Title: Impaired Plantar Sensitivity among the Obese is Associated with Increased Postural Sway

Authors: Xuefang Wu[a] and Michael L. Madigan[b], Ph.D.

Affiliation: [a] Virginia Tech
Grado Department of Industrial and Systems Engineering (0118),
Blacksburg, VA, 24061
xfwu@vt.edu

[b] Texas A&M University
Department of Biomedical Engineering (3120),
College Station, TX 77843-3120
mlm@tamu.edu

Corresponding author: Michael L. Madigan
mlm@tamu.edu

Tel: 1-979-862-1214

Abstract

Impaired foot plantar sensitivity has been hypothesized among individuals who are obese, and may contribute to their impaired balanced during quiet standing. The objective of this study was to investigate the effects of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and balance during quiet standing. Thirty-nine young adults from the university population participated in the study including 19 obese and 20 non-obese adults. Plantar sensitivity was measured as the force threshold at which an increasing force applied to the plantar surface of the foot was first perceived, and the force threshold at which a decreasing force was last perceived. Measurements were obtained while standing, and at two locations on the plantar surface of the dominant foot. Postural sway during quiet standing was then measured under three different sensory conditions. Results indicated less sensitive plantar sensitivity and increased postural sway among the obese, and statistically significant correlations between plantar sensitivity and postural sway that were characterized as weak to moderate in strength. As such, impaired plantar sensitivity among individuals who are obese may be a mechanism by which obesity degrades standing balance among these individuals.

Keywords

Obesity, plantar sensitivity, postural sway, postural balance

Introduction

An estimated 500 million people worldwide were obese in 2008, and the prevalence of obesity has nearly doubled since 1980 [1]. One of the concerns with the high prevalence of obesity is its association with an increased risk of falls. Each year, obese adults fall almost twice as frequently (27%) as their non-obese counterparts (15%) [2]. This is problematic because falls can be injurious [3]. The biomechanical and/or physiological mechanisms leading to the higher rate of falls among the obese are unclear. Understanding these mechanisms could lead to more effective fall prevention programs.

One mechanism by which obesity could contribute to falls is by degrading balance due to impaired plantar sensitivity on the bottom of the feet. Human standing balance control relies on feedback from the proprioceptive system [4]. This system includes cutaneous mechanoreceptors which detect pressure and deformation in the skin [5]. Studies have demonstrated that impairments in plantar sensitivity influence balance control among older adults and individuals with chronicle ankle instability [6, 7, 8]. Obesity increases postural sway during quiet standing [9], and may do so, at least in part, due to impaired plantar sensitivity. Higher plantar pressures have been reported among individuals who are obese [10], but no studies to our knowledge have investigated the effect of obesity on plantar sensitivity, or the association between plantar sensitivity and balance as a mechanism by which individuals who are obese exhibit impaired balance.

The objective of this study was to investigate the effect of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and postural sway. Our first hypothesis was that obesity would adversely affect plantar sensitivity. Our second hypothesis was that plantar sensitivity would be associated with postural sway. The results from this study will provide insight to the mechanisms by which obesity impairs balance, and potentially guide future efforts aimed at developing interventions to mitigate the delirious effects of obesity on balance.

Materials and Methods

Thirty-nine young (age=21.3±2.6 years) adults recruited from the university population participated in the study. Participants included 19 obese (body mass index or BMI = 33.0±2.9kg/m^2; 14 females and 5 males) and 20 non-obese (BMI = 22.2±2.2 kg/m^2; 14 females and 6 males) adults. Body fat percentage was also measured using skinfold caliper measurements at the front of the upper arm, back of the upper arm, below the scapula, and on the abdomen (1cm to the right of the navel). Obese participants were required to have a body fat percentage above 35% for women and above 25% for men from these caliper measurements [11], as well as a BMI above 30 kg/m^2. All participants were free from any self-reported foot pain or known neurological conditions that might affect their performance in this test.

