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Short-Term Effects of Increased Body Mass and Distribution on Plantar Shear, Postural Control, and Gait Kinetics: Implications for Obesity

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ABSTRACT

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Context: Obesity is a growing global health concern. The increased body mass and altered mass distribution associated with obesity may be related to increases in plantar shear that may lead to physical functional deficits. Measurement of plantar shear may provide unique insights on the effects of body mass and body distribution on physical function or performance.

Purpose: 1) To investigate the effects of body mass and distribution on plantar shear; 2) To examine how altered plantar shear influences postural control and gait kinetics.

Hypothesis: 1) a weighted vest forward distributed (FV) would shift the center of pressure (CoP) location forward during standing compared with a weighted vest evenly distributed (EV); 2) FV would increase plantar shear spreading forces more than EV during standing; 3) FV would increase postural sway during standing while EV would not; and 4) FV would increase peak braking force, plantar impulses, and plantar shear spreading forces during walking more than EV.

Methods: Twenty healthy young males participated in four different tests: 1) static test (for measuring plantar shear and CoP location without acceleration; 2) two-leg standing postural control test; 3) one-leg standing postural test; and 4) walking test. All tests were executed in three different weight conditions: 1) NV; 2) EV with 20% added body mass; and 3) FV, also with 20% added body mass. We measured plantar shear stresses using a pressure/shear device and extracted several shear and postural control metrics. Repeated measures ANOVAs with Holms post hoc test were used to compare each metric among the three conditions ($\alpha = 0.05$).

Results: FV and EV increased both anterior-posterior and medial-lateral plantar shear forces in single-foot trials compared to NV. FV shifted CoP forward. FV and EV showed decreased CoP range and velocity and increased time-to-boundary (TTB) during postural control compared to NV. While EV increased medial-lateral plantar shear spreading force, FV increased anterior-posterior plantar shear spreading force during walking.

Conclusion: Added body mass increases plantar shear spreading forces. Body mass distribution had greater effects during dynamic tasks. In addition, healthy young individuals seem to quickly adapt to external stimuli to control postural stability. However, the interactive effects between body mass and distribution may disrupt physical function and/or performance in other populations such as obese. Plantar shear may play a critical role in clinical diagnosis.

Keywords: evenly distributed weight vest, front-loaded weight vest, plantar shear spreading force

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Introduction

Obesity is a growing global health concern with numerous biomechanical complications that affect quality of life [1,2,3]. Approximately 39% of the world's population is either overweight or obese [4]. In 2017 alone, the U.S. federal government invested $300 billion in addressing obesity [5]. One major health concern in this population is fall risk, with obese individuals reporting a fall history twice as frequent as nonobese individuals [2]. Falling is closely associated with postural instability, which has a strong biomechanical component [6]. For example, one study found that body mass accounted for over 50% of the variance of center of pressure (CoP) speed—a major indicator of postural control status—while age, body height, and foot size had no statistical contribution [7]. In addition to postural instability, obese individuals have demonstrated other related biomechanical changes including altered gait mechanics [8], foot deformities [9], increased muscle activity of the lower extremities [10], and decreased plantar cutaneous mechanoreceptor sensitivity [11,12]. Although it is generally accepted that obesity is closely associated with these complications [7,13], the biomechanical mechanisms causing them are still unclear. Studying the effects of added body weight on the interaction between the foot and the ground may help elucidate some of these mechanisms.

Added body weight by itself affects plantar loading. Because a heavy body mass inherently causes greater pressure under the plantar surface of the foot during weight-bearing activity [14,15,16], it is suggested that the increased plantar pressure may contribute to a decrease in plantar mechanoreceptor sensitivity [7], which is indispensable to postural control [17,18]. According to Weber's law, it is harder for those who have a heavy body mass to distinguish changes in external stimuli [19]. In fact, obese individuals demonstrated an increased threshold of plantar sensitivity to tactile stimuli during standing while concurrently exhibiting decreased static postural stability [12]. However, body mass by itself does not fully explain increased postural sway. For instance, an evenly distributed weighted vest appears to support upright posture [20], while an unevenly distributed weighted load (i.e., backpack) substantially aggravates postural control [21]. In addition, an unevenly distributed heavy load increased postural sway while a relatively light equally distributed load did not cause postural sway [20]. It appears as if postural instability occurs by interactive effects between body mass and body distribution [22]. Because plantar pressure is increased by an added load regardless of the way the load is carried [23], there are likely other factors to postural control response to heavy weight.

Obesity also alters body mass distribution. Obesity tends to build excessive adipose tissue in the abdomen that changes body configuration to a so-called pear shaped body. The changed body distribution theoretically shifts the location of the the center of mass (CoM) forward, increasing the anterior-posterior (AP) distance between the CoM and the ankle joint that eventually increases gravitational torque with an elongated joint moment arm. This destabilizing torque due to gravity makes it more difficult to generate corrective torque or reduce postural sway [24]. In addition, obese individuals must generate relatively greater ankle joint torque to compensate for the increased gravitational torque. This increased muscle activity is suggested to facilitate lower extremity muscle fatigue that leads to postural control deficits [10]. Yet, muscle strength appears to play only a minor role in postural control [25], suggesting that muscle weakness does not fully explain postural instability in the obese population. Although previous studies have looked at effects of mass distribution on postural control, they have primarily observed loaded applications where the added mass is on the back (e.g., military backpacks) [20,21]. Little has been done to investigate the mechanical effects of obesity by adding weight to the front of the body.

