UNDERSTANDING THE INDEPENDENT EFFECTS OF INERTIA AND WEIGHT ON BALANCE

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ABSTRACT

While human balance is known to be affected by altered sensory feedback, altered dynamics may also contribute to balance deficiencies in certain populations. The goal of this study was, therefore, to investigate the effects of altered dynamics, namely increased inertia and increased weight, on standing balance. Sixteen normal-weight male participants completed quiet standing in a custom-built backboard under four conditions: baseline, increased inertia only, increased weight only, and increased inertia and weight. Increased inertia did not affect body center of mass movement (COM) or center of pressure (COP) movement, suggesting that no additional ankle torque was necessary to control the increased inertial forces. Increased weight caused increased body COM movement (increased backboard angle range and angular speed) and greater acceleration of the COM (as evidenced by increased COP-COM), requiring an increased level of corrections needed to maintain upright posture (as evidenced by increased COP speed) and increased ankle torques (as evidenced by increased range of COP position). Increasing inertia and weight simultaneously had the same effects as increasing weight except that there was no increased COM movement when both inertia and weight were increased. This indicates that there may be a slight mediating effect of increasing inertia on the extreme changes in balance observed when only weight is increased. These results indicate that altered dynamics of the body have an effect on human standing balance, just as altered sensory function has an effect on balance.

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CHAPTER 1 – OVERVIEW

Obesity is an increasingly prevalent medical condition that is associated with numerous adverse health conditions, including an increased risk for falls and injuries resulting from falls. Standing balance, a common indicator for fall risk, is also impaired in the obese. Obesity is associated with both sensory deficits and altered dynamics, both of which may contribute to impairments in standing balance. While numerous studies have investigated the effects of altered sensory feedback on balance, few have investigated the effects of altered dynamics such as altered inertia and weight. A better understanding of the effects of altered dynamics on balance will give insight into human standing balance and fall risk in the obese, as well as in other populations exhibiting altered dynamics.

With this in mind, Chapter 2 explores the current literature on the effects of obesity on falls, balance, sensory impairment, and dynamics. Chapter 3 examines the independent and combined effects of increased inertia and increased weight on balance from a fundamental viewpoint in an effort to better understand the effects of altered dynamics on human standing balance. Chapter 4 reports the conclusions of the study presented in Chapter 3.

CHAPTER 2 – INTRODUCTION

Increasing Prevalence of Obesity as a Public Health Concern

Obesity is an epidemic. Over one billion people worldwide are overweight, with 300 million of those considered obese (body mass index, BMI > 30 kg/m^2) [1]. In the U.S. alone, two out of every three adults is considered to be overweight or obese, with 32.2% of adults (20+ years old) obese and 4.8% morbidly obese [2, 3]. Additionally, the prevalence of obesity is increasing. Globally, the prevalence of obesity increased from 200 million to 300 million people from 1995 – 2004 [4, 5]. In the U.S., the prevalence of obesity has more than doubled since 1980 [6] while the prevalence of morbid obesity has quadrupled in nearly the same timeframe [7].

The high and increasing prevalence of obesity is particularly worrying because of the numerous health complications associated with obesity. Increased BMI causes an increased risk for type II diabetes, cardiovascular disease, kidney disease, high cholesterol, osteoarthritis, obstructive sleep apnea, certain cancers, and early mortality [4, 5, 8-15]. These medical conditions associated with obesity lead to increased medical costs of $147 billion annually in the U.S., or 9% of the total national annual healthcare expenditures [9, 16, 17]. This amounts to approximately $732 more per year in medical care costs for an obese individual compared to a non-obese individual [18]. In addition to the numerous metabolic and cardiovascular impairments associated with obesity, obese individuals also experience a decreased quality of life compared to non-obese individuals [9, 19], potentially resulting from a higher level of musculoskeletal impairment.