Participants completed one experimental session during which multiple measurements of plantar sensitivity were obtained while standing. Plantar sensitivity was operationalized as the force threshold at which an increasing force applied to the plantar surface was first perceived, and the force threshold at which a decreasing force was last perceived. Measurements were obtained immediately upon standing, and at two locations on the plantar surface of the dominant foot including the calcaneus and the head of the third metatarsal. Postural sway was then evaluated under three different conditions. Participants wore a T-shirt, tight-fitting pants, no shoes, and no socks during testing. Room temperature was controlled at 74ºF.

The setup and methodology was based upon a recent study investigating the effects of added weight on plantar sensitivity while upright standing [12]. Plantar sensitivity was assessed using a custom-designed platform (Figure 1) and a digital force gauge (Extech, Model 475040, Nashua, NH, USA. The aluminum platform (40 × 81 cm) was covered in vinyl floor tile and included a 1.5mm diameter hole so that a small stainless steel probe tip (diameter = 1mm) attached to the force gauge could pass through and come into contact with participants' foot sole while standing. The position of the probe tip was controlled from beneath the platform via a manual lab jack (LJ750, Thorlabs Inc, Newton, New Jersey, USA) (Figure 1).

Figure 1

Two practice trials were performed on each participant at the beginning of the experiment. Practice trials were performed at a random site on the plantar surface of the foot not including the two testing sites. Participants were then asked to sit for 10 minutes. To start testing, participants were asked to stand on the platform while the investigator positioned their foot so that the testing site was aligned with the hole in the platform. Participants were instructed to stand as still as possible, look straight forward, hold onto the bars in front of them to help stand still, and give verbal indication when they were able to feel the force by saying "Now". At the start of each trial, the probe tip was initially below the surface of the platform and not in contact with the plantar surface of the foot. After a random delay of up to 10 seconds, the investigator began manually rotating a dial on the lab jack (Figure 1) in increments of approximately 60 degrees every half second until given a verbal indication by the participant. Once the probe tip translated upward far enough to contact the plantar surface of the foot, this rotating pattern increased the force applied to the foot in a step-wise manner at a rate of ~5 grams every half second. After the participant detected the force, this force threshold was recorded, and the investigator continued to raise the probe tip until the force reached 180 grams (a value well above all participants' force threshold). The lab jack was then used to translate the force probe tip downward, resulting in the force applied to the foot decreasing in a step-wise manner at a rate of ~5 grams every half second. Participants were instructed to give verbal indication when they were no longer able to feel the force by saying “Now”. A total of four trials were performed at each site, with each trial involving the force increasing and decreasing one time. The order of the two sites was counterbalanced within each group. The experimenter also randomly picked one of the four increasing trials to check whether the participant was giving false verbal indication on their foot sensitivity, or experiencing phantom sensation, by delaying the initiation of the trial for 30 seconds after indicating the start of the trial. None of the participants gave indication before the start of the trial during the experiment.

Postural sway was then evaluated while participants attempted to stand as still as possible with bare feet, arms at sides and feet pointed forward and 7.5cm apart. The trials were collected under three different sensory conditions: eyes-open (baseline), eyes-closed (impaired visual feedback), and eyes-closed with the head tilted backward (impaired visual, vestibular, and proprioceptive feedback). Tilting the head backward is thought to render balance-related vestibular information unreliable by placing the otolith organs outside their normal working range [13]. These conditions were imposed because impairing the visual and vestibular systems would make the balance control system more dependent upon the proprioceptive system (e.g. plantar sensitivity), and may strengthen the relationship between plantar sensitivity and balance. Participants were required to tilt their head backwards at least 30º, as measured by investigator observation. Three trials of 75 s were collected under each condition, and two minutes of rest were allowed in between consecutive trials. The order of the trials was randomized within each group.

During standing trials, ground reaction forces were sampled at 1000 Hz using a force platform (Bertec Corporation, Columbus, OH, USA), and low-pass filtered at 7 Hz using a 4th order Butterworth zero-lag filter . Dependent variables during standing included center of pressure (COP) mean velocity, and COP root mean square (RMS) distance from the mean position in the radial direction. Mean velocity was defined as the total COP distance traveled divided by collection time. The RMS distance was defined as the standard deviation about the mean COP position in radial direction [14]. For each standing trial, the initial and final 5 seconds of the data was removed to avoid initial transients and termination anticipation effects, respectively [14].