In addition to static physical performance, body mass by itself seems to affect dynamic physical performance. Obese individuals and healthy individuals with a heavy backpack have similarly demonstrated abnormal gait mechanics—narrower stride length, wider width between feet, increased double support, and slow speed [8,26]. These altered spatiotemporal parameters may be a protective mechanism to avoid falling [27]. In addition, body mass directly affects walking efficiency [28,29]. However, in obese individuals, the increased energy amount is not as high as would be expected due solely to mass alone [28]. suggesting an improved recovery of mechanical energy that is not quite clear [30].

Investigating the effects of added body mass and mass distribution on plantar shear forces may increase our understanding of the mechanical effects of obesity and provide insights into postural control and movement energetics. It is reasonable that elevated body mass alters not only vertical pressure, but also horizontal shear stresses. Most prior research has focused on the greater vertical plantar pressure that results from a heavier body mass. However, our feet are impacted by both vertical and horizontal components of ground reaction force (GRF) [31]. As body mass increases, the foot is loaded to a greater extent, resulting in horizontal plantar tissue spreading as demonstrated by increased foot contact area in obese individuals [15]. In addition, obese individuals have shown greater plantar pressure in forefoot region,[8] which implies that forward-shifted body CoM may move CoP forward with greater plantar spreading forces. Despite the potential contribution, little is known about plantar shear forces and how they are affected by body mass and distribution due to measurement difficulties. However, newer technology allows for direct measurement of shear stresses during dynamic situations, and could be utilized to study the response to added body mass.

Due to the numerous confounding factors associated with obesity, we sought to isolate the biomechanical effects of added body mass and distribution (i.e., even distribution vs. forward distribution) on plantar shear stresses. Therefore, the purpose of this study was to investigate the effects of added body mass and distribution on plantar shear stresses in nonobese individuals, both in standing and walking. We also sought to preliminarily assess their effects on postural stability. We hypothesized that 1) a weighted vest forward distributed (FV) would shift the CoP location forward during standing compared with a weighted vest evenly distributed (EV); 2) FV would increase plantar shear spreading forces more than EV during standing; 3) FV would increase postural sway during standing while EV would not; and 4) FV would increase peak braking force, plantar impulses, and plantar shear spreading forces during walking more than EV.

Methods

Participants

A sample of 20 healthy male participants between ages of 18 and 40 were recruited (age = 23 ± 3.10 years; height = 180 ± 0.05 cm; mass = 75 ± 8.02 kg; BMI = 23 ± 1.85 kg/m^2). In consideration of the distinct body fat accumulation tendency between male and female, we recruited only male participants to unify the body mass distribution. The number of participants was determined from a previous study [23]. Age criteria was based on previous research showing no differences in muscle strength[32] or spatiotemporal gait factors in this age group [33]. Inclusion criteria consisted of normal body weight—BMI: 20–25% defined by World Health Organization (WHO) [34].

The participants were screened and excluded if they were out of range of the normal BMI level or had issues with the lower limb musculoskeletal system affecting mobility or balance such as symptomatic osteoarthritis, any known heart disease, any neurological conditions such as diabetes, or lower and upper extremity injuries in the past six months. Screening was done using the Lower Extremity Functional Scale (LEFS) [35]. Participants also had to be able to stand and walk with an added load unassisted. All participants signed an IRB-approved informed consent form before involvement in the research.

Experimental Procedures

Participants performed four tests in the following order: 1) static stance, 2) double-leg postural control, 3) single-leg postural control, and 4) walking. Three different weight conditions were used in each test: Unweighted (NV), evenly distributed weighted vest (EV), and front-loaded weighted vest (FV). The order of the weight conditions was randomized within each test. For the weighted vest conditions, 20% of the participant's body mass was added, in the form of individuals 0.45 kg sand bags. The 20% added weight increased mean BMI to 27.80 (± 2.22 kg/m^2). According to WHO's obesity definition, this would classify the participants with the added weight in the range of overweight. For EV, weights were added to the front and back of an exercise vest (Valeo VA4471, Houston, TX, USA), while for FV, the same amount of weight was added to the front of the body using a baby carrier (Sunveno sv22094, New Delhi, India) (Figures 1 & 2). Participants were given several minutes to acclimate to the vest conditions prior to testing. In addition, to minimize muscle fatigue, participants were provided with a 1-minute seated rest period between each trial and a 3-minute seated rest period between each test.

A pressure and shear measurement device (FootSTEPS, Innovative Scientific Solutions, Inc., Dayton, OH, USA) was used to collect all plantar force data. Details regarding the device hardware and measurement validity have been published previously [36]. Briefly, the device consists of a glass plate with an embedded stress sensitive polymer film. A camera underneath the plate captures film displacements which are converted to vertical pressure and mediolateral and anteroposterior shear stress distributions using a finite element analysis model. A force plate (AMTI, Watertown, MA, USA) mounted underneath the device is used for calibration. The camera measures 0.42 by 0.28 meter area and is limited to a 50 Hz sampling rate for short duration activities (e.g., walking) and 25 Hz for longer duration collections (e.g., standing). Adjustable height staging panels (StageRight Z-HD, Clare, MI, USA) were used to make a walkway that was flush with the sensing surface. A hole was cut in the center panel for the sensor and a small (< 1 cm) gap was maintained around the device perimeter (Figure 3).