Musculoskeletal Disability Associated with Obesity

Activity limitations are common in the obese [20, 21] and may result from decreased strength relative to body weight, increased levels of perceived exertion, faster time to fatigue, and altered movement patterns. While obese individuals have greater absolute strength in push and pull tasks [22], hand grips tasks [23], and knee extension [24], obesity is associated with significantly lower peak knee extensor torques when normalized to body weight [25], suggesting that the advantages of increased absolute strength in the obese are diminished by having to move additional body mass. This translates to worse performance by obese children in weight-bearing fitness tests such as standing broad jumps, bent-arm hang, and both speed and endurance shuttle runs [23].

Obese individuals also have higher energy expenditure during both sleep/sedentary activities and walking than non-obese individuals (even after normalization for body weight) [26, 27] and decreased muscle oxidative capacity versus non-obese individuals [28]. Additionally, there is some evidence that obese individuals have faster rates of fatigue than non-obese individuals [29], which has been documented to alter the parameters controlling standing balance. A study by Hlavackova *et al.* [30] found that localized muscle fatigue of the plantar flexor muscles increased center of pressure (COP) speed during standing balance. The COP represents location of the net ground reaction force under the feet and its movement is typically used as a measure of postural sway (with greater COP movement, and thus larger postural sway indicating reduced balance ability).

Valgus deviation of the knees is very prevalent in the obese [31-33] and has been suggested to be a risk factor for osteoarthritis due to the abnormal loading of the lateral compartment of the knee joint [34]. Valgus deviation also leads to alterations in gait and can cause pain and difficulty in movement [33], two common complaints in the obese.

During gait, morbidly obese women walk at slower self-selected walking speeds with lower cadence, smaller stride length, and a wider support base than normative data [35]. These alterations (slower speeds, smaller strides, etc.) are similar to those seen in other patient populations, such as the elderly [36, 37] or individuals with osteoarthritis [38], and are thought to be a protective measure aimed at maintaining dynamic stability and balance. Additionally, obese individuals have increased heel forces [39], an increased rating of perceived exertion [24], and increased ankle, but not knee or hip, joint torques and power [40] during walking.

These findings on musculoskeletal limitations in the obese suggest that obese individuals may have to "work harder," – use more strength and expend more energy – in order to achieve the same movement goals (e.g. walking, balance) as non-obese individuals.

Obesity, Balance, and Falls

The musculoskeletal disability and increased effort required by obese individuals during common movement tasks may be responsible for the increased rate of falls in obese individuals. Obese individuals have a greater prevalence of falls and ambulatory stumbling than non-obese individuals [41], and falls are the leading cause of injury in the obese (~36% of all injuries in obese individuals result from a fall) [42]. Falls are a large issue in the obese population because obese individuals are more likely to sustain injuries due to falls [43] and are more likely to have prolonged recovery following injury [44-46] as compared to non-obese individuals.

Consistent with this data on increased fall rate in the obese, obese individuals exhibit increased postural sway compared to non-obese individuals, a measure that is often used to quantify balance [47]. Body weight is strongly correlated with postural sway as measured by COP speed [48, 49], and both obese men and women have been shown to exhibit increased COP speed and amplitude in the anterior-posterior (AP) direction than non-obese individuals [50]. Additionally, morbidly obese individuals have shown faster increases in postural sway during prolonged quiet standing than non-obese individuals [51], suggesting that not only is the balance of obese reduced in comparison to non-obese, but the rate of balance decline during prolonged standing is also higher in the obese than the non-obese.

Effects of Altered Sensory Feedback

Some of the balance deficits observed in obese individuals may result from impaired sensory feedback. The main sensory changes resulting from obesity include increased levels of pain and decreased plantar sensation.

