Kinetic and Kinematic Differences of Barefoot versus High-Heeled Gait in Healthy, Young Adult Females: A Pilot Study

By

Jennifer Fath, Shelby Jurek, Jessica Secoy

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Department of Physical Therapy
Angelo State University
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ABSTRACT

Background & Purpose: High-heeled shoes (HHSs) are a well known part of women's footwear repertoire that can influence ankle and foot position during functional activities. Few studies have quantified the impact of HHSs on static standing posture, balance, and gait, leaving numerical values as to why HHSs are unfavorable to be determined. The purpose of this study was to identify the differences in weight distribution, area of sway during gait initiation (GI), and gait parameters between barefoot and HHSs. **Methods:** 12 healthy young adult female participants ambulated barefoot, then while wearing HHSs in a motion analysis laboratory. Data was collected using a Tekscan Mat (Tekscan Inc., South Boston, MA, USA) and Vicon Motion Analysis system (Vicon, Centennial, CO, USA). Paired t-tests were utilized for statistical analysis. Statistical significance was set at $\alpha = 0.05$. **Results:** Weight shifting primarily occurs through the rearfoot in barefoot static standing, whereas in HHSs it occurs primarily through the forefoot. In addition, wearing HHSs anteriorly displaces the center of force (COF) in standing ($p < 0.05$). During barefoot GI, the area of COF movement was significantly greater ($p < 0.05$). Cadence and walking speed significantly decreased, and double-support phase of gait significantly increased in HHSs ($p < 0.05$). Vertical ground reaction forces at initial contact were significantly greater in HHSs ($p < 0.05$). **Conclusion:** With the results of this study, it appears that healthy young female adults demonstrate more caution with gait in HHSs as well as a disruption of natural postural alignment, imposing greater stresses on the lower extremities. While HHSs do not have a different motor strategy to execute the first step based on relative time of GI, it does require a compensated motor response to adjust to the altered postural alignment during both gait initiation and walking. This contributes to the theory that long-term use of HHSs could predispose women to lower extremity pathology through chronically altered biomechanics.

Key Words: high heel, barefoot, gait, lower extremity, female, healthy

INTRODUCTION

Historically, walking in high-heeled shoes (HHSs) has been viewed negatively as detrimental to the lower limb posture of women. It is a generally accepted concept that HHSs alter gait pattern and lower extremity biomechanics. Previous research has demonstrated a direct relationship between footwear and its impact on function.[1] Walking in HHSs has been aesthetically accepted for visual rather than functional purposes because they increase femininity of gait including reduced stride length and increased rotation and tilt of the hips.[2] However, recent research suggests that from 2002 to 2012, high-heeled related injuries have nearly doubled in the female population with age groups 20 to 29 (18.38 per 100,000 females) and 30 to 39 (11.07 per 100,000 females) demonstrating the highest rates of injury.[3] In Ebbeling et al. 1994, it was shown that heel heights greater than 5.08 cm predispose women to discomfort and foot positions at a biomechanical disadvantage.[4] This suggests that while HHSs may be considered attractive, they may be detrimental to one's health when compared to traditional gait.[5]

Previous studies suggest that HHSs alter the body's ability to balance with static standing, gait initiation, and functional mobility due to a change in the center of gravity superiorly and a forward weight shift.[6] Research has demonstrated that gait initiation from a toe-standing position results in a decreased posterior displacement of the center of pressure as well as an increase in forward momentum.[7] Lord and Bashford compared the ability of older women to balance under four footwear conditions including barefoot, walking shoes, high-heeled shoes, and their own shoes, noting that HHSs resulted in the worst balance performance as assessed by postural sway in static standing as well as maximal balance range and coordinated stability tests for dynamic balance.[8] Additionally, walking with HHSs is viewed as an unstable condition that may require different motor strategies for dynamic postural control to reduce the risk of falling.[9]

Thus, in order to successfully ambulate with HHSs, individuals typically alter their gait pattern in order to maintain balance and lessen the fear of falling. Clinically, basic gait parameters such as walking speed, cadence, and stride length can indicate one's ability to successfully maintain dynamic balance.[10] Prolonged double stance time is considered a compensatory strategy utilized in order to decrease the likelihood of falling due to an increase in time required for an individual to reestablish stability from one step to the next.[11] Consequently, this increase in double stance time is associated with a decrease in overall gait speed. As a result, individuals ambulating in high-heeled shoes typically demonstrate a more cautious gait pattern to minimize the risk of falling.