A three-way mixed-model analysis of variance (ANOVA) was performed on force threshold measurements with independent variables including group (obese or non-obese), location (head of the third metatarsal or calcaneus) and force direction (increasing or decreasing). A two-way mixed-model ANOVA was performed on postural sway measures with independent variables including group (obese or non-obese) and condition (baseline, impaired vision, or impaired vestibular feedback). Tukey's Honestly Significant Difference procedure was used to investigate pair-wise comparisons of interest in the event of significant interactions. A log transform on force threshold measurements and postural sway measures was performed prior to the analyses to achieve a normal distribution of residuals.

Bivariate correlation analysis was performed between plantar sensitivity and postural sway measures, and the strength of the correlation was quantified using the Pearson product-moment correlation coefficient. Mean of the four plantar sensitivity trials under each testing location and force direction was correlated with the mean of the three postural sway trials under each condition. Two data points were associated with substantially larger Cook's distance values [15], and thus were excluded from the bivariate analyses to avoid a disproportionate influence on the correlation. The strength of correlations were characterized using the correlation coefficient

(r) as strong (0.6-0.8), moderate (0.4-0.6), and weak (0.2-0.4) [16]. JMP 10 (SAS Institute Inc., Cary, NC, USA) was used to carry out the statistical analyses, and statistical significance was concluded if $p \leqslant 0.05$.

Results

Force threshold measurements exhibited a mean value of 29.8 grams and a range of 2 - 156 grams across all groups (Figure 2). It exhibited a group by location by force direction interaction (p=0.040; Figure 3). Under the third metatarsal, the obese group exhibited a 56% higher force threshold when force was increasing (p<0.001), and a 22% higher force threshold when force was decreasing (p=0.008). Under the calcaneus, the obese group exhibited a 30% higher force threshold when force was increasing (p=0.019), and a 22% higher force threshold when force was decreasing (p<0.001). The force threshold across both groups was 79% higher at the calcaneus compared to the third metatarsal when force was increasing (p<0.001), but not significantly different between these locations when force was decreasing. Lastly, the force threshold across both groups was 183% higher under the calcaneus when force was increasing (p<0.001), and 91% higher under the third metatarsal when force was increasing (p<0.001).

Figure 2

Figure 3

Both mean velocity and RMS distance exhibited no group by condition interaction (p=0.344 for mean velocity and p=0.179 for RMS distance), but did exhibit main effects of group (p=0.008 for mean velocity and p=0.016 for RMS distance) and condition (p<0.001 for mean velocity and RMS distance). The obese group exhibited 5% higher mean velocity and 7% higher RMS distance compared to the non-obese group.

Bivariate analyses revealed positive correlations of weak to moderate strength between two plantar sensitivity measurements (calcaneus/decreasing force and 3rd metatarsal/increasing force) and both postural sway measurements under the three different conditions (Figure 4). No other correlations were significant.

Figure 4

Discussion

The objective of this study was to investigate the effect of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and postural sway. We hypothesized obesity would adversely affect plantar sensitivity. The threshold to detect force onset and offset were both higher among obese participants at both locations tested. These results indicated obese participants were unable to sense plantar forces as small as non-obese participants while standing. As such, this hypothesis was accepted. We also hypothesized that plantar sensitivity would be associated with postural sway. This hypothesis was accepted because weak to moderate correlations were found between the postural sway and plantar sensitivity measures.

The mean force thresholds, and the effects of obesity on force thresholds, measured here are comparable to a prior study that investigated the effects of adding body mass on plantar sensitivity [12]. Handrigan et al. [12] measured the minimum force threshold while standing, and compared these measurements before and after adding 23 kg of body mass with a weighted vest. This added body mass corresponded to an increase in BMI from 24.7 to 31.9 kg/m^2. The mean force threshold reported by Handrigan et al. [12] was 20.0 g before adding weight, which was comparable to the 25.6 g threshold measured among non-obese participants in the current study across all conditions. Handrigan et al. [12] also reported a 30% increase in force threshold after adding 23 kg of body mass, which was comparable to the 29.3% higher threshold among the obese group in the current study across all conditions. Together, these studies provide evidence for the added body mass associated with obesity eliciting impaired plantar sensitivity.

