Effects of heel base size, walking speed, and slope angle on center of pressure trajectory and plantar pressure when wearing high-heeled shoes

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Abstract

High-heeled shoes are associated with instability and a high risk of fall, fracture, and ankle sprain. This study investigated the effects of heel base size (HBS) on walking stability under different walking speeds and slope angles. The trajectory of the center of pressure (COP), maximal peak pressure, pressure time integral, contact area, and perceived stability were analyzed. The results revealed that a small HBS increased the COP deviations, shifting the COP more medially at the beginning of the gait cycle. The slope angle mainly affected the COP in the anteroposterior direction. An increased slope angle shifted the COP posterior and caused greater pressure and a larger contact area in the midfoot and rearfoot regions, which can provide more support. Subjective measures on perceived stability were consistent with objective measures. The results suggested that high-heeled shoes with a small HBS did not provide stable plantar support, particularly on a small slope angle. The changes in the COP and pressure pattern caused by a small HBS might increase joint torque and muscle activity and induce lower limb problems.

Keywords: high-heeled shoes, heel base size, walking speed, slope angle, center of pressure trajectory

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1. Introduction

High-heeled shoes are widely worn in the society. However, long-term wearers are vulnerable to foot problems such as hallux valgus, forefoot pain, and calluses (Al-Abdulwahab & Al-Dosry, 2000; Goud, Khurana, Chiodo, & Weissman, 2011; Menz & Morris, 2005). These problems are thought to be associated with the redistribution of plantar pressure (Lee & Hong, 2004; Mandato & Nester, 1999; Snow, Williams, & Holmes, 1992) and changes in gait patterns (Barkema, Derrick, & Martin, 2012; Esenyel, Walsh, & Walden, 2003; Ho, Blanchette, & Powers, 2012; Opila-Correia, 1990) caused by the heel elevation. Unsurprisingly, fashion models easily lose balance on the runway when wearing stiletto-heeled shoes. Some studies have reported that walking stability decreased when wearing high-heeled shoes, increasing the risk of sprain or fall (Ebbeling, Hamill, & Crussemeyer, 1994; Lee, Shieh, Matteliano, & Smiehorowski, 1990; Menz & Lord, 1999). Ankle sprain or fall might cause serious damage, such as permanent damage to ligaments and bone fractures (Lee et al., 1990; Menz & Lord, 1999). Thus, achieving a more comprehensive understanding of how the design factors of high-heeled shoes affect walking patterns and influence foot balance is crucial.

In addition to the heel height, the heel base size (HBS) might be another major factor influencing the walking stability of those wearing high-heeled shoes (Menz & Lord, 1999; Chien, Lu, & Liu, 2013). A broader base of support has been thought to enhance stability (Edelstein, 1987; Finlay, 1986). Therefore, similar suggestions have been provided for high-heeled shoe design in response to the recognition that narrow heels (common in most high-heeled shoes) may cause instability. However, direct evaluation on the importance of HBS on the stability of high-heeled shoes is little (Menz & Lord, 1999; Chien et al., 2013). To date, only the redistribution of the plantar pressure caused by changes in the HBS has been reported (Guo et al., 2012). More insights into the effects of the HBS on stability, particularly when walking speed and slope angle are changed, should be provided. Walking speed and slope angle factors have been thought to exert marked effects on postural stability and gait (England & Granata, 2007; Flavel, Nordstrom, & Miles, 2003; Sun, Walters, Svensson, & Lloyd, 1996).

Walking stability during the stance phase can be measured according to the abnormal trajectory of center of pressure (COP), which is caused by the postural adjustments such as the straightening of the spinal column, verticalization of the sacrum and flexion of the hip and knee when the body attempts to establish stability (Gerber et al., 2012; Leroux, Fung, & Barbeau, 2002). The COP is also regarded as a measurement of the tilting movements of the foot (Hoogvliet, van Duyl, de Bakker, Mulder, & Stam, 1997) and can be interpreted as a moment arm for the vertical plantar force (Kim, Uchiyama, Kitoaka, & An, 2003). Deviation in the COP results in an extra inverting or everting moment on the foot (Gefen et al., 2002). This increased foot tilting movement causes the COP to diverge from the center of mass (COM). The body might have difficulty to bring the COM back to be right above the COP (Chien et al., 2013) and thus cause the instability. The COP parameters, such as the medial-lateral and anterior-posterior COP positions, the average distance of COP from the mean position, and the COM-COP inclination angles, were therefore sensitive in order to assess the changes in walking stability (Chien et al., 2013; Gefen et al., 2002; Han, Paik, & Im, 1999; Lemaire, Biswas, & Kofman, 2006; Murray, Seireg, & Sepic, 1975). Some of these parameters have been used to represent the stability when wearing high-heeled shoes (Chien et al., 2013; Cho & Choi, 2005; Gefen et al., 2002; Gerber et al., 2012; Zhang & Li, 2014). Although wearing high-heeled shoes generated a larger displacement of the COP in static trials and reduced the COM-COP inclination angle during walking compared with a barefoot condition (Chien et al., 2013; Cho and Choi, 2005), the reduced heel base was suggested to be more important factor for the reduced stability (Chien et al., 2013). However, the effects of the heel base size on the walking stability in terms of the COP have not been well documented particularly when the walking speed and the slope angle change, which are common in daily activities.

This study evaluated the effects of the HBS of high-heeled shoes on the COP trajectory, plantar pressure and perceived stability. Different walking speeds and slope angles were assessed to determine possible interactions among the HBS, walking speed, and slope angle. The results can be a reference for designing ergonomic high-heeled shoes and walking healthily.