In physical therapy clinics, patients regularly attend therapy visits wearing athletic shoes as they are commonly associated with exercise. However, 37% to 69% of women are required to wear HHSs on a daily basis.[3] As clinicians, it is important to consider the implications that HHSs have on a female's ability to balance, ambulate, and maintain proper posture as we work to progress them towards their normal activities of daily living. Additionally, footwear is designed to help attenuate shock and assist the body's natural shock absorption capabilities during ambulation. However, HHSs are limited in their ability to absorb shock at initial contact due to their rigid structure and lack of cushioning capability and are often not considered in a patient's return to a normal routine.[12] With research suggesting a relationship between footwear and pathology, healthcare professionals, particularly physical therapists, must consider the implications that footwear have on the evaluation and treatment of lower extremity kinetic chain dysfunctions.[13] Due to the large number of females who wear HHSs on a daily basis, it becomes clinically significant to understand the altered gait biomechanics and gain quantifiable data regarding relevant and preventable musculoskeletal pathologies.

High-heeled shoes naturally position the ankle into plantarflexion and feature a narrow toe box, a rigid hindfoot, and increased longitudinal arch which alter an individual's balance, gait, and posture.[14] These positional changes may lead to musculoskeletal injuries such as muscle fatigue, osteoarthritis, and shortened calves.[15] Additionally, previous research suggests that the "altered anatomical position of the foot results in functional changes that include a shift in ground reaction forces toward the medial forefoot, a reduction in foot pronation during midstance, and an increase in the vertical ground reaction force at heel strike".[16] Kinematically, research has shown that "higher heels contribute to slower self-selected walking speeds, shorter stride lengths, and greater knee flexion, plantarflexion, trunk extension, and anterior pelvic tilt".[17] Kinetically, previous research demonstrates higher heels result in "greater peak vertical and anterior-posterior ground reaction forces throughout stance, medial forefoot pressures, and peak knee extensor moments, peak external knee adduction moments, and lower plantarflexion moments.".[17] While HHSs are considered to be stylish and at times necessary for the female population in the professional world, they create a cascade of events that alter movement patterns of the entire kinetic chain.

To our knowledge, the effects of HHSs on balance, gait initiation, and gait pattern has not been sufficiently presented in previous research. It is important to consider that with altered motor strategies and different walking patterns, individuals may be more susceptible to an abnormal plantar pressure distribution as well as altered joint mechanics.[18] Therefore, the purpose of this study was to identify the differences in weight distribution, area of sway during gait initiation, and gait parameters between barefoot and HHSs in order to provide numerical evidence of HHSs interfering with normal female gait and pre-gait activities.

It was hypothesized that walking in HHSs would decrease walking speed, decrease step length, decrease cadence, and increase double-support phase as compared to barefoot walking. It was also hypothesized that HHSs would demonstrate a greater weight shift onto the forefoot during static standing compared to barefoot. During the gait cycle, it was hypothesized HHSs would exhibit a greater vertical peak ground reaction force during initial contact of gait, subsequently leading to impaired shock dissipation through the loading response phase versus that of barefoot walking.

METHODS

Participants

A total of 12 healthy, young female adults were recruited from San Angelo, Texas. Inclusion criteria were as follows: must be female, age 18-40, experienced in ambulating in HHSs, and shoe size 6 US to 8 US. Exclusion criteria were defined as any acute or chronic lower extremity injury or pathology, any limb deformities or abnormalities, or pain during ambulation. Demographic characteristics that were collected included height, weight, leg length, pelvic width, knee width, and ankle width. See Appendix A for demographic profile of participants. This study was approved by the Institutional Review Board at Angelo State University in San Angelo, Texas. Health Insurance Portability and Accountability Act (HIPPA) guidelines were adhered to throughout the course of this study, and participant privacy, consent, and full disclosure were maintained.

Research Design

The present study utilized a within subjects design on a healthy, young female adult population.