Two characteristics of mechanoreceptors may help to explain the impaired plantar sensitivity among obese participants found here. First, the relationship between mechanical stimulus intensity and internal mechanoreceptor voltage potential is nonlinear, and suggests less sensitivity at higher intensities [17]. At lower stimulus intensities, only slight changes in stimulus intensity are needed to markedly increase the mechanoreceptor potential. At higher stimulus intensities, however, an equivalent increase in stimulus intensity only results in a slight increase in mechanoreceptor potential. The higher plantar pressure associated with obesity [10] may result in the operating range of the plantar mechanoreceptor to be closer to the range over which changes in stimulus intensity only result in slight changes in mechanoreceptor potential. Second, the Weber-Fechner Principle states that the amount of change in mechanical stimulus intensity necessary for detection is proportional to the stimulus intensity [18]. As such, the higher plantar pressure associated with obesity would require a larger change in plantar mechanical stimulus to be detected, and result in reduced sensitivity on the plantar surface of the foot.

Force thresholds were lower when force was decreasing than increasing. We have two possible explanations for these results. First, Meissner's corpuscles and Pacinian corpuscles are two rapidly adapting mechanoreceptors responsible for detecting the onset and offset of a mechanical stimulus [19]. Meissner's corpuscles are thought to be more amenable to detecting light touch due to their relatively superficial distribution within the skin, and Pacinian corpuscles are thought to be more amenable to detecting deep pressure due to their deeper distribution within the skin [19]. Given these differences, detecting the onset of an initially zero force that increased may be mostly dependent upon Meissner's corpuscles, whereas detecting the offset of an initially high force (that stimulated deeper Pacinian corpuscles) and decreased may have involved both Meissner's and Pacinian corpuscles. Involvement of Pacinian corpuscles may have offered more sensitivity while force was decreasing because Pacinian corpuscles have a lower response threshold than Meissner's corpuscles [20]. Our second possible explanation is related to the mechanical properties of the skin [21]. Skin is viscoelastic, which means its mechanical response to force depends upon time. Measurements as force increased were made shortly (less than 18 seconds, on average) after the force was applied, and measurements as force decreased were made after the force was applied for at least 30 seconds. Therefore, the viscoelastic stress relaxation that would have occurred during testing likely resulted in differences in the mechanical state (i.e. tissue strain) at the testing site that may have contributed to differences in force thresholds.

Our results indicated that plantar sensitivity is weakly to moderately correlated with postural sway. Mechanoreceptors are preferentially distributed in the anterior, lateral border and heel regions of the plantar surface [20], which could correspond to the critical regions of the foot that support the majority weight of the body under the weight-bearing condition [22]. Similarly, mechanoreceptors under the anterior and posterior regions of the feet provide feedback that regulate the body to tilt posteriorly and anteriorly [22]. The central nervous system may be able to extract a spatial distribution cue according to the plantar pressure which could be transformed into a body position cue indicating the direction and the amplitude of the whole-body inclination [23]. Therefore, deficits in plantar sensitivity could have a direct influence on balance control [6, 7, 8]. However, the mechanism of why only calcaneus/decreasing force and 3rd metatarsal/increasing force were correlated with postural sway measurements under the three different conditions is not clear.

The postural sway condition with eyes closed and head tilted backward was included based upon the expectation that any impairment in plantar sensitivity among obese participants would have a greater effect on postural sway when input from the visual and vestibular systems were eliminated or altered. Based upon the lack of a group by condition interaction in sway measures, this expectation was not met. This may have been due to not tilting participants head backward sufficiently far, or input from other components of the proprioceptive system offsetting the altered input from the plantar surface among the obese. Nevertheless, postural sway exhibited weak to moderate correlation with plantar sensitivity despite not adequately impairing vestibular input during sway measurements.

Several limitations are to be noted for this study. First, changes in force were stepwise and in increments of approximate 5 gram per half second, rather than at a steady rate. This may have limited our force threshold resolution 5 grams, but this amount of force was ostensibly sufficiently small to identify the effects of interest. Second, the results may be dependent upon the nature and duration of activities performed before testing (someone may have been standing for hours while another may have been laying down). However, these data were not obtained. Third, plantar sensitivity was only obtained from the dominant foot, and it is unclear if any left/right asymmetry in plantar sensitivity is common. Fourth, the amplitude of head tilting was not strictly controlled in the current study, and may have contributed to some inter-subject variability in the vestibular feedback during this condition.