2. Methods

2.1. Participants

The study participants comprised 15 female adult volunteers with a foot size of 38 (European standard). The participants were aged 20–38 years, with a mean age of 22.5 years (standard deviation [SD] = 4.7 years), and had a mean body weight and height of 51.3 kg (SD = 4.9 kg) and 1.61 m (SD = 0.04 m), respectively. Most of the participants wore high-heeled shoes one to three times per week. All participants reported that they did not have any foot related disorders, skin lesions, or health-related problems. Moreover, they all were right-foot dominant, and their feet comfortably fit into the experimental shoes. Signed written consent was obtained from all participants before the experiments.

2.2. Experimental equipment and materials

The F-Scan In-Shoe System (Tekscan Inc., USA) was used to measure the plantar pressure at different anatomical regions of the foot and the foot COP. The sampling frequency was set at 100 Hz. Two pairs of high-heeled shoes were customized according to the exact same design but had different HBSs (Fig. 1), namely 0.88 cm^2 (small) and 8.92 cm^2 (large). The heel height was 3 inches (7.62 cm) and the shoe size was 38 (European standard). The walking speed and slope angle were controlled using a treadmill. Two walking speeds (slow: 112 cm/s; fast: 143 cm/s) and two slope angles (small: 0°; large: 10° upward inclination) were used (Opila-Correia, 1990; Snow & Williams, 1994; Sun et al., 1996). To assess perceived stability, the visual analogue scale (VAS) was applied in this study (Müncrmann, Nigg, Stcfanyshyn, & Humble, 2002). Participants responded to the question, “How comfortable do you feel about stability?” for each HBS setting during walking. The participants indicated their perception levels on a 100-mm long scale, a continuous line with indicators ranging from 0 (*Not comfortable at all*) to 100 (*The most comfortable condition imaginable*). The VAS results enabled parametric statistical analyses, the probability of which is statistically and clinically crucial.

(a) (b)

Fig. 1. The experimental shoes with two different HBSs (from left to right: a small size and a large size) (a) Side view of experimental shoes (b) Heel base sizes

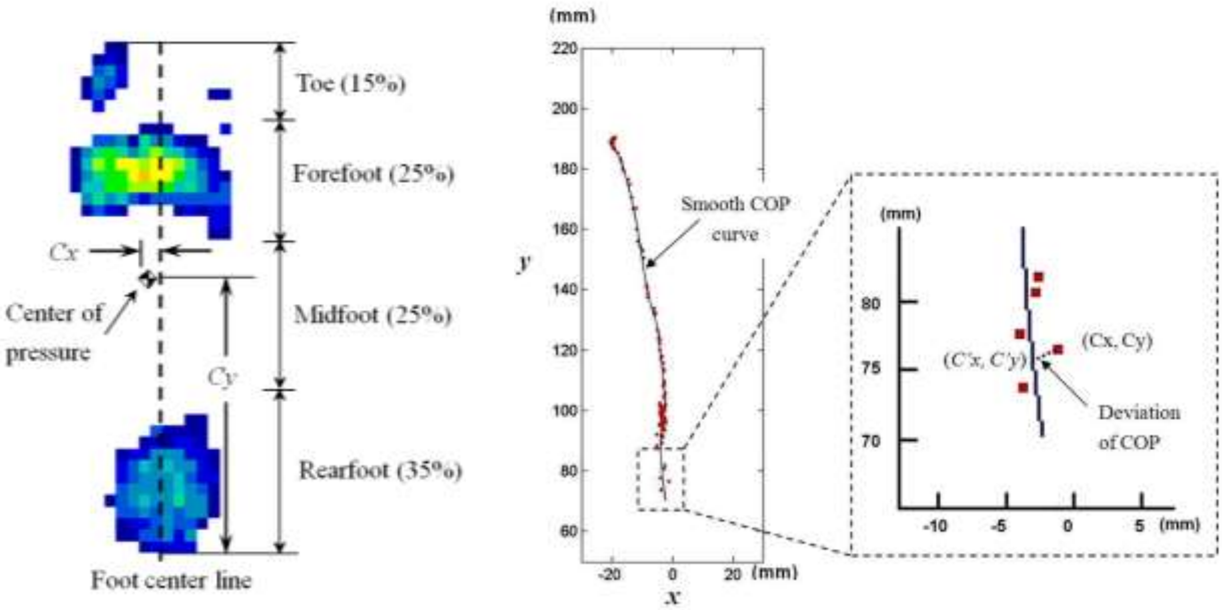
2.3. Experimental procedure

The experiment was conducted in a locomotion laboratory, and eight conditions (2 HBSs × 2 speeds × 2 slopes) were examined. The sequence of these conditions was randomized to minimize the possible confounding effects. After the F-Scan insole sensor was calibrated according to each participant’s body weight, the participants first walked on the treadmill for 3 minutes to become habituated to each HBS, walking speed, and slope angle. During data collection, the plantar pressure and COP of the right foot were recorded using the F-Scan System for 10 seconds; two trials for each condition were conducted. Questionnaires on the perceived stability level of each pair of shoes were distributed to the participants after each HBS setting. To determine uniform perceptions of stability, participants were required to not consider the effect of shoe cosmetics and style in the rating. There was a 2-minute rest between every condition in order to prevent fatigue.

