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Effect of wearing high-heeled shoes on postural control and foot loading symmetry

Michalina Blazkiewicz[1]*, Anna Hadamus[2]

[1]Faculty of Rehabilitation, The Józef Piłsudski University of Physical Education in Warsaw, Warsaw, Poland

[2]Department of Rehabilitation, Faculty of Dental Medicine, Medical University of Warsaw, Warsaw, Poland

*Corresponding author: Michalina Blazkiewicz, Department of Rehabilitation, Faculty of Dental Medicine, Medical University of Warsaw, Warsaw, Poland, e-mail address: michalinablazkiewicz@gmail.com

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1. Introduction

High-heeled shoes (HHS) play a particular role in the history of footwear [9]. The ancient Egyptians and Greeks wore shoes with raised heels for ceremonial and practical purposes. HHS gained popularity in Europe during the 15th century. Persian-inspired riding shoes with heels became fashionable among European aristocrats, both men and women. These shoes were a symbol of elite status. HHS fell out of favor for men in the latter half of the 18th century as fashion moved towards more practical and comfortable footwear [24]. However, heels remained popular among women, are still seen as a symbol of femininity and elegance, and have become an essential part of fashion [25].

Nowadays, the goal of footwear is to provide: protection (against potential hazards, such as sharp objects, extreme temperatures, or falling objects); support (fitted shoes with adequate arch support can help prevent foot and ankle pain and reduce the risk of sprains or strains); comfort (provide comfort by cushioning the feet and reducing the impact of walking or running on hard surfaces); performance enhancement (in sports, footwear should provide the necessary traction, flexibility, and support for particular activities, optimizing performance and reducing the risk of injury); style and fashion preferences (shoes can reflect individual choices, cultural trends, and social norms); medical and orthopedic support (specialized shoes or orthotics help address specific foot conditions, such as flat feet, high arches, or plantar fasciitis) [2; 26].

Reutimann et al. [29] showed that shoes significantly affect postural control by altering the base of support through changes in shape and size. Most studies have shown that wearing HHS with heels larger than 7cm declines balance [13; 16], reduces gait stability [39], and increases the risk of falls and ankle injuries [3; 27]. All this occurs because HHS places the feet in a more plantarflexed and supinating position [35]. This configuration reduces the range of motion of the ankle joint and thus affects the effectiveness of ankle strategies for postural control [37]. Many studies have highlighted that the effects of wearing HHS are not limited to the foot-ankle complex. Kinematic effects are transmitted up the lower limb in a chain reaction [7], ultimately leading to changes in kinematics [19], kinetics [38], muscle activity

[12], and energy expenditure [8]. Silva et al. [33], reviewing the papers in this area, showed that usage of HHS promotes the appearance of postural disorders, particularly forward head tilt, lumbar hyperlordosis, pelvic anterior tilt, and knee valgus. They also observed that heel height and width most affected posture and imbalance. Available evidence suggests that walking in HHS requires special neural control that differs from that used during barefoot walking [1]. If this is the case, it is most likely that a different neuronal control occurs during free-standing. Nonlinear parameters provide insight into such control, as reported by Kedziorek and Blazkiewicz [18]. Nonlinear measurements capture the variability, adaptability, and coordination of movement patterns, allowing insight into the complexity and dynamics of postural control. Of all the measures of nonlinear dynamics, sample entropy (SampEn), fractal dimension (FD), and Lyapunov exponent (LyE) appear to be the most commonly used to assess postural control [4-6; 20; 21]. SampEn quantifies the irregularity or unpredictability of a time series signal. Lower SampEn values indicate more regular and predictable movement patterns, while higher values indicate the system's readiness for an unexpected stimulus [20]. FD assesses the complexity of body sway during standing. This measure quantitatively measures the self-similarity or self-repeating patterns presented during body adjustment when maintaining balance. A higher FD indicates greater complexity and adaptability, which means the body is constantly making fine adjustments to stabilize itself [10]. LyE is a measure that assesses the resistance of the human control system to perturbations. Low LyE values indicate the rigidity of the system and its inability to adapt to the environment. High LyE values indicate the ability to respond more quickly to destabilizing factors and better balance control [18].