In conclusion, obesity impaired plantar sensitivity among a cohort of young adults. This impairment was associated with increased postural sway during quiet standing, and may be a contributing factor to the increased fall risks among individuals who are obese.

Acknowledgements

This work was supported by award R01OH009880 from the CDC-NIOSH. The contents of this paper are solely the responsibility of the authors and do not necessarily represent the official views of the sponsor.

Conflict of Interest

None

Reference

[1] World Health Organization (WHO), Obesity and overweight 2011. http://www.who.int/mediacentre/factsheets/fs311/en/. Last accessed March 20th, 2014, Blacksburg, Virginia, USA.

[2] C. Fjeldstad, A.S. Fjeldstad, L.S. Acree, K.J. Nickel, A.W. Gardner, The influence of obesity on falls and quality of life, Dynamic Medicine 7 (4) (2008) 27.

[3] K.C. Matter, S.A. Sinclair, S.G. Hostetler, H. Xiang, A comparison of the characteristics of injuries between obese and non-obese inpatients. Obesity 15(10) (2007) 2384-2390.

[4] S. Zhang, L. Li, The differential effects of foot sole sensory on plantar pressure distribution between balance and gait, Gait Posture 37(4) (2013) 532-535.

[5] D.A. Winter, A.E. Patla, J.S. Frank, Assessment of balance control in humans, Medical Progress through Technology 16 (1990) 31-51.

[6] M.R. Powell, C.J. Powden, M.N. Houston, M.C. Hoch, Plantar cutaneous sensitivity and balance in individuals with and without chronic ankle instability, Clinical Journal of Sports Medicine 2014 Epub Ahead.

[7] O. Bretan, R.M. Pinheiro, J.E. Corrente, Balance and plantar cutaneous sensitivity functional assessment in community dwelling elderly, Brazil Journal of Otorhinolarynogoly 76(2) (2010) 219-224.

[8] H.B. Menz, M.E. Morris, S.R. Lord, Foot and ankle characteristics associated with impaired balance and functional ability in older people, Journal of Gerontology 60(12) (2005) 1546-1552.

[9] N. Teasdale, M. Simoneau, P. Corbeil, G. Handrigan, A. Tremblay, O. Hue, Obesity alters balance and movement control, Current Obesity Reports 2 (2013) 235-240.

[10] A.P. Hills, E.M. Henning, M. McDonald, O. Bar-Or, Plantar pressure differences between obese and non-obese adults: a biomechanical analysis, International Journal of Obesity 25 (2001) 1674-1679.

[11] World Health Organization (WHO) Physical status: the use and interpretation of anthropometry: report of a WHO Expert Committee. Published 1995. Geneva, Switzerland: WHO Technical Report Series 854; p378 http://whqlibdoc.who.int/trs/WHO_TRS_854.pdf Accessed March 31st, 2014.

[12] G.A. Handrigan, M. Simoneau, N. Teasdale, P. Corbeil, The effects of added mass on plantar sole sensitivity in upright standing, Journal of Biomechanics 45(Supplement 1) (2012) S233.

[13] N. Vuillerme, O. Chenu, N. Pinsault, A. Fleury, J. Demongeot, Y. Payan, Can a plantar pressure-based tongue-placed eletrotactile biofeedback improve postural control under altered vestibular and neck proprioceptive conditions? Neuroscience 155 (1) (2008) 291-296.

[14] D. Lin, H. Seol, M.A. Nussbaum, M.L. Madigan, Reliability of COP-based postural sway measures and age-related differences, Gait & Posture 28(2) (2008) 337-342.

[15] D.C. Montgomery, E.A. Peck, G.G. Vining, Introduction to Linear Regression Analsis, 5th Edition, Wiley, New York, 2012.

[16] M.H. Kunter, C.J. Nachtsheim, J. Neter, W. Li, Applied linear statistical Models. Irwin: McGraw-Hill, Columbus, 2004.