2.4. Data analysis

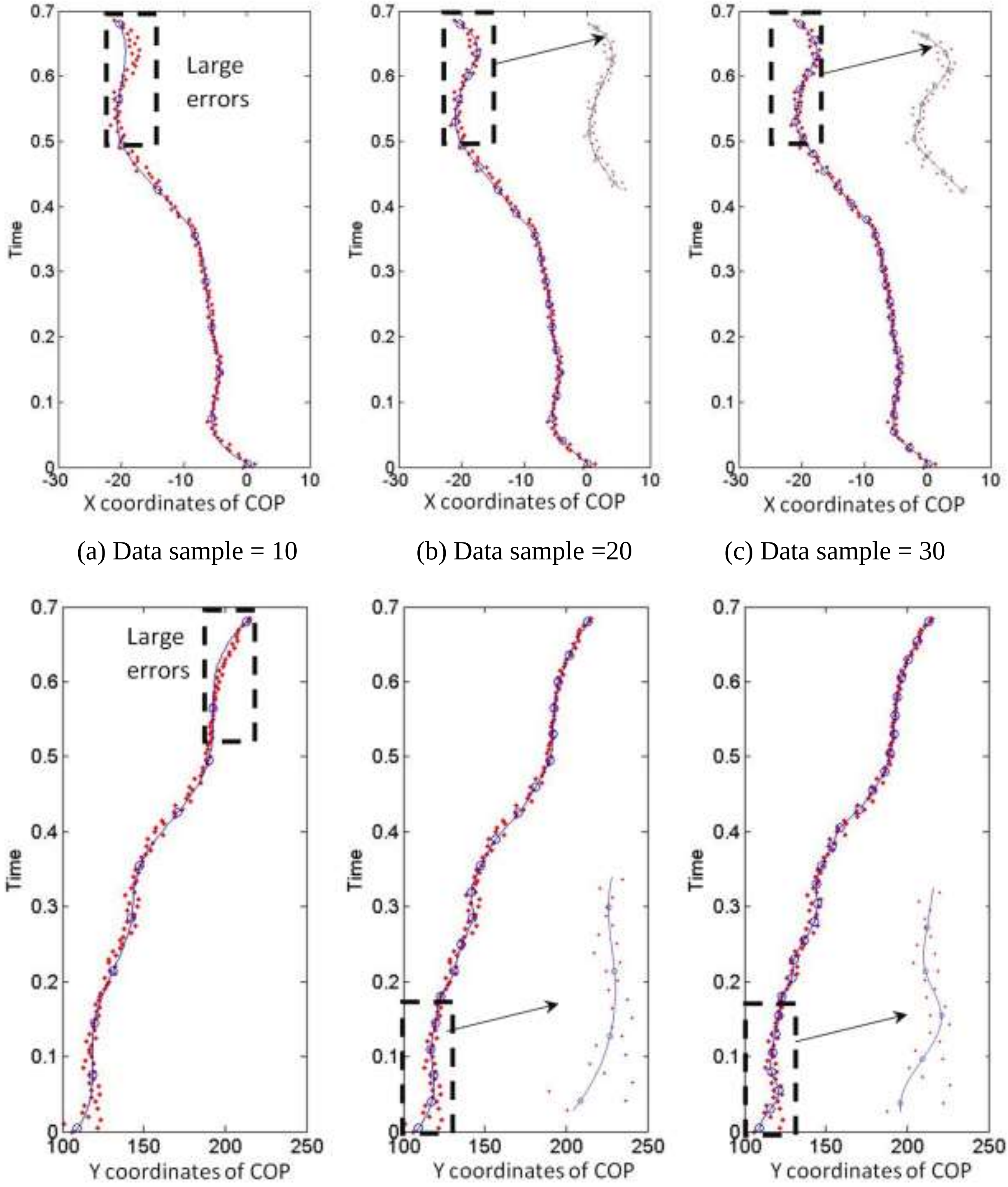
A full factorial within-subject experimental design was applied in this study. All participants walked on the treadmill at two walking speeds and two slope angles wearing two HBSs. The maximal peak pressure, pressure time integral, and pressure contact area during the stance phase were calculated in the four foot regions of interest: toe, forefoot, midfoot, and rearfoot (Fig. 2[a]). The coordinates of and deviations in the COP during the stance phase were also measured and calculated. The stance phase was separated into four phases (Han et al., 1999; Perry, 1992): loading response (LR), midstance (MS), terminal stance (TS), and preswing (PS). The LR spanned from initial contact until contact with the first or second metatarsal head. The MS followed the LR, continuing until the heel rose, which marked the beginning of the TS. The TS ended when the second pressure peak occurred and was followed by the PS, which continued until the stance phase ceased.

The location of the COP (Fig. 2 [a]) was described by the coordinates (C_x, C_y), where C_x is the distance between the COP and foot center line in the mediolateral direction, and C_y is the distance between the COP and the end of the heel in the anteroposterior direction. A negative value of C_x indicates a medial shift, and a positive value of C_y indicates an anterior shift. The average C_x ($\overline{C_x}$) and the average C_y ($\overline{C_y}$) were calculated for each subphase during the entire stance phase.