So far, no assessment of the complexity, variability, and adaptation of postural control while standing in high-heeled shoes has been found in the current literature. Therefore, the aim of this study is to analyse the impact of 11cm high heels and prior experience with high heels on foot pressure distribution and balance by analysing linear and nonlinear oscillation parameters of the center of pressure in mediolateral (ML) and ~~anterioposterior~~anteroposterior (AP) directions.

2. Materials and Methods

2.1. Participants

Sixteen young female HHS wearers experts (HHE) and sixteen young females occasionally wearing high-heeled shoes (HHO) participated in this study (Table 1). The groups' size was determined based on Zeng's et al. [39] review and meta-analysis, where the authors reported sample sizes ranging from 3 to 71, with 15 being the most common.

Table 1. Characteristics of the participants (mean ± standard deviation).

Group	Age [years]	Body weight [kg]	Body height [cm]
HHE: n = 16	28.06 ± 6.46	60.31 ± 5.87	165.81 ± 4.61
HHO: n = 16	31.81 ± 9.68	62.25 ± 7.56	166.75 ± 3.86

HHS wearers experts were women who had worn shoes with a minimum heel height of 7cm three or more times per week in the past two years. All women from the HHE group are dancers or ex-dancers in high-heels with about five years of experience in this dance style. During those years, training was scheduled four times a week for about two hours each. These dancers danced on stilettos with a height of eight to ten centimeters. Women from the HHO group declared that they wear this type of footwear only occasionally (no more than ten times a year) and have no experience in dancing on stilettos. For the study, the selected shoes were those with an 11cm heel and a thin stiletto ($1cm^2$) (Fig. 1). It aimed to create unfamiliar conditions in both groups. All participants were tested using the same pair of shoes.
All participants had a shoe size of EU 38-40 and reported being free from lower limb injuries for a minimum of six months before the study. Moreover, all of them declared to have a dominant right leg. According to Promsri et al. [28], the dominant leg was the preferred leg for kicking the ball. All participants gave their informed consent to participate in the research, which had previously been approved by the university's institutional review board (no. SKE01-15/2023). The study followed ethical guidelines and the principles of the Declaration of Helsinki.

2.2. Measurement protocol

The four tests were evaluated in random order (Fig. 1). The tests consisted of standing on both legs with eyes open and upper limbs alongside the trunk, wearing shoes with 11cm heels or being barefoot (BF). The tests were performed on both the Biodex Balance System SD tilting platform (Biodex, Shirley, NY, USA) and the Zebris FDM platform (Zebris Medical GmbH, Germany). Each test lasted 20 seconds. A two-minute rest was provided between each condition to prevent fatigue. On the Biodex plate, each participant underwent a Fall Risk Test (FRT), during which the platform changed stability from very unstable to slightly unstable (from 6 to 2). On the Zebris platform (100Hz), the participants were instructed to stay still, looking at the white wall a meter ahead of them. The tested subjects were instructed to position their feet identically for both measurements to align feet identically for both measurements, maintaining a distance between their feet equivalent to their hip width. For the Biodex platform, it was possible to determine the coordinates of foot position, which remained the same for standing barefoot and in heeled shoes.

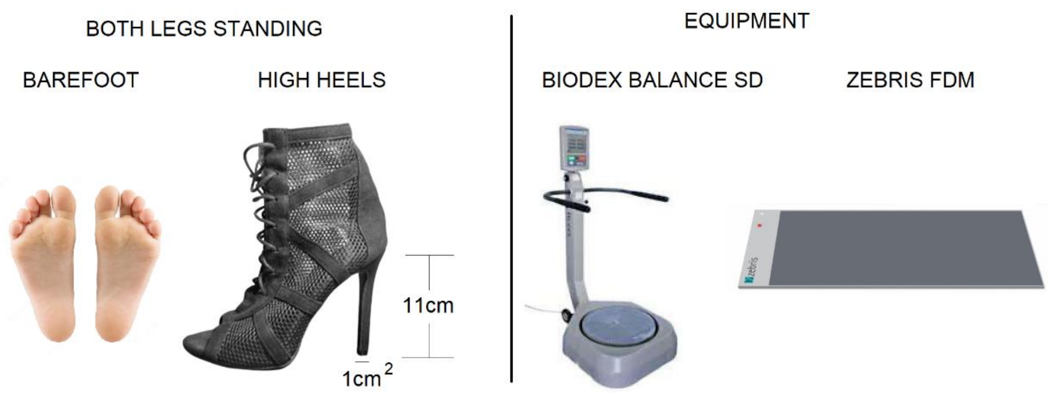

Figure 1. Both legs standing conditions (Barefoot and High-heeled shoes) and equipment used (Biodex and Zebris platforms) for the four test conditions.