The static stance test was used to evaluate the influence of added mass on plantar forces, without the influences of the small accelerations that can accompany balance tests. Thus, this test consisted of an assisted standing posture. The participant stood barefoot in front of a pressure/shear measuring plate. Once given a starting sign from a researcher, the participant took a step onto the plate. The participant was instructed to initially step his dominant foot on the force plate and then to sequentially settle the opposite foot next to the dominant foot. Assistance was provided in the form of a fixed bar placed in front of the force plate. After putting feet together on the force plate, the participant was instructed to lightly touch the bar only with his fingertips, but not grasp and push on it (Figure 1). To standardize posture, the participant was instructed to watch a screen in front of the participant that showed the participant's face in profile superimposed on a grid. The grid was provided by a motion capture system (Qualisys, Inc., Göteborg, Sweden) linked to the screen. Four reflective markers were used to enable the participant to maintain an upright standing position. These markers were attached to the lateral shoulder, greater trochanter, femoral epicondyle, and malleolus on the left side, forming a line. The participant was asked to maintain alignment of these four markers on a line of the grid in order to maintain consistent upright position during the test. Three trials of 25 seconds each were recorded at 25 Hz. To avoid extraneous movements at the beginning and end of a trial, we removed the first 7 and last 5 seconds, thus, the actual recording period was 13 seconds.

For both postural control tests, the participant took one step onto the force plates once a researcher gave a starting sign, placed hands on iliac crests, and stood as still as possible on the pressure/shear plate. The bar and grid were not used (Figure 2). But the participant was instructed to visually focus on a marked area at approximately eye level. For the two-foot tests, the participant was instructed to place both feet so that they were touching, in as narrow a stance as possible. In the single-leg standing, a participant stood on his dominant foot, bending his opposite knee at 45º (Figure 2). Three successful trials of 25 seconds each were recorded for each test at 25 Hz. Trials were discarded if hands were taken off the hip or the toes of the opposite leg touched the ground (single-leg trials) during the recording. Again, the first 7 and last 5 seconds were also removed prior to processing.

In the walking test, preferred walking speed was first determined in order to minimize confounding factors, considering that walking speed by itself affects gait mechanics [37]. While a participant walked barefoot back and forth on the walkway (18 feet long and 3 feet width, (Figure 3), average walking speed was measured using laser timers placed near each end of the middle block of the walkway. The same laser timers were used to monitor walking speed during the actual test. The participant walked at his preferred speed across the walkway containing the pressure/shear plate in the middle of it (Figure 3). The participant was instructed to walk as normal as possible and to keep looking straight ahead. The starting position was adjusted by a researcher to ensure full contact of the dominant foot on the force plate. Moreover, the participant received verbal feedback after a walking trial in order to maintain the walking speed within ± 10% of his average walking speed. Three successful trials (i.e., clean foot contacts at average walking speed) were collected at 50 Hz.

Data Analysis

Data from the pressure/shear plate was first processed in manufacturer-supplied software, as described by Goss et al. [36]. Raw shear stress and reconstructed pressure data were then imported into custom LabView software (National Instruments, Austin, TX, USA) for data analysis.

For static stance, mean AP and medial-lateral (ML) plantar shear spreading forces and AP CoP location were extracted. To determine plantar shear spreading forces, all directional shear stresses were summed and multiplied by the area to create directional shear GRFs. Specifically, anterior-directed shear stresses were separated from posterior-directed shear stresses; likewise, medial and lateral shear stresses were separated from each other. AP and ML spreading was then determined as the amount of each directional force that opposed the main directional force. For example, if anterior force exceeded posterior force, the absolute value of the posterior force was used to represent plantar spreading. CoP was calculated as the weighted average of all pressure pixel locations (x_i), weighted by vertical force (F_{vi}) (Equation 1). This was expressed in the AP direction as a percentage of foot length, relative to the heel.

$$CoP_{AP} = \frac{\sum(F_{vi} * x_i)}{\sum F_{vi}} \qquad [1]$$

To do this, a composite plantar pressure image was constructed (max value of each pixel across time) and the maximum anterior and posterior boundaries of the footprint were identified using pressure thresholds.

For the postural control tests, the following dependent variables were extracted: AP CoP location, mean CoP speed, AP and ML CoP range, and AP and ML time-to-boundary (TTB).

These measures were chosen to represent a broad range of previously published metrics from the postural control literature. CoP location was calculated in the same way as it was in static stance. Mean CoP speed was obtained by summing instantaneous CoP displacements (i.e., path length) and dividing by total trial duration (13 seconds). CoP range was represented as the difference between maximum and minimum CoP values in both AP and ML directions. TTB was calculated as previously described for both AP and ML directions [38]. Briefly, rectangular AP and ML boundaries were identified from a composite pressure image. The distance between the boundary and the instantaneous CoP location were divided by instantaneous velocity (central difference) in that direction. For example, if CoP was moving toward the metatarsal heads or anterior boundary, the distance between the anterior boundary and the CoP was divided by the instantaneous velocities which corresponded to the anterior direction. AP and ML TTB(s) were obtained separately. From the TTB time series, we calculated the mean of all local signal minima.

For the walking test, the following dependent variables were extracted: peak braking (posterior) force, braking impulse, propulsive (anterior) impulse, and AP and ML plantar shear spreading at midstance. Peak braking force was obtained as the maximum net posterior force. Braking and propulsive impulses were also obtained from the net AP forces using trapezoidal integration. For braking impulse, the area from heel strike to midstance was obtained, while for propulsive impulse, the area from midstance to toe-off was obtained. AP and ML plantar shear-spreading forces were calculated in the same way as was done for the static test, but extracted at midstance rather than taking the mean across stance. All metrics were averaged across trials.