(a) (b)

Fig. 2. An example of the foot COP (a) in a time frame (b) the trajectory of COP with a simulated curve



(a) Data sample = 10 (b) Data sample =20 (c) Data sample = 30

(d) Data sample = 10 (e) Data sample =20 (f) Data sample = 30

Fig. 3. An example of simulated COP splined curve calculation with different number of data samples

In order to compare the COP deviations for different participants and between different conditions, an approximated COP curve was simulated using a spline curve according to the COP coordinate points (C_x, C_y) for each stance phase. Separate spline curves were generated for

C_x and C_y with respect to time, since there are a one-one mapping between C_x and C_y with respect to time. To generate the spline curve sampled data points ranging from 5 to 30 with a step of 5 sampled points were investigated. If less points were selected (sampled data point <= 15), the errors between actual COP points and the approximated curve were large (Fig. 3[a] and 3[d]). If more points were selected (sampled data point >= 25), the approximated COP curve was not very smooth and had undesirable ripples (Fig. 3[c] and 3[f]). When the number of sampled points was equal to 20, the spline curve seemed to better approximate the COP points (Fig. 3[b] and 3[e]), hence the smoothed mean COP curve was simulated using 20 sample points to accurately approximate the COP path. The deviation of the actual COP from the simulated COP curve was calculated using the following equation (Fig. 2[b]):

$$D = \sqrt{(C_x - C_x^{'})^2 + (C_y - C_y^{'})^2} \tag{1}$$

where $C_x^{'}$ and $C_y^{'}$ are the x and y coordinates, respectively, in the simulated COP curve of the corresponding location in the actual COP.

The mean of D ($\overline{D}$) and it standard deviation (s_D) were calculated for each stance phase. The value of $\overline{D}$ indicates the average COP deviation from the smooth COP curve, and s_D indicates its variability. The walking stability decreases when $\overline{D}$ and s_D are large.

All of the aforementioned parameters were calculated from the two complete gait cycles randomly selected within each trial. A factorial analysis of variance was used to investigate the effects of the HBS, walking speed, and slope angle in different stance phases. The Student–Newman–Keuls (SNK) test was used to determine the differences in significant interaction effects. Additionally, a *t* test was applied to examine the effect of the HBS on the subjective measure of walking stability.

3. Results

3.1. Center of pressure

For the locations of the COP ($\overline{C_x}$ and $\overline{C_y}$), a small HBS caused a medial shift in the COP compared with a large HBS during the LR, MS, and TS phases (Table 1 and Fig. 4). The HBS affected the location of the COP in the anteroposterior direction only at the end of the stance phase, and the location was nearer to the toe when the HBS was small (Table 1). The slope angle mainly affected the COP locations in the anteroposterior direction before the PS phase, and the COP was situated in a posterior location when the slope angle was large. The walking speed mainly affected the locations of the COP in the anteroposterior direction during the MS, TS and PS phases, and the COP location was situated more toward to the heel when the speed was slow. Interactions between the HBS and slope angle for $\overline{C_x}$ in the MS phase (p = .012), for $\overline{C_y}$ in the LS (p = .019) and MS (p = .044) phases, and between the walking speed and slope angle for $\overline{C_x}$ in the LS phase (p = .014) were significant. The results of the SNK test are shown in Fig. 5. The changes in the COP location were greater when the HBS was large compared with when the HBS was small when the slope angle changed.

In terms of the deviations of COP ($\overline{D}$ and s_D), both a small HBS and small slope angle significantly increased the magnitudes, particularly in the LR phase (Table 1). Further SNK testing revealed that the interaction between the HBS and the slope angle for $\overline{D}$ (p = .033) and for s_D (p = .018) in the LR phase was significant. A small HBS had significantly larger $\overline{D}$ and s_D than did a large HBS when the slope angle was small during the LR phase. However, when the slope angle was large, the differences in $\overline{D}$ and s_D between the two HBSs were small (Fig. 5). The deviations of the COP ($\overline{D}$ and s_D) were not significantly different between walking speeds.

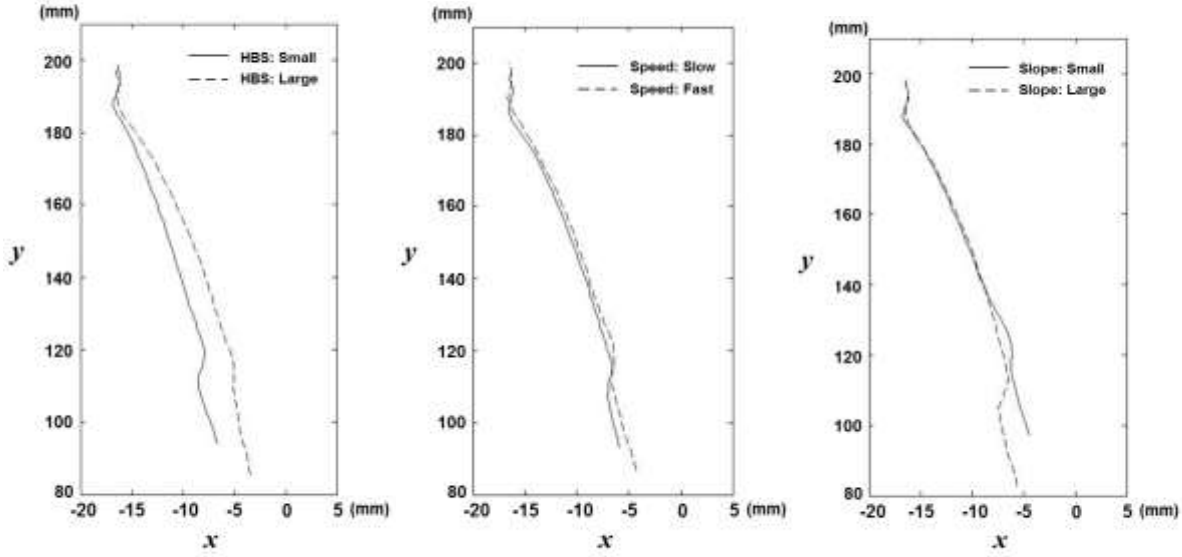
Table 1

Main effects of the HBS, walking speed and slope angle on COP variables (Mean, Standard Deviation and p value are listed)