Materials & Equipment

Gait assessment was standardized according to the protocol developed by the Nordic Vicon User Group (Nordic Vicon User Group, 2013), which is based on recommendations for reflective marker placement set-up for the Plug-in gait (PiG) model (Vicon Motion Systems, 2012). For each data collection session, six infrared and two digital cameras were calibrated with a five marker wand L-frame. Prior to each session, anthropometric data was collected, and included the following: weight (mechanical scale), height (measuring tape), bilateral leg length (anterior superior iliac spine to ipsilateral medial malleolus with knee extended; measuring tape), pelvic width (distance between anterior superior iliac spines; measuring tape), knee width (distance between lateral and medial femoral condyles; dial caliper), and ankle width (distance between lateral and medial malleoli; dial caliper). These data were exported to the Vicon Nexus 1.7.1 software program (Vicon Motion Systems, Oxford, UK) for the purpose of calibration of each participant.

After individual measurements were taken, sixteen reflective markers (each of 14 mm diameter) (Vicon Motion Systems, Oxford, UK) were affixed to anatomical landmarks using double-sided adhesive tape according to the PiG model (Vicon Motion Systems, 2012) and guidelines from the Nordic Vicon User Group (2013). The reflective gait marker landmarks were: anterior superior iliac spines, posterior superior iliac spines, right lateral thigh, left lateral thigh, right knee, left knee, right ankle, left ankle, right toe, left toe, right heel, and left heel. See Appendix B for specific description and visual representation of marker placement. Participants wore form-fitting clothing, and were instructed to tuck loose shirts into pants prior to reflective marker placement.

Participants initiated each trial from a Tekscan High Resolution Floor Mat System (Tekscan Inc., South Boston, MA, USA), which was utilized to record ground reaction forces, location of weight shift, and the time required for these tasks, as well as the center of force (COF) displacement. The active sensing area of the Tekscan Mat was 48.8 cm x 47.7 cm.

Standardized HHSs were utilized in this study and were provided to each of the participants by assessors. All HHSs were 7 cm high.

Procedures

Each participant was instructed to stand barefoot in the center of the data collection room in order to calibrate the eight Vicon cameras specific to each individual's anatomy and posture in static standing. Then, prior to the task of gait initiation (GI), participants were instructed to stand statically on the Tekscan Mat (Tekscan Inc., South Boston, MA, USA), where weight distribution and center of force (COF) were recorded for each trial. Participants were then instructed to walk at a self-selected, comfortable pace along a 10-meter walkway, making sure to contact each of two AMTI force platforms (AMTI, Watertown, MA, USA) concurrently during each trial. Force platforms were located approximately at the halfway point of the walkway. Each participant was instructed to maintain consistent gait velocity approximately three feet beyond the force plates to a pre-positioned cone to avoid confounding data collection results with gait velocity deceleration. Participants were given a verbal cue for initiating each trial. Prior to each data collection session, participants were permitted a single test trial to become familiar with the position of the force platforms and length of the walkway. Ten trials in total were recorded for each participant, five trials under barefoot conditions and five trials while wearing HHSs.

Balance was evaluated utilizing the area of sway (cm^2) from Tekscan data to determine postural stability during GI. The absolute temporal events were obtained from the vertical GRFs measured by the Tekscan Mat corresponding to the swing and the stance limb during GI.[19, 20] In order to normalize the temporal parameters of GI, according to each subject, the time from onset of movement to stance limb toe-off was considered at 100% of the GI cycle. The temporal events measured were: (1) time to swing limb toe-off (anticipatory postural adjustment [APA]), (2) DST_{GI} (duration between swing limb heel strike and stance limb toe-off [SLToff]), and (3) time to SLToff.[21, 22]

Vicon motion analysis was utilized to assess gait kinematics under both barefoot and HHSs conditions of walking. Specific values that were recorded included: (1) cadence (2) double support time, (3) step length, (4) step time, (5) step width, (6) stride length, (7) stride time, (8) walking speed, (9) shock dissipation, and (10) peak vertical GRF. Shock dissipation and peak vertical GRF were assessed using the force platforms, whereas all other gait characteristics were recorded using the Vicon system.

For the purposes of this study, shock dissipation was defined as the moment of shock absorption that occurs immediately following the impact phase of gait as the stance limb transitions from initial contact to loading response.[23] Center of force (COF) is referred to by Tekscan Inc. as "the center of all of the forces on the sensor", which shows how plantar pressures are balanced on the sensor as an individual stands upon it. Optimal location of COF, according to Tekscan Inc., is "at the intersection of the midline of the feet and at the level of the midtarsal joint". Area of sway, or COF movement according to Tekscan Inc., is "the amount of COF movement in standing", which is optimally minimal and characterized by symmetrical oscillations in the COF as it moves (Tekscan Inc., South Boston, MA, USA).