2.3. Parameters and statistical analysis

A total of fifteen parameters were used for statistical analysis. From the Biodex platform, the FRT index was acquired, with a higher value indicating an increased risk of falling.

Eight parameters were extracted from the Zebris platform. Two variables, namely center of pressure (CoP) path length [mm] and average velocity [mm/s], were used to assess stability.

Six additional parameters: average forefoot force [%], backfoot force [%], and total force [%] for both the left and right lower limbs were calculated to evaluate foot loads. According to the Zebris FDM software manual [14], the measurement presents the distribution of relative forces as a percentage, divided between the left and right foot or between the forefoot and heel. Therefore, the total should be 100% within the body or for each individual foot, respectively.

Additionally, based on the center of pressure time series in the anterior-posterior (AP) and mediolateral (ML) directions, values for three nonlinear parameters were computed: sample entropy (SampEn), fractal dimension (FD), and the largest Lyapunov exponent (LyE).

SampEn calculates the probability that a sequence of N-data points, having repeated itself within a tolerance r for m points, will also repeat itself for m+1 points, without allowing self-matches: $SampEn(m,r,N) = -ln\left(\frac{A^m(r)}{B^m(r)}\right)$. B represents the total number of matches of length m, while A represents the subset of B that also matches for m+1. Thus, a low SampEn value arises from a high probability of repeated template sequences in the data, hence greater regularity. For calculating the SampEn, we used the MatLab codes obtained from the the Physionet tool [15] with "default" parameters: m = 2, r = 0.2*SD, where SD is standard deviation.

The FD was calculated using Higuchi's algorithm [17], which is particularly well applied to short time series.

LyE is a measure of the local stability of a system, i.e., its resistance to small internal perturbations, such as the natural fluctuations that occur while maintaining an upright stance [31]. The concept of using LyE to identify chaos in a system comes from the idea that if the average distance between two points grows exponentially, the system is sensitive to a change in initial conditions, and the value of LyE is greater than zero. Thus, LyE is defined by the following equation: $d(t) = Ce^{LyEt}$, where: d(t) is the average divergence at time t, and C is a constant that normalizes the initial separation. Therefore, the presence of a positive LyE is considered a necessary and sufficient condition for the presence of chaos in the system often.

Statistical analysis was performed using Statistica 13.1 (TIBCO Software, Inc., Palo Alto, CA, USA), and the cut-off p-value was set at 0.05.

The normality of the distributions of the abovementioned parameters was assessed using the Shapiro-Wilk test. Using the factorial ANOVA with post-hoc Tukey HSD the effect of group (HHE/ HHO), standing conditions (BF/ HHS standing), and interaction effects (groups x standing conditions) were assessed. Then, within groups, the effects foot loading parameters (left/ right and forefoot/ backfoot) were examined using the t-test for depended groups. A partial eta squared (η^2) value was assigned for each parameter as a measure of effect size. The interpretation of the η^2 value follows the study [30], where $0.01 \leq \eta^2 < 0.06$ denotes a small effect, $0.06 \leq \eta^2 < 0.14$ indicates a moderate effect, and $0.14 \leq \eta^2 \leq 1$ signifies a large effect.

3. Results

The results presented in this chapter include those describing foot loading while standing barefoot and in heeled shoes, postural stability assessed using linear parameters, and nonlinear parameters (Table 2).

When analysing the combination of comparisons: forefoot force vs. backfoot force within the right and left lower limb separately, as well as the comparison of forefoot and backfoot force between the right and left foot, no statistically significant differences were found in either group when standing barefoot and in high-heeled shoes (Fig. 2).