Main Effect		The HBS		Walking speed		Slope angle	
		Small	Large	Slow	Fast	Small	Large
$\overline{C_x}$ (mm)	LR	**-8.01 ±3.42**	**-4.79 ±3.21**	**-6.78 ±3.98**	**-6.02 ±3.32**	**-5.87 ±3.37**	**-6.94 ±3.88**
		0.000		**0.026**		**0.002**	
	MS	**-8.21 ±3.16**	**-5.56 ±4.13**	-6.89 ±3.98	-6.83 ±3.83	-6.94 ±3.72	-6.83 ±4.08
		0.000		0.918		0.722	
	TS	**-12.95 ±3.77**	**-11.37 ±4.90**	-12.09 ±4.59	-12.24 ±4.28	**-12.70 ±4.18**	**-11.63 ±4.59**
		0.000		0.778		**0.018**	
	PS	-16.42 ±5.15	-16.01 ±5.25	-16.12 ±5.51	-16.27 ±4.85	-16.27 ±5.20	-16.12 ±5.15

		0.443		0.819		0.761	
$\overline{C_y}$ (mm)	LR	106.66 ±21.87	104.82 ±22.46	106.08 ±24.11	105.41 ±20.07	**113.55 ±18.04**	**97.94 ±23.16**
		0.376		0.747		**0.000**	
	MS	118.35 ±21.50	119.47 ±20.30	**116.56 ±20.13**	**121.25 ±21.41**	**125.75 ±19.44**	**112.07 ±20.08**
		0.567		**0.017**		**0.000**	
	TS	162.65 ±10.89	162.69 ±20.04	**160.43 ±17.26**	**164.91 ±14.58**	**166.37 ±14.12**	**158.97 ±17.13**
		0.976		**0.004**		**0.000**	
	PS	**193.18 ±12.00**	**189.35 ±14.18**	**189.46 ±15.21**	**193.07 ±10.71**	191.36 ±13.14	191.18 ±13.41
		0.004		**0.006**		0.891	
$\overline{D}$ (mm)	LR	**2.30 ±1.51**	**1.82 ±0.76**	2.04 ±1.20	2.08 ±1.24	**2.20 ±1.40**	**1.91 ±0.99**
		0.000		0.728		**0.015**	
	MS	1.22 ±0.98	1.16 ±0.67	1.15 ±0.77	1.24 ±0.90	1.23 ±0.94	1.15 ±0.72
		0.483		0.294		0.323	
	TS	0.97 ±0.63	0.90 ±0.44	0.93 ±0.50	0.94 ±0.58	0.95 ±0.63	0.91 ±0.45
		0.217		0.744		0.469	
	PS	1.23 ±1.03	1.20 ±1.45	1.25 ±1.38	1.18 ±1.12	1.28 ±1.32	1.15 ±1.18
		0.760		0.592		0.277	
s_D (mm)	LR	**1.90 ±1.46**	**1.54 ±0.80**	1.63 ±1.02	1.81 ±1.33	**1.91 ±1.37**	**1.53 ±0.94**
		0.002		0.111		**0.001**	
	MS	0.81 ±0.69	0.75 ±0.48	0.75 ±0.54	0.81 ±0.64	0.81 ±0.68	0.75 ±0.49
		0.368		0.254		0.304	
	TS	0.77 ±0.56	0.71 ±0.40	0.72 ±0.42	0.76 ±0.54	0.75 ±0.54	0.74 ±0.43
		0.272		0.475		0.807	
	PS	0.82 ±0.66	1.07 ±1.84	1.01 ±1.77	0.88 ±0.84	1.00 ±1.38	0.89 ±1.39
		0.071		0.321		0.395	

Note: Bold values show significance at p< .05.



(a) (b) (c)

Fig. 4. The average trajectory of the center of pressure under different conditions: (a) HBS (b) Walking speed (c) Slope angle