All metrics were compared across weight conditions using repeated measures analysis of variance (ANOVA). For statistically significant main effects ($p < 0.05$), a post hoc test, Holm method, was used for pairwise comparisons. In addition, eta squared effect size was calculated

(η^2). Based on Cohen's guideline, eta effect size was defined as small ($\eta^2 = 0.01$), medium ($\eta^2 = 0.06$), and large ($\eta^2 = 0.14$) effects [39]. Mauchly's test was used to check sphericity violation. If sphericity was violated, Greenhouse-Geisser test or Huynh-Feldt test was used to correct the violation depending on 0.75 of the epsilon values. If the epsilon is bigger than 0.75, Greenhouse-Geisser test was used to correct sphericity violation. Otherwise, Huynh-Feldt was selected for correction of sphericity violation. Furthermore, the Benjamini-Hochberg procedure with a false-discovery rate of 0.10 was used on the ANOVA main effects to account for the multiple tests performed.

Results

Of the 20 total metrics analyzed (3 static, 12 postural control, 5 walking), 17 showed significant main effects due to the weight conditions (Tables 1–3). One of these (AP CoP range both feet) had a p-value greater than 0.05 (0.058), but was identified as significant by the Benjamini-Hochberg procedure.

Static Test

ANOVA showed main effects on plantar spreading forces in both AP ($p < .001$) and ML ($p < .001$) directions. The pairwise comparisons revealed that both AP and ML plantar shear forces were greater in FV and EV than in NV while differences were not seen between FV and EV (Figure 4). There were no differences in CoP location between the weight conditions ($p = 0.176$) (Figure 4). Table 1 provides the outcomes of the static test in summary.

Postural Control Test

As with CoP location in the static test, CoP location in the both-feet trials showed no differences ($p = .36$) (Figure 5). However, there were main effects on CoP location in the single-foot trials ($p = .011$). The pairwise comparisons revealed that the CoP location was more anterior in FV as compared to NV and even EV (Figure 5). The statistical analysis showed main effects on CoP speed in both-feet trials ($p < .001$) and single-foot trials ($p = .012$) with FV and EV decreased compared to NV, and FV further reduced compared to EV just in the both-feet trials (Figure 5).

The Benjamini-Hochberg procedure found main effects on AP CoP range in both feet trials ($p = .058$), but no pairwise comparisons were significant (Figure 6). No main effects on AP CoP range in single-foot trials ($p = .392$). Meanwhile, ANOVA showed main effects on ML CoP range in both-feet ($p = .002$) and in single-foot trials ($p = .005$), respectively. The post hoc tests revealed that FV and EV decreased ML CoP range in comparison to NV (Figure 6).

For AP TTB, ANOVA found main effects in both-feet trials ($p < .001$) and in single-foot trials ($p = .015$). The post hoc tests observed that FV and EV increased AP TTB in both-feet trials as compared to NV while only EV showed an increase in AP TTB in single-foot trials compared to NV (Figure 7). Likewise, the statistical analysis showed main effects on ML TTB in both-feet trials ($p < .001$) and in single-foot trials ($p = .029$). The post hoc tests observed that, similar to the outcomes of AP TTB, FV and EV increased ML TTB in both-feet trials as compared to NV, while only EV showed an increase in AP TTB in single-foot trials compared to NV (Figure 7). Table 2 provides the outcomes of the standing test in summary.

Walking Test

The average walking speed was 1.22 m/s (± 0.28). ANOVA found main effects on peak braking force ($p < .001$), with FV and EV increased compared to NV. For braking and propulsive impulses, main effects occurred ($p < .001$, $p < .001$, respectively) (Figure 8). According to the pairwise comparisons, FV and EV increased braking impulse and propulsive impulse compared to NV. Meanwhile, EV showed further increase in braking impulse and propulsive impulse as compared to FV (Figure 8). For plantar shear spreading forces, there were main effects in both AP ($p < .001$) and ML ($p < .001$), respectively. Based on the pairwise comparisons, FV and EV increased AP plantar shear forces and ML plantar shear forces as compared to NV. Interestingly, while AP plantar shear spreading forces were even greater in FV than in EV, ML plantar shear forces were even greater in EV than in FV (Figure 8). Table 3 provides the outcomes of the walking test in summary.

Discussion

Static Test

The purpose of the static test was to isolate the effects of added body mass on plantar loading. As expected, added body mass resulted in increased plantar spreading in both AP and ML directions. While this result was expected, this study represents the first measurement of plantar shear forces in this context. Plantar spreading forces were slightly increased above the 20% increase in net vertical force, with ML spreading (34% FV and 25% EV) slightly greater than AP spreading (21% FV and 29% EV), but both suggestive of deforming effects on the transverse and medial longitudinal foot arches. A few previous studies have indirectly investigated the effects of mass on medial longitudinal arch deformation. Comparing seated and standing arch height, Butler at al. reported a 3 mm change in arch height [40] while Xiong et al. reported a 6% change in foot length [41]. Wright et al. added 10 kg to the knee of seated participants, measuring a 1 mm drop in arch height that decreased as the tibia moved anteriorly [42]. However, these studies examined normal weight loading, rather than changes due to loading beyond normal body weight. Kern et al. used a weighted vest during walking, finding no difference in arch height with added mass [43]. Yet, obese individuals have lower arches [44]—this plantar spreading could represent a mechanism for long-term arch collapse. We did not see differences in spreading between FV and EV conditions, nor did we see a change in CoP location in the static tests. This is likely due to the controlled posture and hand assistance, resulting in elevated activity of the neuromuscular system in order to maintain the upright position. It is also plausible that the weight distribution in the FV condition did not change the CoM far enough anterior to see larger effects on these variables.