Obese individuals experience musculoskeletal pain more frequently than non-obese individuals, with both obese children and adults often reporting chronic pain in the low back, hip, knee, and foot [24, 52, 53]. Induced pain in the ankle musculature or feet has been shown to increase postural sway (range, standard deviation, and mean speed of COP) [54, 55], suggesting that musculoskeletal pain due to obesity may increase risk of falls. Additionally, induced pain in the hands does not increase postural sway, suggesting that it is not merely the pain sensation that causes balance deficits but the fact that the pain is located on the limb responsible for controlling balance [55].

Obese individuals also experience various differences in foot structure and function that may lead to additional deficits in balance ability due to altered sensory information and pain. For one, obese individuals experience increased plantar pressures [56-60]. Increased plantar pressures have been associated with increased pain [61] and even tissue damage [62] in other populations and are present in the obese despite an increased foot-ground contact area over which the additional weight of the obese individual could be distributed [56-60, 63]. This increased contact area comes about in part because of a flatter foot with a lower arch in obese individuals [64]. Any deviations from a neutral foot position have previously been reported to cause poorer postural control during standing [65], therefore the altered foot structure in obese is likely to cause increased postural sway during standing balance.

Obese individuals also have an increased incidence of plantar heel pain [66]. This may result in part from the increased risk of developing plantar fasciitis, the most common cause of inferior heel pain, with increasing body mass index (BMI) [67]. Additionally, obese individuals have increased heel fat pad thickness [66, 68, 69] and similar [68] or increased heel pad compressibility indices (i.e. decreased elasticity of the heel pad) [66, 69], either of which could impair plantar mechanoreception, which has also been shown to increase postural sway [70, 71].

Effects of Altered Dynamics

While perturbing sensory feedback has been well documented to cause balance deficits and may explain some of the reduced balance ability observed in obese individuals, perturbing the dynamics of the body may also affect balance ability. Altered dynamics include any physical aspect of the body that results in changes in the governing equations of motion of the body. For example, changes in mass moment of inertia (inertia) or forces acting on the body due to the effects of gravity (weight) result in changes in the governing equations of motion of human standing and thus are considered altered dynamics. In obese individuals, both increased inertia and increased weight have the potential to contribute to the balance deficits observed in this population. However, the biomechanical mechanisms by which increased inertia and increased weight independently and simultaneously contribute to balance are not well understood.

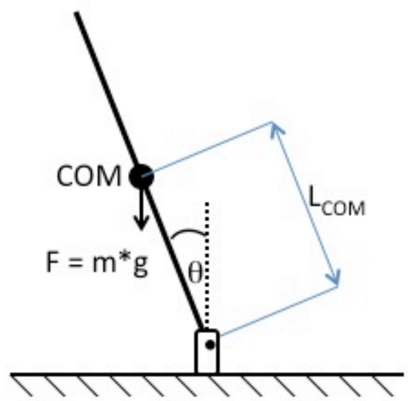
Effects of Inertia on Human Movement

Previous studies have investigated the effects of increased inertia on multiple aspects of human movement with varying results. Teunissen *et al.* [72] examined the effects of manipulating inertia independently of body weight on the metabolic cost of running in ten recreational runners and found that the metabolic cost of running with increased inertia was not substantially different than that of normal running. In this study, inertia was increased by adding mass to a belt around the participant's waist while unloading the additional weight of the added mass through a support harness applying a constant upward force. Grabowski *et al.* [73] also investigated the effects of increased inertia on metabolic cost using ten participants walking on a treadmill and the same methods of Teunissen *et al.* and found, contrary to Teunissen *et al.*, that participants had increased metabolic cost with increased inertia. Some of the differences in the results of these two studies may have occurred because Grabowski *et al.* increased inertia by increasing body mass by 25-50% and tested during walking while Teunissen *et al.* increased inertia by increasing body mass by 10-30% and tested during running. Therefore, the different amounts of added inertia or different locomotion types could be responsible for the contrasting results regarding the effect that increased inertia has on the metabolic cost of human movement. DeWitt *et al.* [74] found the effect of inertia on gait parameters to be dependent on locomotion type, lending further support to the idea that the effect of increased inertia depends on the movement task involved. Additionally, the effects of increased inertia on muscle function during locomotion were found to be dependent on the primary action of the muscle [75]. Therefore, inertia does have a role in the biomechanics of human movement but the specific effects of increased inertia may depend on the function or movement being performed.