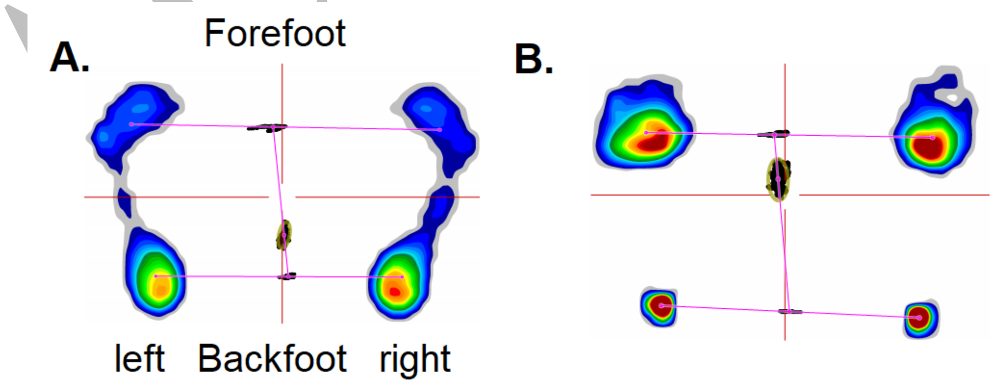

Figure 2. Examples of average foot loading for one person while A. barefoot, both legs standing, and B. both legs standing in high-heeled shoes.

Statistically significant differences were found in both groups for the total force left and right parameter (Fig. 3). While barefoot standing, all individuals (HHO and HHE group) loaded the right foot significantly more strongly. In heeled shoes, the load on the foot was the opposite. All women loaded the left foot significantly more strongly (Fig. 3, Table 2).

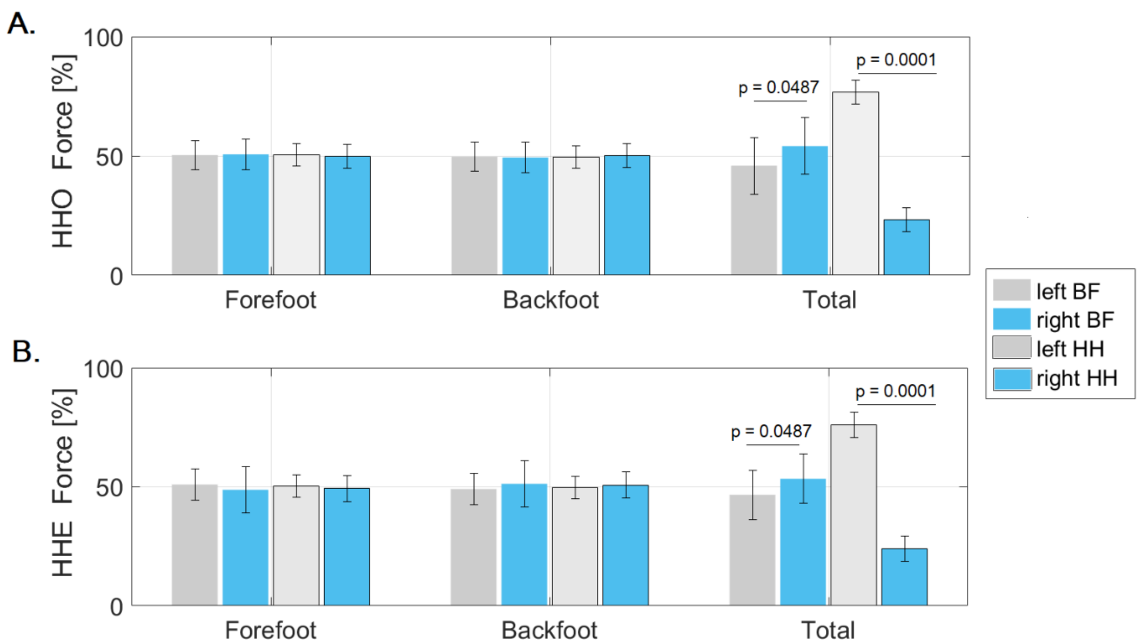

Figure 3. Forefoot, backfoot and whole foot loading for group A. HHO and B. HHE, where only statistically significant differences ($p < 0.05$) between the right and left lower limbs for standing barefoot (BF) and in heeled shoes (HH) are marked.