Postural Control Test

Similar to CoP location in the static test, CoP location in the both-feet trials also did not show any differences. However, in the single-foot trials FV elicited a 3% forward-shift in CoP. The disparity between the different trials on CoP location implies that the harder a given task, the more remarkable changes occur. In the single-foot standing, the participants were instructed to slightly bend the knee of the support leg in order to avoid confounding effects, such as joint locking. The flexed knee may have contributed to the forward shift of the CoP location by preventing compensatory trunk extension.

In contrast to our hypothesis, FV and EV consistently showed lower CoP range, lower CoP speed, and greater TTB(s). These changes are traditionally thought to represent an increase in postural stability. However, this perspective should be interpreted cautiously. The results may be due to the use of healthy participants that adapted to the added loads easily. The weighted vests may have acted as external stimuli causing the participants to focus more intently on achieving postural equilibrium. In fact, one previous study observed that young subjects demonstrated increased variability in neutral position as compared to AP leaning conditions [38]. On the other hand, in the same study, older participants exhibited decreased variability in a neutral position as compared to leaning conditions. In the present study, the average age of the participants was early 20s (23 ± 3.10 years), and according to the online survey, they participated in active workouts 3–4 times for 1.5 hour per week on average. Unlike a population that suffers from degenerative neuromuscular function, healthy young individuals may be able to easily adapt to the acute effects of physical stimuli. In other words, the seemingly increased postural stability in FV and EV may result from the ability to flexibly adapt to changing conditions, rather than representing real improved postural stability [45]. Interestingly, variability in postural measure is considered a more complicated notion than traditionally thought. Both van Emmerik and van Wegen stated that reduced variability does not always indicate healthy or good postural stability [45]. In this respect, it is reasonable that the seemingly increased stability occurred not only in EV but also in FV, although FV shifted CoP location forward in single-foot trials.

Increased plantar pressure and shear spreading may influence standing postural control, potentially in both positive and negative ways. Previous studies showed increased tactile perception thresholds with added weight, suggesting reduced sensitivity with increased sensory feedback [12], even in healthy individuals [17]. No studies have investigated the relationship between plantar shear stresses and plantar sensitivity, but mechanoreceptors are expected to respond to both pressure and shear. One previous study utilizing plantar shear stress observed greater peak shear in overweight participants with diabetic neuropathy. In fact, obese individuals have also demonstrated reduced plantar tactile sensitivity simultaneously with reduced postural stability [12]. However, the results in the present study imply that the influence of increased plantar loading may depend on the population. It is possible that the greater plantar shear forces may actually provide some postural control benefit to the young, healthy participants. For example, one prior study found that subthreshold stimuli under the feet improved postural control during standing [46]. Alternately, the seemingly improved postural control with added mass may be due to greater reliance on other physical systems (e.g., visual and vestibular systems) in order to compensate for degraded plantar mechanoreceptor sensitivity caused by greater plantar pressure and shear forces. One prior study reported that while subthreshold vibratory stimuli increased vibratory perception threshold under whole-foot area, the healthy young participants improved postural control [47]. In fact, from a meta-analysis, Song et al. noted that sensory reweighting occurs following malfunction of the body system, such as chronic ankle instability [48]. In this regard, FV may trigger greater activities of the visual and vestibular systems as alternatives to insensitive plantar mechanoreceptors, which consequently results in improved postural control. To support the suggestion, however, follow-up studies are obviously necessary.

Walking

Walking was included in this study to evaluate the manner in which added body mass and distribution affects plantar loading during a common dynamic movement, and how this might ultimately influence balance. As expected, we found strong influences in all loading metrics from both conditions, with several differences between conditions. Peak braking force was higher in both mass conditions but not different between them, suggesting that independent of body distribution, a heavy mass causes greater peak braking forces under the foot. This is in line with previous research showing that both asymptomatic young adults with an evenly loaded vest and obese individuals, who tend to have excessive body fat at the abdomen, have showed higher plantar pressure under the heel during walking [14,16].

Unlike standing, excessive body mass produces greater inertial acceleration back and forth as body moves. The greater peak braking forces in the present study may be to compensate greater inertial acceleration in order to prevent the body from falling. In fact, healthy young individuals with a heavy backpack showed increase in double support and decrease in stride length during walking [26], which apparently protects them from falling. This has several clinical implications. For one, a greater braking force has been associated with higher risk of injuries, such as osteoarthritis [49], and could be an indicator of altered gait mechanics.

Similar to peak braking force, both EV and FV also showed increased braking and propulsive impulses compared to NV. However, EV also had higher braking and propulsive impulses compared to FV. Body mass by itself may require more energy and cause greater stresses to the body than mass distribution does. In other words, walking with EV may need greater forces to move forward as greater resistance was overcome. However, the increased impulses may be due to different body positions dependent on load distribution. In fact, one prior study observed that participants walking with a heavy backpack tend to flex their trunks with increased back muscle activation [50,51]. To compensate for a forward-leaning CoM, the participants in FV condition may have hyperextended their trunk during walking in order to reduce stresses to body joints. For a future study, adding kinematic variables will be helpful to better understand the relationship between body distribution and impulses.