Statistical Analysis

Statistical analysis was performed using SPSS version 21 (IBM, Chicago, IL). Paired t-tests were used to identify the significant differences between barefoot and HHSs for analyzing Tekscan and Vicon data. Statistical significance was set at $\alpha = 0.05$. For statistical analyses, the means of three trials were calculated out of a total five trials for each condition and for each subject.

RESULTS

Paired t-tests revealed that there is a significant effect under barefoot circumstances on weight bearing onto the rearfoot (64%) compared to the forefoot (34%) ($p=0.011$). HHSs also showed a significant weight distribution increase of the forefoot (78%) versus the rearfoot (22%) ($p=0.000$) during static standing. Accordingly, center of force (COF) location with HHSs showed a forward shift of 3.4 cm greater than the COF location in barefoot standing. With respect to motor strategies of dynamic postural control, barefoot (179cm^3) has about 1.9 times greater area of COF movement than HHSs (92cm^3) during GI ($p=0.000$). However, no significant differences were found for any relative temporal parameters of GI ($p > 0.05$).

Additional paired t-tests of Vicon data demonstrated significant differences between barefoot cadence (115.92 steps/min) compared to HHSs (112.38 steps/min, $p = 0.008$). A significant difference was found to exist between barefoot walking speed (1.20 m/s) and HHSs (1.16 m/s, $p = 0.023$). A significant difference was also found between barefoot double stance time (0.16 secs) and HHSs double stance time (0.23 secs, $p = 0.001$). No significant differences were found between single support stance time, step length, step time, step width, stride length, and stride time when comparing barefoot and HHS conditions ($p > 0.05$). See table 1 for full results regarding gait parameters.

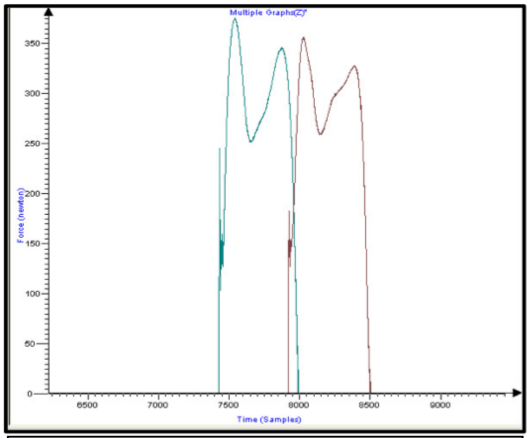


Figure 1: Single subject vertical ground reaction forces generated under barefoot conditions. Initial spike: force plate registration of heel strike. First Peak: body weight and force as accelerating mass. Downward deflection: force drops as body weight moved across force plate. Final Peak: body weight and force as accelerating mass.[24]

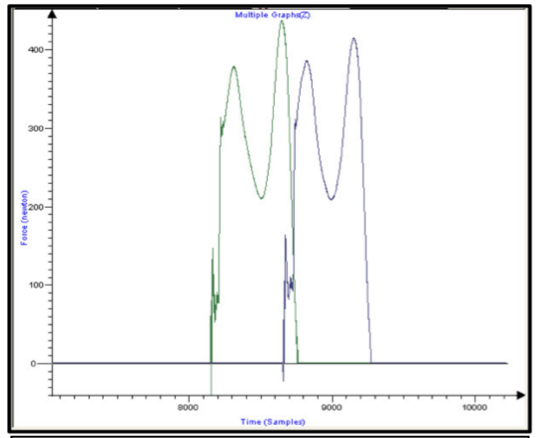


Figure 2: Single subject vertical ground reaction forces generated under high-heeled shoes (HHSs) conditions. Initial spike: force plate registration of heel strike. First Peak: body weight and force as accelerating mass. Downward deflection: force drops as body weight moved across force plate. Final Peak: body weight and force as accelerating mass.[24]

Vertical peak ground reaction forces (GRFs) at initial impact demonstrated a statistically significant difference when comparing both the right and left foot of the barefoot and HHSs conditions. The R foot peak GRF of barefoot measured at 317.13 (N/s) compared to HHSs of 335.38 (N/s) ($p = 0.024$). The L foot peak GRF of the barefoot condition was 306.52 (N/s) compared to HHSs of 321.29 (N/s) ($p = 0.018$). While discrepancies were noted between both R and L shock dissipation when comparing the two conditions, no statistically significant differences were found. Figure 1 and Figure 2 represent barefoot and HHSs vertical GRFs as plotted by Vicon software. See table 2 for full results regarding force plate measurements.