~~Insert Table 2~~

~~Table 2. Mean and standard deviation values of parameters for between-group comparisons (HHO and HHE) under different standing conditions (barefoot (BF) and in heeled shoes (HH)). Statistically significant differences are denoted by *, with a significance level of $p < 0.05$. ML refers to the medio-lateral direction, and AP refers to the anterio-posterior direction.~~

Through analysis of linear parameters assessing postural stability (Table 2), there were no statistically significant differences for the fall risk assessment index between the groups and between BF and HH standing. However, in the HHE group, standing in heeled shoes increased the FRT index value in a significant way. Furthermore, wearing heeled shoes significantly increased CoP path length and sway velocity in both groups.

Table 2. Mean and standard deviation values of parameters for between-group comparisons (HHO and HHE) under different standing conditions (barefoot (BF) and in heeled shoes (HH)). Statistically significant differences are denoted by *, with a significance level of $p < 0.05$. ML refers to the medio-lateral direction, and AP refers to the anterio-posterior direction.

Parameters	**Groups (HHO vs. HHE)**	**Conditions (BF vs. HH)**	**Interaction (Groups x Conditions)**
Foot loading assessment			
Forefoot force left [%]	(50.39 ± 5.48 vs. 50.62 ± 5.65), $p = 0.8735$, $\eta^2 = 0.06$	(50.63 ± 6.29 vs. 50.38 ± 4.73), $p = 0.8735$, $\eta^2 = 0.06$	$p = 0.0722$, $\eta^2 = 0.01$
Forefoot force right [%]	(50.27 ± 5.79 vs. 49.03 ± 7.79), $p = 0.4782$, $\eta^2 = 0.06$	(49.68 ± 8.21 vs. 49.62 ± 5.27), $p = 0.9697$, $\eta^2 = 0.06$	$p = 0.0722$, $\eta^2 = 0.01$
Backfoot force left [%]	(49.60 ± 5.48 vs. 49.37 ± 5.65), $p = 0.8735$, $\eta^2 = 0.06$	(49.36 ± 6.29 vs. 49.61 ± 4.73), $p = 0.8570$, $\eta^2 = 0.06$	$p = 0.1317$, $\eta^2 = 0.01$
Backfoot force right [%]	(49.72 ± 5.79 vs. 50.96 ± 7.79), $p = 0.4782$, $\eta^2 = 0.06$	(50.31 ± 8.21 vs. 50.37 ± 5.27), $p = 0.9697$, $\eta^2 = 0.06$	$p = 0.1317$, $\eta^2 = 0.01$
Total force left [%]	(61.67 ± 17.89 vs. 61.34 ± 16.98), $p = 0.8795$, $\eta^2 = 0.06$	(46.57 ± 10.98 vs. 76.45 ± 5.21), $p = 0.0001$*, $\eta^2 = 0.14$	$p = 0.0394$*, $\eta^2 = 0.06$ HHO BF < HHO HH, $p = 0.0001$* (46.52 ± 11.99 < 76.83 ± 5.24) HHO BF < HHE HH, $p = 0.0001$* (46.52 ± 11.99 < 76.07 ± 5.31) HHO HH > HHE BF, $p = 0.0001$* (76.83 ± 5.24 > 46.62 ± 10.26) HHE BF < HHE HH, $p = 0.0001$* (46.62 ± 10.26 < 76.07 ± 5.31)
Total force right [%]	(38.32 ± 17.89 vs. 38.65 ± 16.98), $p = 0.8795$, $\eta^2 = 0.06$	(53.42 ± 10.98 vs. 23.54 ± 5.21), $p = 0.0001$*, $\eta^2 = 0.14$	$p = 0.0021$*, $\eta^2 = 0.06$ HHO BF > HHO HH, $p = 0.0001$* (53.47 ± 11.99 > 23.16 ± 5.24) HHO BF > HHE HH, $p = 0.0001$* (53.47 ± 11.99 > 23.92 ± 5.31) HHO HH < HHE BF, $p = 0.0001$* (23.16 ± 5.24 < 53.37 ± 10.26) HHE BF > HHE HH, $p = 0.0001$* (53.37 ± 10.26 > 23.92 ± 5.31)
Linear measures of postural control assessment			
FRT index	(0.98 ± 0.44 vs. 0.94 ± 0.37), $p = 0.8046$, $\eta^2 = 0.06$	(0.86 ± 0.39 vs. 1.05 ± 0.41), $p = 0.0673$, $\eta^2 = 0.05$	$p = 0.0209$*, $\eta^2 = 0.06$ HHE BF < HHE HH, $p = 0.0209$* (0.77 ± 0.26 < 1.11 ± 0.40)