Unlike walking impulses, plantar shear spreading forces showed different interactive effects between body mass and body location. As we expected, FV and EV showed greater plantar shear forces than NV. Interestingly, however, ML plantar shear spreading forces were higher in EV than FV, while AP plantar shear spreading forces were higher in FV than EV. The increased plantar shear forces may provide a mechanism for altered foot morphology over time. While we did not measure foot kinematics, Kern et al. did not show any differences in foot kinematics with added body mass [43]. However, the increased plantar shear forces in the present study may imply that altered stresses under the feet could eventually lead to damage of soft tissues over time. However, to clarify the effects of plantar shear forces on foot morphology, follow-up studies are necessary.

Furthermore, as mentioned above, the increased plantar shear spreading forces in both FV and EV may be pertinent to altered plantar mechanoreceptor sensitivity in regards to greater plantar shear stresses in diabetes with plantar neuropathy [52]. In fact, another previous study reported that after walking with a backpack (30% of body weight) for 10 minutes, healthy young participants showed reduced somatosensory function concurrently with increased postural sway [53]. However, the participants did not alter somatosensory function after walking with a double-pack that was the same weight of the backpack. Perhaps the results of the prior study was that they only investigated AP postural sway. If ML postural sway was also examined, it is possible that the double-pack showed ML postural sway. To authenticate the relationship between plantar shear and plantar mechanoreceptor sensitivity, more information is definitely needed from future studies.

Finally, the increased plantar shear spreading could also represent altered energetics. Increased plantar shear stresses are likely associated with increased energy dissipation in the form of heat, which could reduce walking efficiency even beyond that due solely to the added energy needed to raise and propel body mass forward [29]. This may make the proportionally lower efficiency in obesity even more impressive [28]. Of course, the full contribution of shear stresses to walking energetics requires additional research.

Conclusion

Increased body mass and distribution increased plantar shear spreading forces. The different results in the postural control test may be due to the use of healthy young participants, who can flexibly adapt to acute external stimuli, not reflecting real improved postural control with the interactive effects between body mass and distribution. Therefore, follow-up studies are necessary in other populations, such as obese individuals who chronically carry excessive mass in the abdomen. Furthermore, including kinematic and neuromuscular variables will provide further insights on how plantar shear influences physical performances. In conclusion, we believe that plantar shear can be an important indicator on physical function and/or performance which can be diagnosed in a clinic.

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Table 1. Descriptive information of static test (Mean ± SD).

	FV	NV	EV	*P*-value (η^2)
AP Plantar Shear Force (N)	27.8 ± 6.2*	23.0 ± 6.4	29.7 ± 5.9*	< 0.001 (0.45)
ML Plantar Shear Force (N)	41.9 ± 8.4*	31.2 ± 10.2	39.1 ± 7.8*	< 0.001 (0.47)
AP CoP Location (%)	47.2 ± 6.1	45.4 ± 5.6	46.88 ± 5.2	0.176 (0.09)

P-value is ANOVA main effect. * is $p < 0.05$ compared to NV.

Table 2. Descriptive information of postural control test (Mean ± SD).

	Bf_FV	Bf_NV	Bf_EV	*p*-value of Bf (η^2)	Sf_FV	Sf_NV	Sf_EV	*p*-value of Sf (η^2)
AP CoP location (%)	44.7 ± 6.1	45.5 ± 6.1	44.4 ± 6.6	0.360 (0.05)	51.6 ± 3.1^{*+}	49.9 ± 2.9	49.9 ± 3.4	0.011 (0.21)
CoP Speed (cm/sec)	5.4 ± 1.2	7.3 ± 1.6$^{+¶}$	5.8 ± 1.3¶	< 0.001 (0.78)	7.0 ± 1.7	7.8 ± 2.6$^{+¶}$	6.8 ± 2.1	0.012 (0.21)
AP CoP Range (cm)	2.5 ± 0.5	2.8 ± 0.5	2.8 ± 0.6	0.058§ (0.14)	3.6 ± 0.6	3.8 ± 0.6	3.7 ± 0.7	0.392 (0.05)
ML CoP Range (cm)	2.9 ± 0.7	3.3 ± 0.7$^{+¶}$	3.1 ± 0.5	0.002 (0.24)	3.7 ± 0.6	4.3 ± 0.9$^{+¶}$	3.8 ± 0.9	0.005 (0.28)
AP TTB (sec)	4.6 ± 0.1*	3.4 ± 0.9	4.4 ± 0.2*	< 0.001 (0.60)	3.3 ± 0.9	3.0 ± 0.7	3.5 ± 0.7*	0.015 (0.20)
ML TTB (sec)	2.4 ± 0.4*	1.8 ± 0.3	2.3 ± 0.3*	< 0.001 (0.76)	1.1 ± 0.3	0.9 ± 0.3	1.1 ± 0.4*	0.029 (0.17)

Bf is both feet. Sf is single foot. * is $p < .05$ compared to NV. ¶ is $p < .05$ compared to FV. $^{+}$ is $p < .05$ compared to EV. § is main effects from Benjamini-Hochberg procedure despite $p > .05$.

Table 3. Descriptive information of walking test (Mean ± SD).

	FV	NV	EV	*P*-value (η^2)
Peak Braking Force (N)	92.7 ± 32.1*	84.2 ± 26.7	97.6 ± 28.7*	< 0.001 (0.43)
Braking Impulse (N•sec)	9.5 ± 2.4*	8.1 ± 26.7	10.1 ± 28.7*¶	< 0.001 (0.61)
Propulsive Impulse (N•sec)	10.7 ± 2.1*	9.0 ± 2.3	11.6 ± 2.4*¶	< 0.001 (0.69)
AP Spread at Midstance (N)	31.3 ± 5.6*+	25.5 ± 1.5	29.0 ± 2.2*	< 0.001 (0.48)
ML Spread at Midstance (N)	21.4 ± 5.2*	18.2 ± 4.3	23.7 ± 4.8*¶	< 0.001 (0.55)

* is p < 0.05 compared to NV. ¶ is p < 0.05 compared to FV. + is p < 0.05 compared to EV.