[17] A.C. Guyton, J.E. Hall, Textbook of medical physiology, pp575, 594, 11th edn. Elsevier Saunders, 2006.

[18] R.G. Lanzara, Weber's law modeled by the mathematical description of a beam balance, Mathematical Biosciences 122 (1994) 89-94.

[19] M.S. Fleming, W. Luo, The anatomy, function, and development of mammalian Aβ low-threshold mechanoreceptors, Front Biology 8(4) (2013) 408-420.

[20] P.M. Kenndy, J.T. Inglis, Distribution and behavior of glabrous cutaneous receptors in the human foot sole, Journal of Physiology 538(Pt 3) (2002) 995-1002.

[21] B.H. Pubols, Factors affecting cutaneous mechanoreceptor response. I. constant-force versus constant-displacement stimulation. Journal of Neurophysiology 47(3) (1982) 515-529.

[22] S.D.Perry, W.E. Mcllroy, B.E. Maki, The role of plantar cutaneous mechanoreceptors in the control of compensatory stepping reactions evoked by unpredictable, multi-directional perturbation, Brain Research 877(2) (2002) 401-406.

[23] A. Kavounoudias, R. Roll, J. Roll, Foot sole and ankle muscle inputs contribute jointly to human erect posture regulation, Journal of Physiology 532 (2001) 869-878.

Figure legends

Figure 1 Custom-designed platform to assess plantar sensitivity while standing (Digital force gauge setup was mounted on lab jack under the platform. The investigators adjusted the vertical position of the probe by manipulating the lab jack, and read off the force threshold from the digital readout when indicated by the participants.)

Figure 2 Force threshold measurements separated by group, location, and force direction. Dots indicate individual measurements, and brackets indicate 95% confidence intervals of the mean.

Figure 3 Group by location by force direction interaction plot illustrating differences between groups and conditions * indicates $p \leq 0.05$, **indicates $p \leq 0.01$. B indicates statistical significance when combining across both groups.

Figure 4 Scatter plots of correlation between plantar sensitivity (calcaneus/immediately upon standing/decreasing force and 3rd metatarsal/immediately upon standing/increasing force) and postural sway measures (mean velocity and RMS distance) with r and p values.