Fundamental mechanics suggest that increased inertia may be detrimental to balance. As the body deviates away from an upright position, inertia makes it difficult to reverse the direction of the body because inertial forces resist changes in direction. When inertia is increased, increased strength may be necessary to overcome the increased inertial forces and return the body to an upright position.

Effects of Weight on Human Movement

Increased weight also likely contributes to the reduced balance ability seen in obese individuals. To further explore the potential effects of increased weight on standing balance, it is helpful to model the human body as an inverted pendulum (Fig. 1). Increasing an individual's weight increases the gravitational moment about the ankles, suggesting that obese individuals will need greater muscle torques to control the greater gravitational moments acting at the ankle.



$$\ddot{\theta}(t) = \frac{mgL_{COM}\sin(\theta(t))}{I}$$

Figure 1: Single segment inverted pendulum model. Model of a slender-rod inverted pendulum with equation for acceleration of an inverted pendulum. *m* = mass of pendulum, *g* = gravitational constant, *COM* = center of mass of pendulum, L_{COM} = distance from pendulum pivot to COM, θ = angle of pendulum from vertical, *t* = time, *I* = mass moment of inertia of pendulum about its axis of rotation, *F* = gravitational force exerted on pendulum.

In addition to observing the effects of inertia on the metabolic cost of running, Teunissen *et al.* [72] also examined the effects of reducing body weight on metabolic cost and found significant decreases in net metabolic rate with decreased body weight. In this case, reduced body weight was achieved through the use of a support harness applying a constant upward force without any additional mass added. Net metabolic rate also decreased with decreased body weight during walking [73]. The effects of increased weight on muscle function during locomotion were, similar to the effects of increased inertia on muscle function, found to be dependent on the primary action of the muscle. The gastrocnemius, which plays a large role in body support but not forward propulsion during locomotion, was more sensitive to changes in body weight than the soleus, which is more responsible for propulsion of the body during locomotion [75].

Studies on load carriage give insight into the combined effects of inertia and weight on balance. A study by Qu and Nussbaum [76] observed the effects of increasing the mass and height of a load worn around the abdomen on COP based sway variables. Increasing the mass of the load increases both inertia and weight simultaneously. Increasing the height of the load increases the height of the center of mass of the body and thus increases both inertia and weight of the body. Both increasing the mass of the load and increasing the height of the load led to increases in sway variables such as AP COP mean speed, root mean square COP position, and centroidal frequency. Two studies investigating backpack load carriage found similar results. Zultowski *et al.* [77] investigated the effect of increasing the mass of a load carried in a backpack on COP speed during standing and found increased COP speed with increased load. Another study involving backpack load carriage by Sako *et al.* [78] investigated the effect of load placement height and found that COP speed increased with increased load height. Increasing both inertia and weight, as seen in these load carrying studies, resulted in similar balance deficits as those seen in the obese.

Summary and Purpose

Obesity is an increasingly prevalent medical condition in today's society, particularly so because it puts individuals at an increased risk for injuries resulting from falls. A better understanding of human balance during standing can give insight into the increased fall risk in the obese. Compared to the numerous studies investigating the effects of perturbing sensory feedback on balance, relatively few studies have investigated the effects of perturbing dynamics on balance. Therefore, the purpose of this study was to investigate the effects of increased inertia and increased weight (two aspects of altered dynamics associated with obesity) on balance during quiet standing. Results from this study can provide insight into human balance, and could eventually lead to interventions to help improve balance (and reduce falls) among obese individuals.