Figure 1. Static Trial in FV and Static Trial in EV. (L = FV; R = EV)

Figure 2. Standing Trials (L = both feet in FV; R = single foot in EV)

Figure 3. Walkway with FootSTEPS and laser timers (L = front view of walkway with FootSTEPS in middle; R = side view of walkway including laser timers).

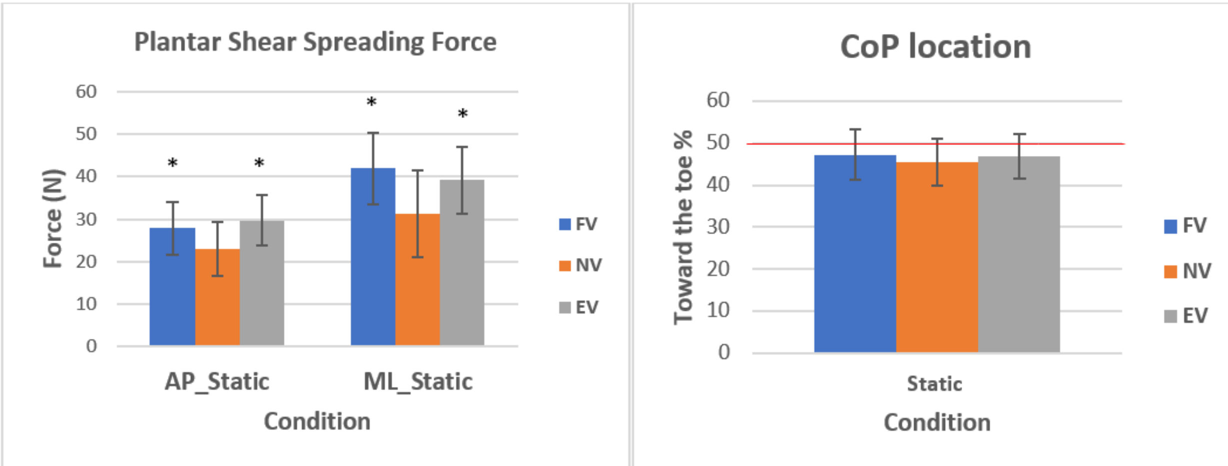

Figure 4. Static AP & ML plantar shear spreading and static CoP location % with standard deviations. The horizontal red line is the central position (50% of a foot length). * is $p < 0.05$ compared to NV.

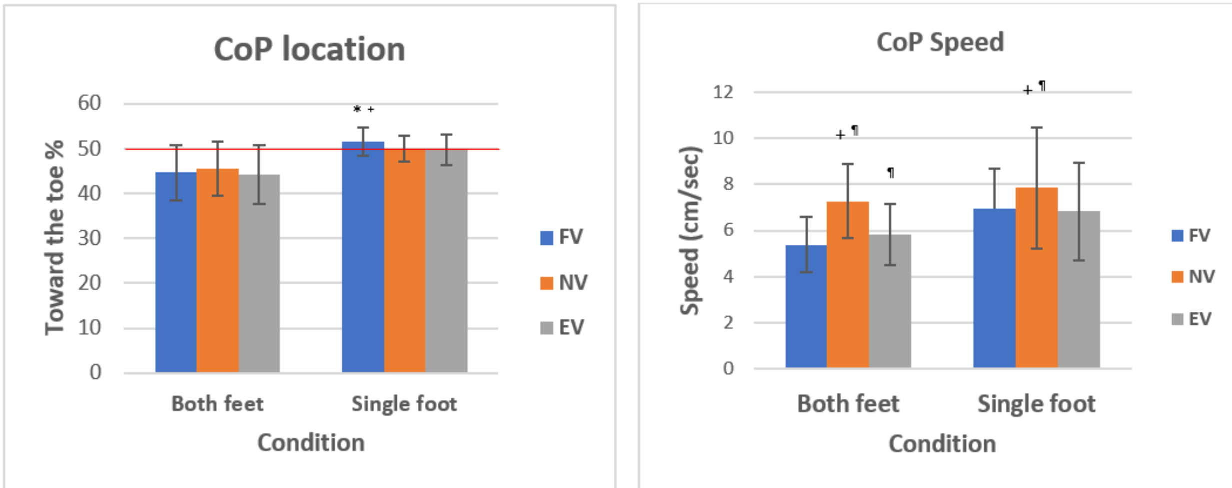

Figure 5. Standing CoP location% in both-feet trials and single-foot trials and CoP speed with standard deviation. The horizontal red line is the central position (50% of a foot length). * is $p < 0.05$ compared to NV. ¶ is $p < 0.05$ compared to FV. + is $p < 0.05$ compared to EV.