CoP path length [mm]	(236.97 ± 125.83 vs. 202 ± 91.57), p = 0.1342, η^2 = 0.07	(158.08 ± 62.44 vs. 280.89 ± 114.75), p = 0.0001*, η^2 = 0.32	p = 0.0001*, η^2 = 0.06 HHO BF < HHO HH, p = 0.0005* (177.60 ± 68.01 < 296.34 ± 143.46) HHO BF < HHE HH, p = 0.0091* (177.60 ± 68.01 < 265.44 ± 78.26) HHO HH > HHE BF, p = 0.0001* (296.34 ± 143.46 > 138.56 ± 51.18) HHE BF < HHE HH, p = 0.0002* (138.56 ± 51.18 < 265.44 ± 78.26)
CoP velocity [mm/s]	(11.84 ± 6.29 vs. 10.10 ± 4.57), p = 0.1342, η^2 = 0.07	(7.90 ± 3.12 vs. 14.04 ± 5.73), p = 0.0001*, η^2 = 0.32	p = 0.0001*, η^2 = 0.06 HHO BF < HHO HH, p = 0.0005* (8.88 ± 3.40 < 14.81 ± 7.17) HHO BF < HHE HH, p = 0.0090* (8.88 ± 3.40 < 13.27 ± 3.91) HHO HH > HHE BF, p = 0.0001* (14.81 ± 7.17 > 6.92 ± 2.55) HHE BF < HHE HH, p = 0.0002* (6.92 ± 2.55 < 13.27 ± 3.91)
Nonlinear measures of postural control assessment			
SampEn ML	(0.15 ± 0.09 vs. 0.21 ± 0.11), p = 0.0152*, η^2 = 0.09	(0.16 ± 0.12 vs. 0.20 ± 0.09), p = 0.0703, η^2 = 0.07	p = 0.0022*, η^2 = 0.07 HHO BF < HHO HH, p = 0.0466* (0.11 ± 0.07 < 0.18 ± 0.09) HHO BF < HHE BF, p = 0.0153* (0.11 ± 0.07 < 0.20 ± 0.14) HHO BF < HHE HH, p = 0.0032* (0.11 ± 0.07 < 0.22 ± 0.07)
SampEn AP	(0.39 ± 0.20 vs. 0.47 ± 0.25), p = 0.1678, η^2 = 0.07	(0.51 ± 0.26 vs. 0.35 ± 0.15), p = 0.0057*, η^2 = 0.23	p = 0.0051*, η^2 = 0.07 HHO HH < HHE BF, p = 0.0038* (0.32 ± 0.14 < 0.55 ± 0.30) HHE BF > HHE HH, p = 0.0353* (0.55 ± 0.30 > 0.39 ± 0.16)
FD ML	(1.35 ± 0.10 vs. 1.41 ± 0.09) p = 0.0086*, η^2 = 0.10	(1.34 ± 0.09 vs. 1.43 ± 0.09), p = 0.0001*, η^2 = 0.23	p = 0.0134*, η^2 = 0.12 HHO BF < HHO HH, p = 0.0010* (1.30 ± 0.09 < 1.41 ± 0.09) HHO BF < HHE BF, p = 0.0264* (1.30 ± 0.09 < 1.37 ± 0.08) HHO BF < HHE HH, p = 0.0001* (1.30 ± 0.09 < 1.46 ± 0.08) HHE BF < HHE HH, p = 0.0085* (1.37 ± 0.08 < 1.46 ± 0.08)
FD AP	(1.51 ± 0.12 vs. 1.55 ± 0.11) p = 0.1849, η^2 = 0.07	(1.55 ± 0.12 vs. 1.51 ± 0.11), p = 0.2781, η^2 = 0.07	p = 0.8572, η^2 = 0.07
LyE ML	(1.66 ± 0.14 vs. 1.63 ± 0.26) p = 0.5616, η^2 = 0.06	(1.58 ± 0.24 vs. 1.71 ± 0.14), p = 0.0180*, η^2 = 0.08	p = 0.0371*, η^2 = 0.07 HHO HH > HHE BF, p = 0.0371* (1.72 ± 0.16 > 1.56 ± 0.34)
LyE AP	(1.69 ± 0.13 vs. 1.70 ± 0.14) p = 0.7478, η^2 = 0.06	(1.67 ± 0.12 vs. 1.72 ± 0.15), p = 0.1509, η^2 = 0.06	p = 0.6467, η^2 = 0.06