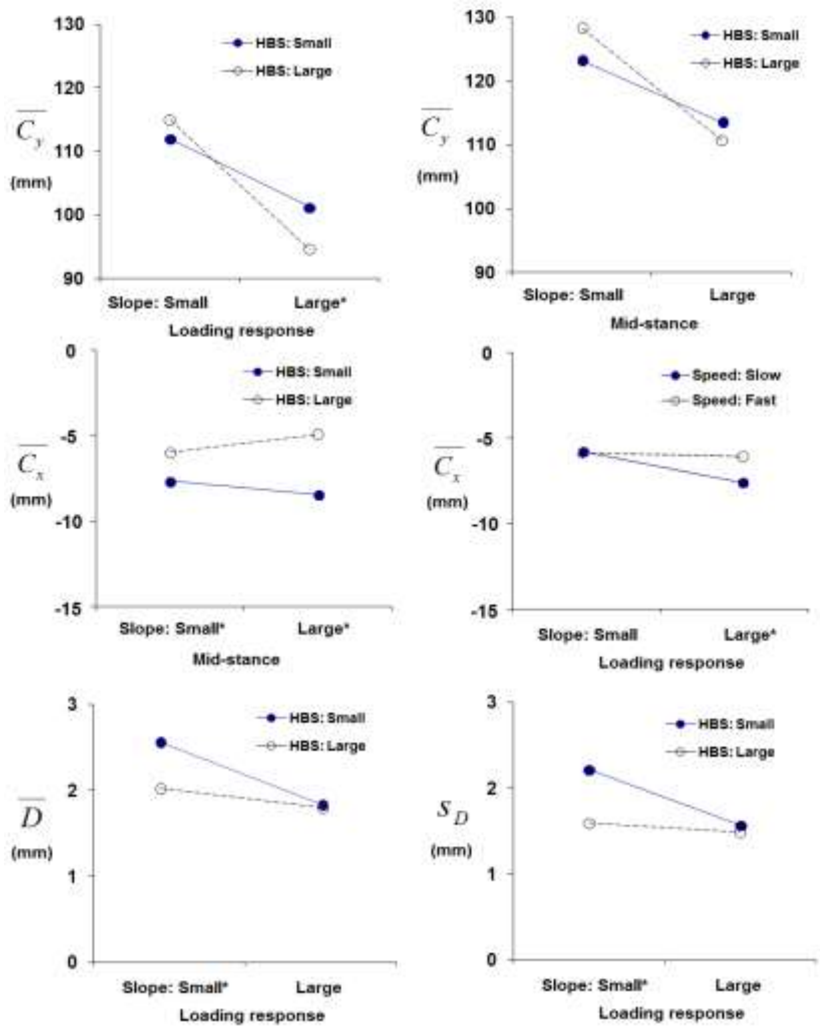


Fig. 5. Interaction effects on the COP variables (*where significant at $p < .05$ for SNK test)

Table 2

Main effects of the HBS, walking speed and slope angle on plantar pressure variables (Mean, Standard Deviation and *p* value are listed)

Main Effect		The HBS		Walking speed		Slope angle	
		Small	Large	Slow	Fast	Small	Large
Max. Peak Pressure (kPa)	Toe	**237.53 ±130.51**	**211.54 ±124.00**	217.29 ±125.12	231.78 ±130.34	221.06 ±127.07	228.01 ±128.76
		0.045		0.262		0.590	
	Forefoot	**517.04 ±247.68**	**578.31 ±312.86**	520.07 ±266.73	575.29 ±297.38	524.86 ±268.56	570.50 ±296.55
		0.032		0.053		0.109	
	Midfoot	**13.36 ±29.02**	**25.54 ±36.38**	20.06 ±33.97	18.84 ±32.95	**9.52 ±24.31**	**29.39 ±38.09**
		0.000		0.698		**0.000**	
	Rearfoot	**174.97 ±47.14**	**210.65 ±87.07**	191.38 ±72.99	194.24 ±71.49	188.39 ±61.12	197.23 ±81.66
		0.000		0.684		0.209	
Pressure Time Integral (kPa*sec)	Toe	49.70 ±19.65	47.58 ±21.48	50.48 ±21.16	46.80 ±19.89	48.20 ±20.82	49.09 ±20.40
		0.308		0.077		0.668	
	Forefoot	79.06 ±32.42	78.09 ±32.36	81.74 ±33.29	75.40 ±31.14	79.03 ±30.98	78.12 ±33.73
		0.767		0.054		0.781	
	Midfoot	**3.38 ±8.63**	**5.66 ±9.25**	4.87 ±9.58	4.18 ±8.40	**2.17 ±6.74**	**6.88 ±10.30**
		0.009		0.422		**0.000**	
	Rearfoot	43.46 ±12.22	44.58 ±17.92	**47.34 ±15.51**	**40.70 ±14.44**	**42.13 ±14.04**	**45.92 ±16.34**
		0.455		**0.000**		**0.012**	
Pressure Contact Area (mm^2)	Toe	**635.13 ±318.67**	**518.87 ±267.60**	**539.86 ±286.31**	**614.14 ±308.55**	568.61 ±300.26	585.39 ±299.42
		0.000		**0.013**		0.571	
	Forefoot	2034.07 ±354.55	2002.32 ±276.95	2004.09 ±311.36	2032.30 ±324.91	2013.87 ±311.16	2022.52 ±325.66
		0.325		0.381		0.788	
	Midfoot	**45.91 ±116.89**	**72.95 ±115.70**	58.68 ±112.01	60.18 ±121.94	**29.03 ±88.53**	**89.83 ±133.13**
		0.016		0.893		**0.000**	

	Rearfoot	1723.21 ±331.17	1727.96 ±402.22	1734.43 ±398.57	1716.73 ±335.33	**1625.69 ±321.41**	**1825.48 ±384.84**
		0.894		0.620		**0.000**	

Note: Bold values show significance at $p < .05$.