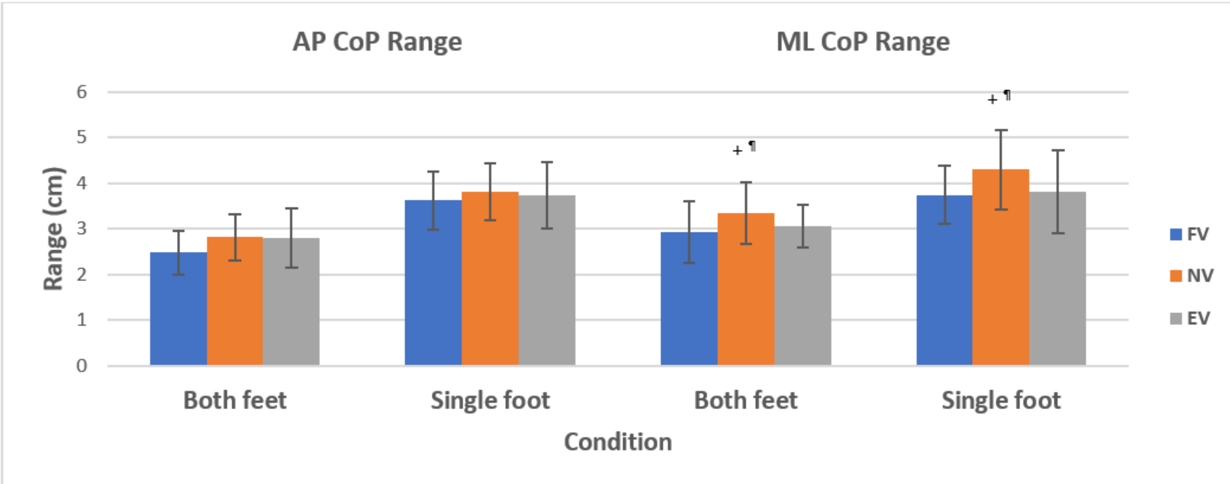

Figure 6. Standing AP & ML foot CoP ranges with standard deviation. ¶ is $p < 0.05$ compared to FV. + is $p < 0.05$ compared to EV.

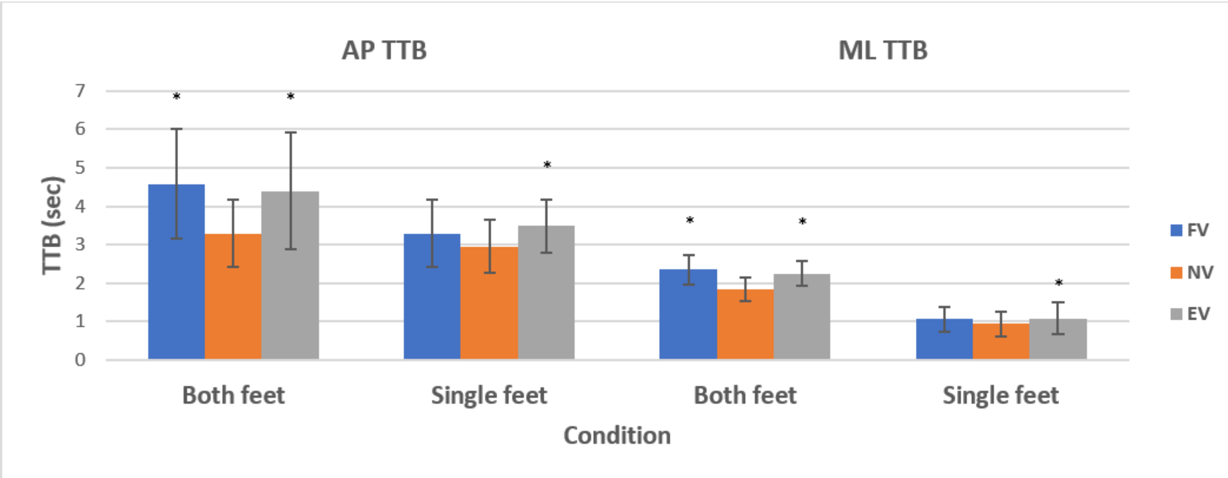

Figure 7. Standing AP & ML TTB with standard deviation. * is $p < 0.05$ compared to NV.

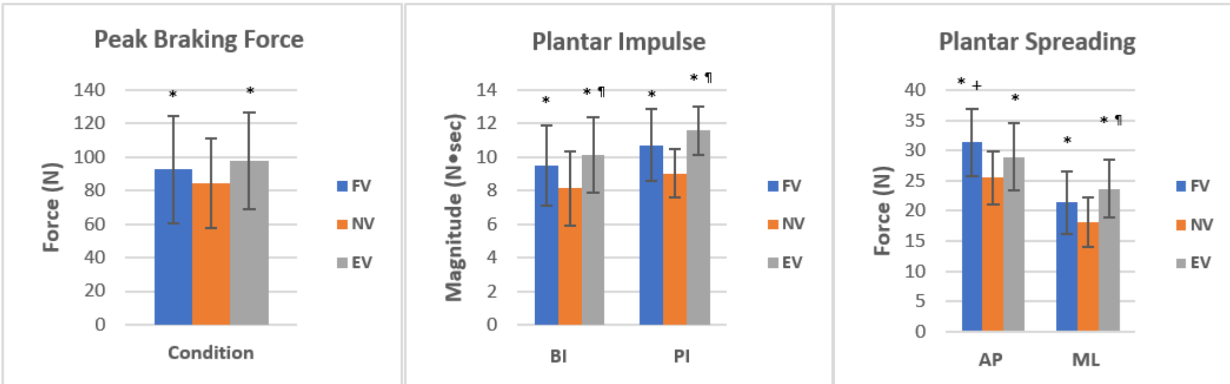

Figure 8. Walking Peak braking force, plantar braking impulse (BI) & propulsive impulse (PI), and AP & ML plantar spreading with standard deviation. * is $p < 0.05$ compared to NV. ¶ is $p < 0.05$ compared to FV. + is $p < 0.05$ compared to EV.