walking in high-heeled shoes when compared to low-heeled shoes, the former subjects showing decreased rotations of the upper trunk and exaggerated motions of the pelvis, and the latter showing the opposite [13]. Habitual wearers also had a shorter stride length than inexperienced wearers. These kinematic changes suggest that both groups adopt a cautious gait style, particularly the inexperienced group [13].

Differences in the changes of gait kinematics and muscle-tendon architecture between habitual and inexperienced wearers suggest that the two groups may have different balance control

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http://dx.doi.org/10.1016/j.gaitpost.2014.01.007

during high-heeled gait. Menant et al. [7] studied the effects of wide base, elevated shoe heels on gait stability using the minimum COM–BOS margin during SLS, but the effects of long term use were not considered. Since the dynamic stability of gait depends on both the position and the velocity of the COM with respect to the BOS [14], the minimum COM–BOS margin alone may not be sufficient to describe the body's stability. The continuously varying BOS during double-limb support (DLS) presents another difficulty in the definition of the COM–BOS margin. Chien et al. [8] used COM–COP inclination angles (IA) to describe the body's dynamic control in experienced wearers. The experienced wearers were found to adopt a conservative balance control strategy when compared with barefoot, with reduced normalized walking speed, reduced frontal IA throughout the high-heeled gait cycle, and also a reduced frontal and sagittal rate of change of IA (RCIA) during DLS. During high-heeled gait the BOS is reduced both in the anterior–posterior and in the medial–lateral directions compared with barefoot. During single-limb support (SLS), the BOS is reduced mainly as a result of the reduced area of the heel. During DLS, the reduced BOS is a result not only of the reduced area of the heel, but also of the reduction in the step width and stride length [8]. The reduced BOS was found to be a major factor affecting the observed conservative control strategy in habitual wearers compared to the barefoot condition [8]. However, whether this strategy was adopted immediately or whether it was a result of long-term adaptation to high-heeled shoes has not been explored.

The purposes of this study were to investigate the long-term effects of wearing high-heeled shoes on the body's COM motion in terms of IA and RCIA during high-heeled gait. It was hypothesized that during high-heeled walking, experienced wearers would decrease the frontal IA throughout the gait cycle, and the frontal and sagittal RCIA during DLS when compared with the inexperienced controls.

2. Materials and methods

2.1. Subjects

Fifteen female habitual wearers of high-heeled shoes (experienced group; age: 24.4 ± 3.4 years; height: 158.9 ± 5.7 cm; mass: 49.2 ± 5.1 kg) and 15 matched controls (inexperienced group; age: 24.9 ± 4.1 years; height: 161.0 ± 4.2 cm; mass: 50.3 ± 3.7 kg) participated in the current study with informed written consent, as approved by the Institutional Research Board. None of the subjects suffered from any neuromusculoskeletal pathology that might have affected their normal gait. Experienced wearers had worn shoes with narrow heels of more than 3 cm height a minimum of three times per week, six hours per day for at least two years. The inexperienced wearers wore high-heeled shoes less than twice per month.

2.2. Data collection

Each subject walked in high-heeled shoes (height: 7.3 cm) at a self-selected pace on an 8-m walkway. The shoes were narrow-heeled (heel base: $2.0\,\text{cm} \times 1.6\,\text{cm}$; mass: 0.2 kg) and were commercially available [8]. Subjects were fitted with the most suitable test shoes from several different sizes, and were allowed to familiarize themselves with the walkway before data collection.

Each subject wore 39 retroreflective markers for tracking motions of the body segments [15]. Markers for the heels, big toes and fifth metatarsal bases were attached to the corresponding positions on the shoes. Markers on the navicular tuberosity and malleoli were not affected by the shoes. Three-dimensional trajectories of the markers were measured using a motion capture system (Vicon 512, OMG, UK) at a sampling rate of 120 Hz, and were low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 5 Hz [16]. The ground reaction forces (GRF) were collected from two forceplates (AMTI, USA) at a frequency of 1080 Hz [15]. Six successful trials, three for each limb, were obtained.

2.3. Data analysis

The body was modeled as a system of 13 rigid segments, namely the head and neck, trunk, pelvis, arms, forearms, thighs, shanks and feet [15,17]. Coordinates of the markers gathered during a static calibration trial were used to define the anatomical coordinate system of each of the body and foot/shoe segments, with the positive x-axis directed anteriorly, the positive y-axis superiorly and the positive z-axis to the right. A Cardanic rotation sequence (z–x–y) was used to describe the rotational movements of each joint [15]. In order to minimize the errors owing to skin movement artifacts, a global optimization method was used [18]. With the measured GRF and kinematic data, inverse dynamics were used to calculate the inter-segmental forces and moments at the lower limb joints. Inertial properties for each body segment were obtained using an optimization method [19], using Dempster's coefficients as the initial guess [17]. In the shod condition, the mass of the foot/shoe unit was determined as the sum of the masses of the foot and the shoe, with the center of mass taken as that of the foot as an approximation. All the calculated joint moments were normalized to body weight (BW) and leg length (LL) that was defined as the sum of the shoe height and the distance between the anterior superior iliac spine and the medial malleolus [20].

The body's COM position was calculated as the weighted sum of the positions of all 13 model body segments. The COP position was calculated using forces and moments measured by the forceplates [21]. The medial/lateral positions of the COM and COP were described relative to the line of progression that bisected the medial/lateral range of motion of the COM during a gait cycle, a positive value being to the side of the contralateral limb [22]. The anterior/posterior positions of the COM and COP were described parallel to the direction of progression, a zero value being the position of heel-strike and a positive value being anterior to that position. The IA in the sagittal plane (α) and frontal plane (β) were then calculated as follows [16] (Fig. 1).

$$\vec{t} = \left(\frac{\vec{P}_{COM\_COP} \times \vec{Z}}{|\vec{P}_{COM\_COP}|} \right) \tag{1}$$

$$\alpha = \sin^{-1}(t_x) \tag{2}$$

$$\beta = \sin^{-1}(t_y) \tag{3}$$

where $\vec{P}_{COM\_COP}$ was the vector pointing from the COP to the COM, and $\vec{Z}$ was the unit vector of the global vertical axis. With the current forceplate setup, α and β were calculated from the beginning of SLS to the subsequent heel-strike. The RCIA for α and β were also calculated by smoothing and differentiating their trajectories using the GCVSPL method [23].

During the gait cycle, the transitions between SLS and DLS, i.e., heel-strike of the contralateral leg and toe-off of the ipsilateral leg, are critical instances at which maintaining body stability is expected to be more difficult [16]. Therefore, the values of the IA, RCIA and the joint moments at these instances were obtained. The values of IA, RCIA and the joint moments were averaged over the DLS and SLS, respectively. The ranges of motion (ROM) of IA during DLS and SLS, as well as the peak RCIA during DLS, were obtained. Temporal-distance parameters (gait speed, stride length, step width, stride time, cadence, stance time, DLS time and SLS time) for all conditions were also calculated. For between-group comparisons, gait speed,