DISCUSSION

This study showed that gait parameters including gait speed, cadence, and double stance time were altered in HHSs when compared to the barefoot conditions. This finding indicates that individuals wearing HHSs may demonstrate a more cautious gait pattern as a compensatory mechanism to achieve greater stability in order to attain steady-state walking. Additionally, weight distribution in HHSs in standing is more significant in the forefoot compared to barefoot due to the excessive plantarflexion at the talocrural joint that HHSs naturally place the ankle in. The significant differences that existed between vertical peak ground reactions forces at initial heel strike signify the altered shock absorption mechanism of the lower extremity during HHSs gait.

During a normal gait cycle, heel strike occurs with the foot in a supinated position. As the foot continues through the loading response phase, shock absorption occurs as the subtalar joint pronates to decrease the impact upon loading.[25] During efficient ambulation, three critical events occur in which energy is consumed. These include: controlled forward movement during terminal swing, shock absorption at initial contact, and propulsion during pre-swing.[26] From the period of initial contact to loading response, the primary shock absorbers during this weight-loading period include the quadriceps and dorsiflexors, particularly the tibialis anterior.[16] During loading response, the entire stance limb should contribute to shock absorption in order to decelerate the limb as the foot transitions into a weight-bearing position.[27] The four primary shock absorbers in gait include: 1) ankle plantarflexion, 2) subtalar pronation, 3) knee flexion, and 4) contralateral pelvic drop. In normal gait, knee flexion and eccentric control from the tibialis anterior provide the initial mechanisms of shock absorption. Additionally, during initial contact, subtalar pronation creates a mechanism in which the foot transitions from highly rigid to mobile in order to provide shock absorption and and weight acceptance.[28]

As the foot progresses through the loading response phase, the body utilizes the ankle rocker as permitted by movement at the talocrural joint to transmit body weight over the entire lower extremity.[27] The medial longitudinal arch serves as an additional shock absorber that can respond and adapt to various ground reaction forces.[29] As the individual continues through the stance phase, one must transition from ankle plantarflexion to ankle dorsiflexion from midstance to the pre-swing phase. This results in the locking of the subtalar and transverse tarsal joints to transmit force across the entire foot for propulsion into the pre-swing phase of gait.[26]

Alterations to these joint motions or decreased muscle activation may subsequently lead to decreased shock absorption capabilities resulting in potential for musculoskeletal injuries.[4] Prior studies have determined that lower body movement coordination during HHSs gait, even as far up the kinetic chain as trunk and hip extensors and cervical paraspinal activation, is altered greatly.[30,31] The findings of this study support the notion that foot positioning with HHSs may predispose women to injury due to the inability of the foot to transition from rigid to mobile at initial contact. While differences in shock dissipation were not significant, it is worth speculating that the excessive plantarflexion of the foot in HHSs may place the tibialis anterior, which is essential for eccentric control during stance phase, on stretch. This would not be advantageous for proper muscle activation during this task of gait, and could contribute to altered motor strategies throughout the gait cycle.

In HHSs, a person's center of mass (COM) shifts forward resulting in an increased anterior pelvic tilt and lumbar lordosis and greater difficulty maintaining balance due to the weight distribution on the forefoot.[32] Due to the rigid posture of the ankle in plantarflexion and supination, the primary use of ankle-strategy to correct balance perturbations is eliminated.[33] Therefore, women in HHSs are forced to resort to various other strategies to maintain their COM during the controlled weight distribution in single limb stance.[33,34] In order to compensate for the increased difficulty of maintaining balance during single limb support, it is likely that the greater double support stance time is a result of the person's less stable posture while ambulating in HHSs when compared to barefoot walking. The combination of decreased balance, increased double limb support, and the implementation of various balance strategies contribute to a general decrease in gait speed while walking in HHSs. Chien et al. supported that the theory of decreased gait speed walking in HHSs can be attributed to the increased balance demand associated with a narrowed base of support.[33] Although step width in HHSs was not significantly different, decreased area of sway was demonstrated in this study by a significant decrease in area of COF movement during gait initiation. Normally, anticipatory postural adjustments regulated by the central nervous system occur prior to voluntary movement, which influence both mediolateral and anteroposterior directional forces in order to propel the swing limb while maintaining the center of mass within the area of sway.[35] It is worth noting that proprioceptive input has a role in anticipatory postural adjustments, and that by increasing muscle spindle input through plantarflexed positions of the feet in HHSs, proprioception may be altered, thereby influencing the motor strategies necessary to weight shift during gait initiation.[36,37]