Figure 1
Click here to download high resolution image

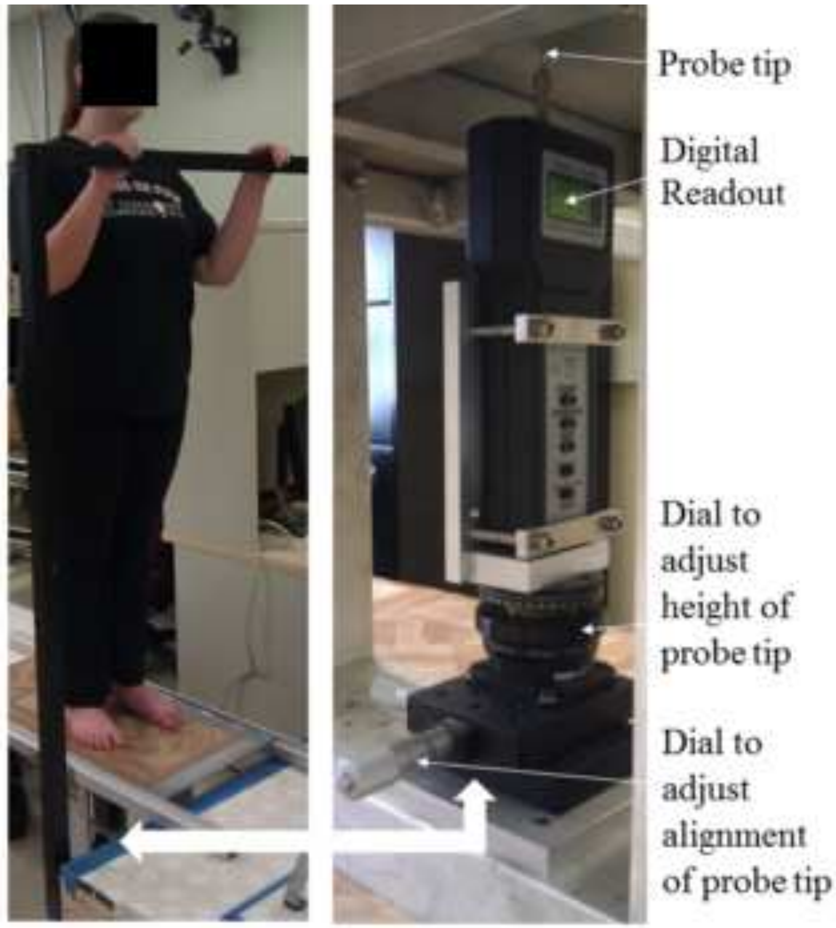

Figure 2

Click here to download high resolution image