Reporting the behavior of nonlinear parameters along the anterior-posterior direction for both FD and LyE coefficients, there were no statistically significant differences between groups, conditions, and interactions between these factors. In this direction, only for SampEn values was the effect of conditions. The SampEn values were significantly higher during BF than those recorded during HH standing. This result was affected by the HHE group. In this group, SampEn values were significantly higher during BF standing than those recorded for HH standing in both the HHE and HHO groups.

Regarding the mediolateral direction, the differences were statistically significant between the HHO and HHE groups only for SampEn and FD. In both cases, the values of these parameters were higher in the HHE group. In contrast, the significant effect of conditions was for FD and LyE, where higher values occurred during HH standing. The same three interaction effects were noted for SampEn and FD. In both cases, these parameter values were significantly lower during BF in the HHO group than those recorded when standing in heeled shoes in both the HHO and HHE groups and when standing barefoot in the HHE group. In addition, only FD showed significantly higher values for standing in heeled shoes versus those recorded for standing barefoot in the HHE group. The LyE values for the interaction showed only one relationship. Significantly higher values were recorded for standing in heeled shoes in the HHO group against those for standing barefoot in the HHE group.

It is worth nothing that within the parameters assessing foot loading and postural control (linear and non-linear measures), the effect size was consistently at a moderate level (0.06) across most comparisons. Is suggests that the observed differences hold significance and are not trivial, indicating a moderate practical importance of the study's findings. Furthermore, among the parameters examined, six exhibited a large effect size. These parameters include Total force in both left and right foot loading, CoP path length, CoP velocity, SampEn AP, and FD ML. These differences were noted particularly between standing barefoot and in heeled shoes. This outcome indicates that these specific parameters play a significant role in evaluating the impact of wearing high-heeled shoes during free standing.

4. Discussion

This study evaluated the effects of high-heeled shoes and high-heeled experience on foot loading and standing balance using linear and nonlinear methods. This study found that using high-heeled shoes, regardless of experience, negatively impacted stability and balance. Experienced wearers faced over a 44% increased fall risk, with a 78% rise in CoP path length and velocity. Nonlinear measures highlighted disruptions in balance, showing around 6-8% increases in specific parameters. Additionally, high heels caused a 70% rise in total foot load asymmetry.

Postural control refers to the ability of an individual to maintain balance and stability while standing, walking, or performing any other physical activity. Linear and nonlinear measures are both commonly used to study human movement control. Linear measures include traditional ones like the center of pressure path length and velocity [18]. These measures are widely used in clinical and research settings and provide important information about the magnitude and direction of postural sway. Nonlinear parameters, on the other hand, are relatively new and offer a different perspective on postural control because they are based on the principles of chaos theory and are used to analyse the complexity and variability of postural sway over time. Combining these two sets of parameters with a foot-loading assessment appeared to provide a comprehensive answer to how postural control changes when standing in high-heeled shoes.

This study proved that the fall risk in the groups of occasional wearers of HHS and those who wear them frequently is at the same level. However, HHS caused a significant increase in FRT of more than 44%, but only in the HHE group. The linear parameter values (CoP path length and CoP velocity) were not significantly higher in the HHO group than those recorded in the HHE group. However, the presence of HHS caused a significant increase in CoP path length and CoP velocity by almost 78%. The values of these parameters increased by more than 67% in the HHO group and by more than 92% in the HHE group. These results are in line with other studies [16; 37]. Hapsari and Xiong [16] showed that heel height starting at