From a healthcare perspective, the selective functional movement assessment (SFMA) utilizes a joint-by-joint approach that suggests that as one moves up the kinetic chain, we alternate between the need for mobility and stability. The talocrural and hip joints are considered to be highly mobile whereas the tibiofemoral joint is viewed to be stable. When the mobile joint (talocrural) becomes immobile, as is the case with high-heeled shoes, the stable joint (tibiofemoral) may be forced into greater mobility as a result of compensation.[38] Previous research has proposed that the differences in kinetics and kinematics between barefoot and HHSs gait can alter joint loading of the lower extremity, resulting in an increased susceptibility to overuse, chronic conditions.[39] With a decrease in subtalar pronation and a talocrural joint pre-positioned in plantarflexion, it becomes more difficult to dissipate shock at initial contact through the loading response.[15] This is consistent with the findings of this study that showed vertical peak ground reaction forces to be significantly increased in HHSs and that differences did exist in shock dissipation throughout the stance phase between barefoot and HHSs conditions. Consequently, the stable talocrural joint forces the knee into a more mobile role as increased knee flexion becomes a more prominent shock absorber in HHSs. Previous research has established that walking in HHSs produces greater torque at the knee joint in the sagittal plane, while also producing heightened rectus femoris activity.[12,40] These ideas contribute to implications that women are more predisposed to knee osteoarthritis as well as overall knee instability. As women are the sole wearers of HHSs, it is possible that prolonged wear of HHSs may further predispose women to musculoskeletal pathologies clinically encountered by physical therapists.

Limitations

The primary limitation of this current study revolves around the small, non-probability sample of convenience that was utilized to recruit participants. Due to the small sample size, the probability of a Type II error increases due to the ability of one outlier to skew the data in a positive or negative manner. In addition, shoe sizes were limited to 6 US - 8 US and thus, most participants were of shorter stature, making it difficult to generalize these results to a heterogeneous population of healthy females. This study solely focused on the gait kinetics and kinematics of a healthy female population and should incorporate those with musculoskeletal pathologies for better comparison of biomechanical differences with gait.

In order to compensate for the limitations in this study, the gold standard of human movement analysis (Vicon) as well as a comparable system to the gold standard force plate (High Resolution Tekscan Mat) were employed. The Vicon system is a three-dimensional motion analysis system that has been proven to deliver accurate and reliable biomechanical analysis of human gait.[41] The HR Tekscan Mat is a pressure sensing mat that can be used for postural stability and sway analysis as well as foot distribution data and has proven to be reliable in detecting plantar pressures.[42] Together, these tools offer greater statistical power when compared to human observation in clinical gait analysis.

Further research should be implemented in order to better understand the acute versus chronic effects of high-heeled gait. The addition of a female sample population with musculoskeletal pathologies such as knee osteoarthritis and Haglund's deformity can offer better comparison of gait abnormalities that exist between barefoot and high-heel ambulation. Subsequently, the effects of fatigue with high-heel gait can be researched by utilizing a treadmill over specified time intervals to look at the potential chronic effects with the use of HHSs.

CONCLUSION

HHSs significantly dampen the biomechanical advantages of normal walking, leading to overall impaired characteristics of gait and stability in young healthy females. By shifting the COF forward and decreasing the area of sway, a female in HHSs must compensate throughout the gait cycle to maintain balance by decreasing cadence, decreasing velocity, and increasing double-support phase. Consequently, this study showed that greater force is generated upon initial contact when ambulating with HHSs. These results provide a basis for further research to investigate not only musculoskeletal pathologies associated with HHSs but also to establish a clinical approach for identifying, treating, and educating regarding functional mobility affected by HHSs.