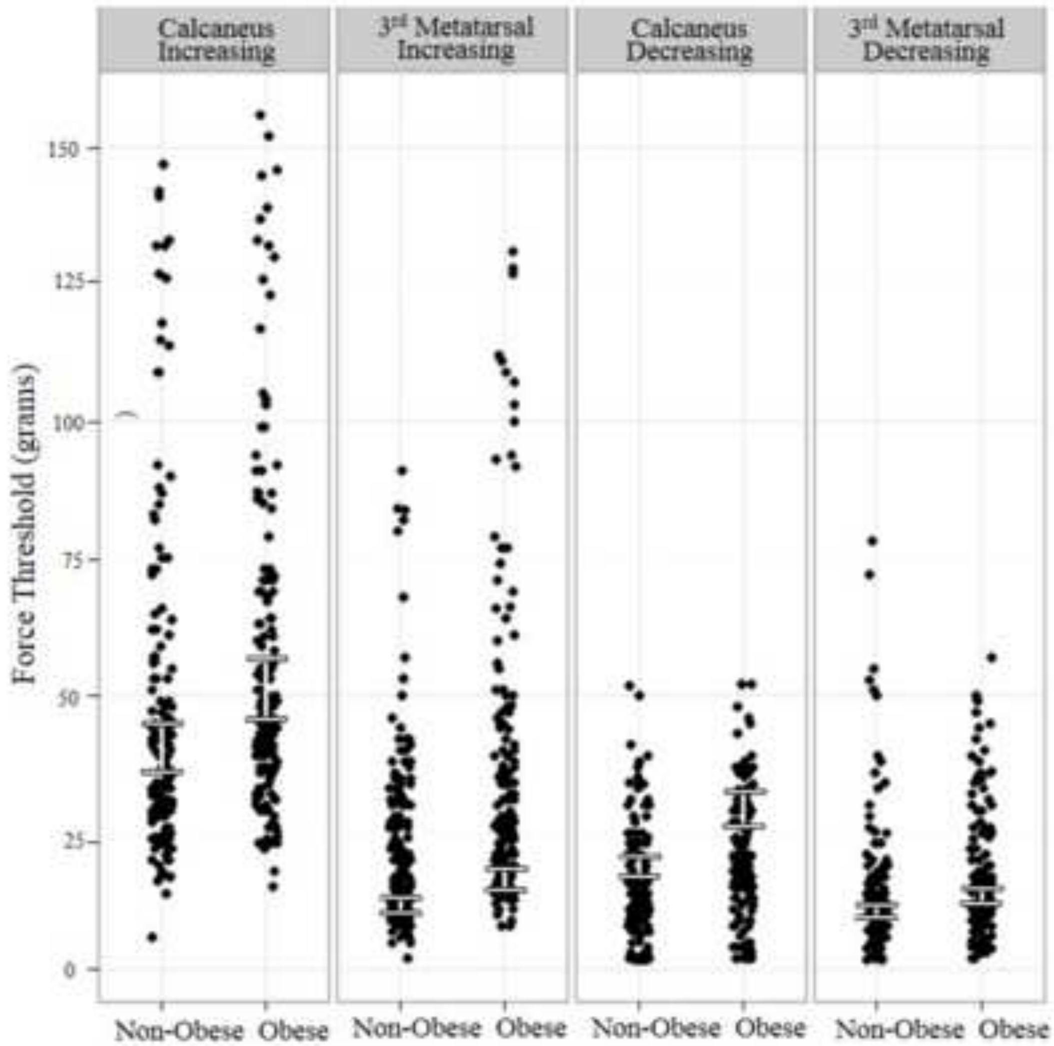

Figure 3
Click here to download high resolution image

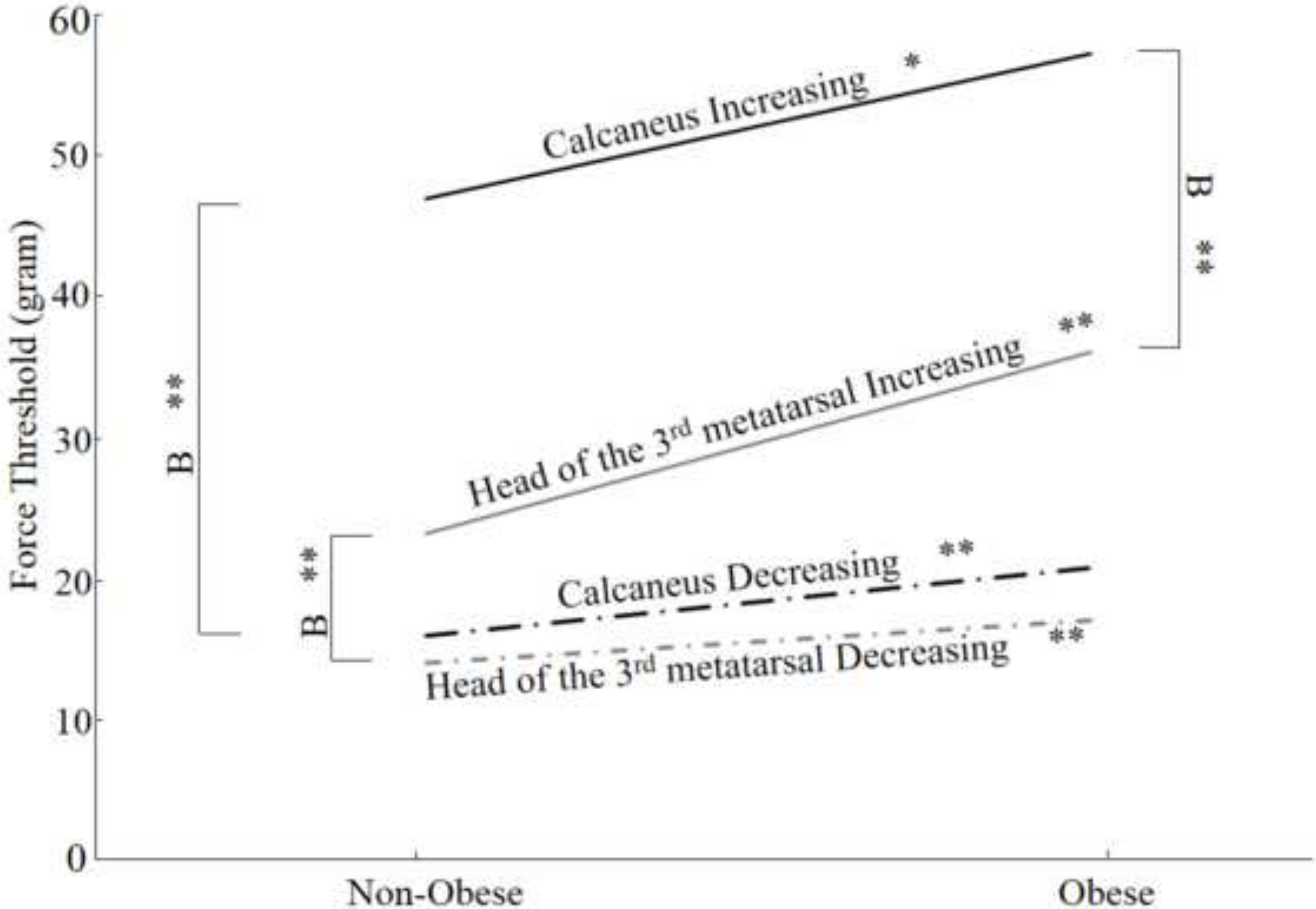

Figure 4
Click here to download high resolution image