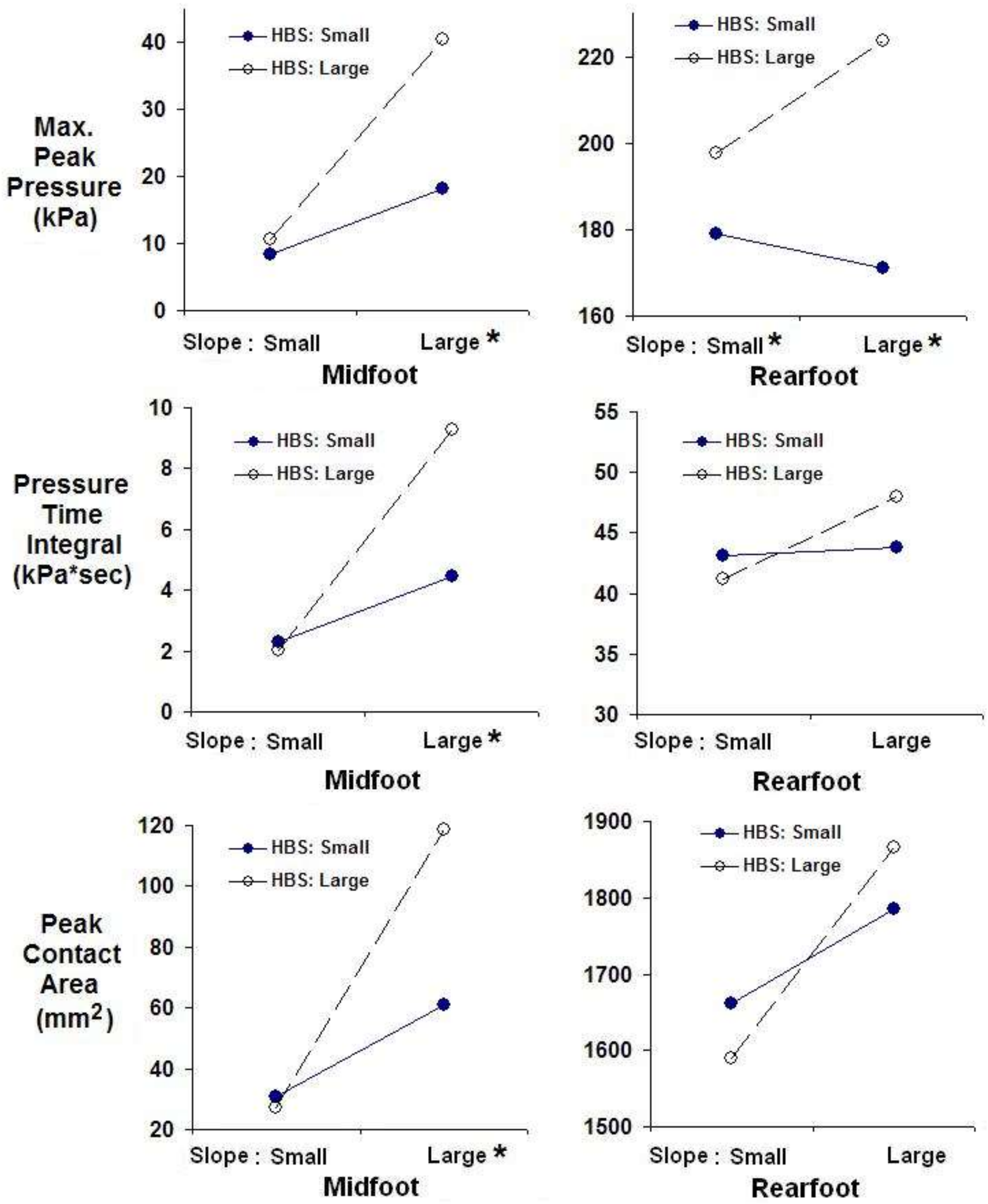


Fig. 6. Interaction effects on plantar pressure variables (*where significant at $p < .05$ for SNK test)

3.2. Plantar pressure distributions

The HBS changed the maximal peak pressure over the entire foot region (Table 2). A large HBS increased the maximal peak pressure over the forefoot, midfoot, and rearfoot, whereas a small HBS caused a higher maximal peak pressure over the toe region. The slope angle mainly affected the pressure over the midfoot and rearfoot. A large slope angle increased the pressure time integral and the peak contact areas over midfoot and rearfoot regions compared with a small slope angle. Interactions between the HBS and the slope angle for maximal peak pressure over the midfoot (p = .001) and rearfoot regions (p = .017), for pressure time integral over the midfoot (p = .003) and rearfoot regions (p = .044), and for peak contact area over the midfoot (p = .007) and rearfoot regions (p = .031) were significant at p < .05. Fig. 6 shows the SNK test results. A large HBS caused significantly larger plantar pressure values in the midfoot than did a small HBS when the slope angle was large. The walking speed exerted less influence on the parameters related to plantar pressure. The fast speed increased only the peak contact area over the toe region.

3.3. Visual analog scale assessment of perceived stability

A significantly higher perception rating (mean ± SD) was revealed when the HBS was large (75.03 ± 4.00) compared with when the HBS was small (39.03 ± 6.20). These results indicated that the participants felt that a small HBS was more unstable to wear.

4. Discussion

In this study, the effects exerted by the HBS of high-heeled shoes, walking speed, and slope angle on the walking stability and perceived stability were investigated. The HBS significantly influenced the trajectories of the COP and the plantar pressure distributions, thus changing the walking stability and perceived stability.

The larger deviation of COP observed in this study (the mean and standard deviation of the COP) suggested that the foot was more unstable with a small HBS than with a large HBS during the LR phase, particularly when the slope angle was small. During the LR phase, the body weight is transferred onto the stance limb, causing a shift in the center of body mass from one leg to the other (Perry, 1992). Additionally, the ankle joint is placed in a plantarflexion position during this period, and the plantarflexion angle is increased when wearing high-heeled shoes. The plantarflexion of the ankle can result in a less stable ankle joint than that in a neutral or dosiflexed position (close packed position) because of the anatomical properties of the talus (Willems, Witvrouw, Delbaere, DeCock, & Clercq, 2005). The shift in the center of body mass and the plantarflexion position both reduce the stability of the body and the ankle joint. A large swing of the foot at the beginning of the gait and less body and ankle joint stability might increase the risk of ankle injuries, particularly when wearing high-heeled shoes with a small heel base.