Table 1. Gait Parameters Barefoot vs. High-Heeled Shoes

Gait Parameter	Barefoot	High-Heeled Shoes	p-value
*Cadence (steps/min)	115.92±5.82	112.38±5.01	0.008
*Double Support (s)	0.1554±0.02	0.2338±0.06	0.001
Single Support (s)	0.44±0.02	0.41±0.07	0.086
Step Length (m)	0.62±0.04	0.63±0.03	0.731
Step Time (s)	0.52±0.03	0.55±0.05	0.130
Step Width (m)	0.13±0.02	0.13±0.02	0.392
Stride Length (m)	1.25±0.08	1.25±0.04	0.825
Stride Time (s)	1.04±0.05	1.15±0.26	0.163
*Walking Speed (m/s)	1.20±0.08	1.17±0.07	0.023

* Statistically significant differences between barefoot and high-heeled shoes

Table 2. Force Plate Measurements Barefoot vs. High-Heeled Shoes

Force Plate Measurement	Barefoot	High-Heeled Shoes	p-value
R Foot Shock Dissipation (N/s)	655.61±284.11	725.96±258.89	0.259
L Foot Shock Dissipation (N/s)	587.11±270.48	619.06±183.81	0.666
*R Foot Peak Force (N)	317.13±55.88	335.38±47.41	0.024
*L Foot Peak Force (N)	306.52±55.51	321.29±48.27	0.018

* Statistically significant differences between barefoot and high-heeled shoes

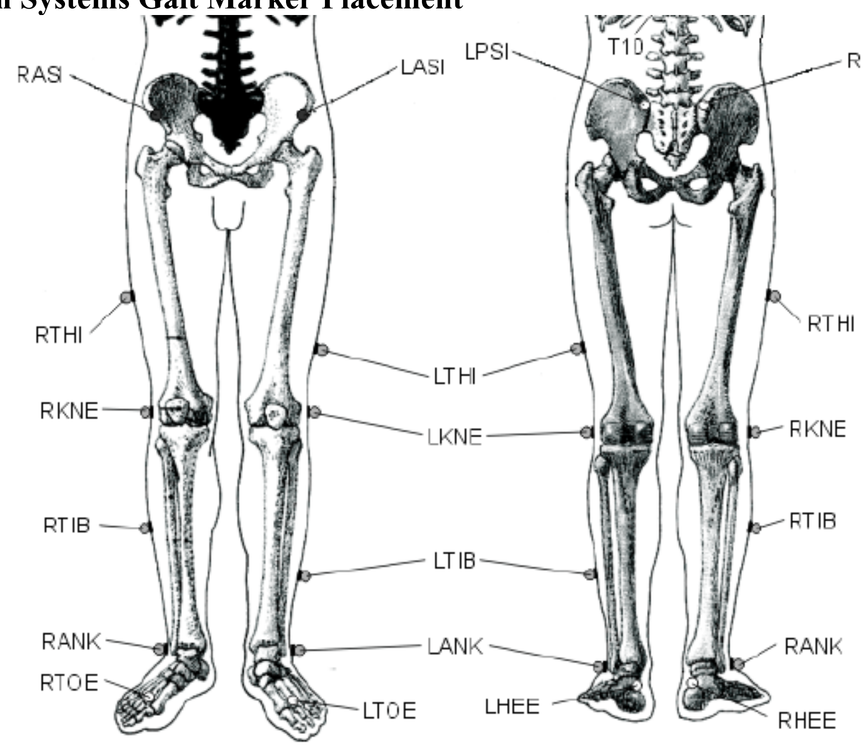
APPENDIX A

Demographic Characteristics of Study Population	
Age (years)	27.67 +/- 4.98
Height (mm)	1.62 +/- 0.06
Weight (kg)	63.91 +/- 9.28
BMI (kg/m2)	24.58 +/- 3.81
Knee Width Left (mm)	95.50 +/- 8.48
Knee Width Right (mm)	96.33 +/- 9.04
Ankle Width Left (mm)	59.67 +/- 3.17
Ankle Width Right (mm)	59.58 +/- 3.01
Leg Length Left (mm)	834.25 +/- 47.68
Leg Length Right (mm)	830.42 +/- 45.79
*N=12 Female College Student	

APPENDIX B

Vicon Motion Systems Gait Marker Placement[43]	
Marker	**Placement**
ASIS	Directly over the anterior superior iliac spine
PSIS	Directly over the posterior superior iliac spine
Knee	On the lateral epicondyle of the knee
Thigh	Over the lower and lateral third of the surface of the thigh, just below the swing of the hand and in the plane of the hip and knee joint centers
Tibia	Over the lower third of the shank and in the plane of the knee and ankle joint centers
Ankle	On the lateral malleolus along the line that passes through the transmalleolar axis
Toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot
Heel	Placed on the calcaneous at the same height above the plantar surface of the foot as the toe marker

Vicon Motion Systems Gait Marker Placement[43]



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