7cm worsens functional lower limb mobility and standing balance. Wan et al. [37] conducted a more detailed analysis focused on the directionality of linear measures. They showed that the variability of CoP in both the ML and AP directions increased with increasing heel height, but the main effect of heel height appeared to be significant only in the ML direction. At this point, it is worth emphasizing that nonlinear measures are directional and allowed to analyse both directionality and intermediate features related to regularity and complexity. The FD and LyE showed no statistically significant differences for group (HHE, HHO) and condition (HHS, BF) effects in the AP direction. In this direction, SampEn was the only nonlinear parameter to show significantly lower values, by as much as 31.37%, when standing in heeled shoes than barefoot. Such a result suggests that when standing in heeled shoes, the system may not respond flexibly to a given destabilizing stimulus [18]. Wan et al. [37] and also Ko et al. [22] proved that the instability introduced by HHS is due not only to an increase in the height of the center of mass and a decrease in the area of the base of support but also to the fact that the feet are more supinated and plantarflexed. These changes in foot posture alter foot loading conditions [23; 34] and reduce the ankle's range of motion in plantar flexion and calcaneal eversion. As a result, the feet may be unable to evert naturally and effectively to maintain balance as heel height increases [11]. A common result may be a different balance strategy that uses different amounts of hip and ankle movement activity to maintain body balance. Such an implication may be confirmed by the results in the mediolateral direction. In this direction, the group effect was only for SampEn, where the HHE group obtained 40% higher values than HHO. This result means that HHE individuals feel comfortable standing in heeled shoes. A similar interpretation of the high entropy results was included in the study of Stins et al. [36]. Similarly, Schmit et al. [32] suggested that increased noisiness of postural movements among dancers indicated greater behavioral flexibility, allowing them to switch between behavioral modes more easily.

In the ML direction, significant condition effects (HHS, BF) appeared for FD and LyE. In both cases, HHS increased those nonlinear measures values by 6.71% and 8.22%, respectively. Higher values in this direction while standing in HHS suggest the ability to react faster to destabilizing stimuli and better balance control related to plasticity and adaptability to new conditions. This finding is confirmed by the change in foot loading. Our study showed that the load on both the forefoot and the backfoot did not differ between the group of women occasionally wearing high-heeled shoes and those wearing them frequently. It also did not change when the forefoot and backfoot loads were compared while standing barefoot and in heeled shoes. However, the value of the total force parameter provided more relevant information. As before, no differences were shown between the HHE and HHO groups, while a significant effect of the heeled shoe on the change in the value of this parameter was noted. In both groups, the total load on the left foot increased on average by 64.2%. The right, on the other hand, decreased twofold. It seems that in both groups, standing in heeled shoes was such a factor that involuntarily caused the transfer of body weight to the safe, supporting leg - in this case, the left one. Of course, such an implication needs to be verified by examining a group of people who have the left, rather than the right (as in this case), dominant lower limb.

In conclusion, wearing 11cm high-heeled shoes affected stability in young women negatively, independent of experience. The presence of HHS notably increased the CoP path length and velocity by 78%. Within the high-heel experienced group, the risk of falls increased by more than 44%. Additionally, the introduction of HHS resulted in a significant increase in FD and LyE values in the ML direction. Moreover, HHS contributed to a substantial increase in foot loading asymmetry, with a notable increase to 70% compared to the baseline of 30%.

Some limitations of this study need to be acknowledged. Firstly, measuring plantar pressure on the Zebris platform with shoes only allowed for the fore- and backfoot pressure distribution analysis. In-shoe pressure measurement systems would likely enable a more detailed analysis of pressure distribution. However, placing the measuring insole accurately within a high-heeled shoe can be challenging. Secondly, an analysis of the lower limb joint torques would provide additional information about joint loading and, consequently, the ability to maintain balance. This aspect should be considered in future studies. Thirdly, it would be valuable to incorporate dynamic stability analysis during balance tests such as the "limits of stability" test or during gait assessment. Information obtained from such evaluations would help analyze the risk of falls when wearing high-heeled shoes. Moreover, it is also worth mentioning that the test subjects from the HHE group are current or former dancers, which could potentially impact their balance. In addition, the type of footwear, specifically the high shoe upper, might have provided additional ankle stabilization and influenced the test results.

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