In this study, the COP locations had a medial and anterior shift from the foot center line, which was similar as a previous study (Han et al., 1999). From the LR to TS phases, the COP locations had a further medial shift when the HBS was small compared with when the HBS was large (Fig. 4[a]). A greater medial shift of the COP might increase the torque at the knee and ankle joints, generated by the ground reaction force (Kerrigan, Lelas, & Karvosky, 2001; Kim et al., 2003). To compensate for the divergence in the COP, greater muscle force is required to maintain body or joint stability, and thus, muscle fatigue is easily induced (Gefen et al., 2002). These changes might increase the risk of instability, and knee and ankle problems.

The HBS significantly influenced plantar pressures over all foot regions. The small HBS significantly increased the maximal peak pressure and peak contact area over the toe region compared with the large HBS, possibly because the stability when wearing high-heeled shoes with a small HBS was reduced. When the HBS is small, the toes must grasp the sole of the shoe when the heel is unstable. This may cause toe problems such as hammertoe.

The slope angle is another critical factor affecting the COP and plantar pressure distribution. In this study, the foot appeared to be more stable when walking on the large slope compared with when walking on the small slope in terms of the COP deviations. This observation might be explained by the fact that walking on a large slope reduces the ankle plantarflexion (Leroux et al., 2002), which is also the reason the COP shifted to a posterior location when the participants walked on the large slope. The posterior shift of the COP caused greater plantar pressure and a larger contact area over the midfoot and rearfoot. A larger contact area provides more foot support, enabling walking stability. The results from this study also revealed that the interactions between the slope angle and the HBS were significant and the effects of the HBS differed with the slope angle. The small HBS had a larger COP deviation ($\overline{D}$ and s_D) than did the large HBS when the participants walked on the small slope; however, the difference was not significant on the large slope (Fig. 5). The large HBS provided more midfoot and rearfoot plantar supports than did the small HBS only on the large slope (Fig. 6). Additionally, the HBS mainly affected the COP position in the mediolateral direction, whereas the slope angle mainly affected the COP position in the anteroposterior direction (Fig. 5). The small slope angle created similar effects as those produced by an increased heel height; thus, the downward inclination might affect walking stability differently, which should be investigated in the future.

Walking speed exerted a limited influence on the COP deviations and plantar pressure. A possible reason for this result is that the two speeds used in this study were within the comfortable speed range (110 to 145 cm/s) for high-heeled shoe wearers reported in previous studies (Opila-Correia, 1990; Snow & Williams, 1994). Future studies should investigate a larger range of walking speeds.

In this study, walking stability was evaluated in separate gait phases, a method that can afford more insight into walking stability as affected by shoe designs in a specific gait instance or phase. Because humans tend to make compensatory postural adjustments when walking stability changes, kinematic studies and the measurement of the electromyography of muscles are warranted. The results from the present study can explain various key aspects of plantar loading changes and facilitate understanding the mechanism underlying how the foot loses and maintains stability. However, only one heel height was investigated in this study. Although researchers claimed that the reduced heel base rather than the increased heel height was the primary factor for the reduced stability during a level walking (Chien et al., 2013), the heel height might cause some potential complication in explaining the interaction effects of the heel base size on the plantar pressure and walking stability in a slope walking. Compared with level walking, the increased slope angle increased the peak pressure over the medial forefoot, midfoot and toe regions when flat shoes were used (Shen, Ma, Li, & Gu, 2013). In our study, the increased peak pressure was not significant over the medial forefoot and toe regions with high-heeled shoe when the slope angle increased. The increased slope angle can have some opposite effects on the body adjustment with the heel height elevation such as the decreased ankle plantarflexion (Leroux et al., 2002). The effects of the HBS and the changed heel height on walking stability warrant further investigation to facilitate a deeper understanding of the combined effects of shoe design factors.

This study was laboratory-based. Although temporal gait parameters between treadmill and overgound walking had very few differences, some changes in muscle activation patterns and joint moments were observed (Lee & Hidler, 2008; Riley, Paolini, Croce, Paylo, & Kerrigan, 2007). Therefore, treadmill walking might cause potential differences in measurement when compared with overground walking. In addition, although the tasks were performed during approximately 2 hours in this study, people might stand or walk for longer durations in a real work environment (Lee & Hong, 2004). A long-term investigation involving real-life conditions is crucial for providing insight into the behavioral and physical adaptations of the foot to maintain stability.

5. Conclusions

The HBS and slope angle significantly influenced the COP trajectory and plantar pressure patterns. Compared with the large HBS, the small HBS increased the COP deviations, shifting the COP more medially at the beginning of the gait cycle. These changes impair walking stability, particularly in the LR phase when the foot is in a plantarflexion position and the body weight is transferred onto the stance limb. The slope angle mainly affected the COP in the anteroposterior direction, and an increased slope angle shifted the COP posteriorly. The posterior shift in the

COP caused greater pressure and a larger contact area over the midfoot and rearfoot, thus providing more support to the foot. However, a significant interaction was observed between the HBS and the slope angle on the COP trajectory. The results revealed that high-heeled shoes with a small HBS increased the COP deviations compared with a large HBS, particularly on a small slope, whereas the differences were not obvious on a large slope. The objective measures suggested that high-heeled shoes with a large HBS might increase stability during level walking, an indication that was consistent with the perceived stability of the participants.

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