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CHAPTER 3 – THE INDEPENDENT AND COMBINED EFFECTS OF INCREASED MASS MOMENT OF INERTIA AND INCREASED WEIGHT ON STANDING BALANCE

Introduction

Numerous studies have focused on understanding the contributions of sensory feedback to human balance. For example, Meyer *et al.* [1], Day *et al.* [2], and Day *et al.* [3] investigated the effects of perturbing somatosensory feedback, visual feedback, and vestibular feedback, respectively, on standing balance. This type of study usually involves impairing one type of sensory feedback and observing the (often detrimental) effects this produces on balance by measuring multiple sway variables. Typical results often include increased center of pressure (COP) speed, increased COP range, etc. Despite the high prevalence of studies related to sensory feedback, relatively few studies have focused on understanding the contributions of altered dynamics of the human body to balance. Altered dynamics can include any physical aspect of the body that would result in changes in the governing equations of motion of the body. Such changes could lead to alterations in either balance performance and/or the control of balance during standing. For example, changes in mass moment of inertia (inertia) and forces acting on the body due to the effects of gravity (weight) both result in changes in the governing equations of motion of human standing and thus can be considered altered dynamics. Altering the dynamics of the human body and observing the subsequent effects on postural sway, similar to the way in which studies alter sensory feedback that contributes to balance, can lead to further insight on human balance.

A real-world example of altered body dynamics influencing human balance occurs in obese individuals. Obesity is associated with both increased inertia and weight of the body, and is associated with increased postural sway during quiet standing [4-8]. However, there is also evidence of obesity altering sensory feedback [9-12]. As such, the increased postural sway reported among obese individuals is likely the net effect of both altered dynamics and sensory performance. While the sensory impairments associated with obesity are thought to be detrimental to standing balance, increases in inertia and weight also likely have their own influence on balance, and may affect individuals other than the obese. However, the contributions of increased inertia and increased weight to human balance are currently not well understood.

Fundamental mechanics suggest an ambiguous effect of increased inertia or weight on balance. Increased inertia may be detrimental to balance during quiet standing by requiring greater strength to control the natural oscillations of the body. As the body begins to sway away from an upright position, it will be more difficult to slow down and reverse the direction of the body in order to return to an upright position because increased inertia makes the body more resistant to changes in angular velocity. For the purposes of this study, increased weight was defined as an increase in the torque about the ankle due to the effects of gravity. Increased weight may be detrimental to balance because increased strength is needed to balance the increased gravitational moments. Additionally, increased weight is likely to affect proprioceptive feedback due to the increased pressure sensed by the plantar mechanoreceptors resulting from increased gravitational forces. If greater strength is not an issue, increased weight may have little effect on standing balance because the body can utilize the altered proprioceptive feedback to control for any alterations in body dynamics due to increased weight.

In an effort to better understand the effects of altering body dynamics on balance, the goal of the present study was to investigate the effects of increased inertia and increased weight on balance during quiet standing. This was done using a custom device that allowed separate manipulation of these two aspects of body dynamics. Improved understanding of human balance may lead to improvements in fall prevention in individuals with conditions such as obesity, where both sensory feedback and body dynamics are altered from normal.

Methods

Sixteen male adults participated in this study (age: 22.1 ± 1.7 yrs, BMI: 22.9 ± 2.0 kg/m^2, height: 174.9 ± 5.1 cm, mass: 70.2 ± 7.7 kg). Participants were excluded if they reported any musculoskeletal injury within the past three months. This research was approved by the Virginia Tech Institutional Review Board, and written informed consent was obtained from all participants prior to participation.

Balance measurements were taken under four experimental conditions: baseline without increased inertia or weight (B), increased inertia only (I), increased weight only (W), and increased inertia and weight (IW). The order of presentation of the conditions was randomized across participants.

A custom-built backboard was designed to allow inertia and weight of participants to be increased independently. The backboard was supported by the ground and pivoted about an axis aligned with the lateral malleoli of participants (Fig. 2). Thus, it limited movement to the anterior-posterior (AP) direction. Wooden boards placed under

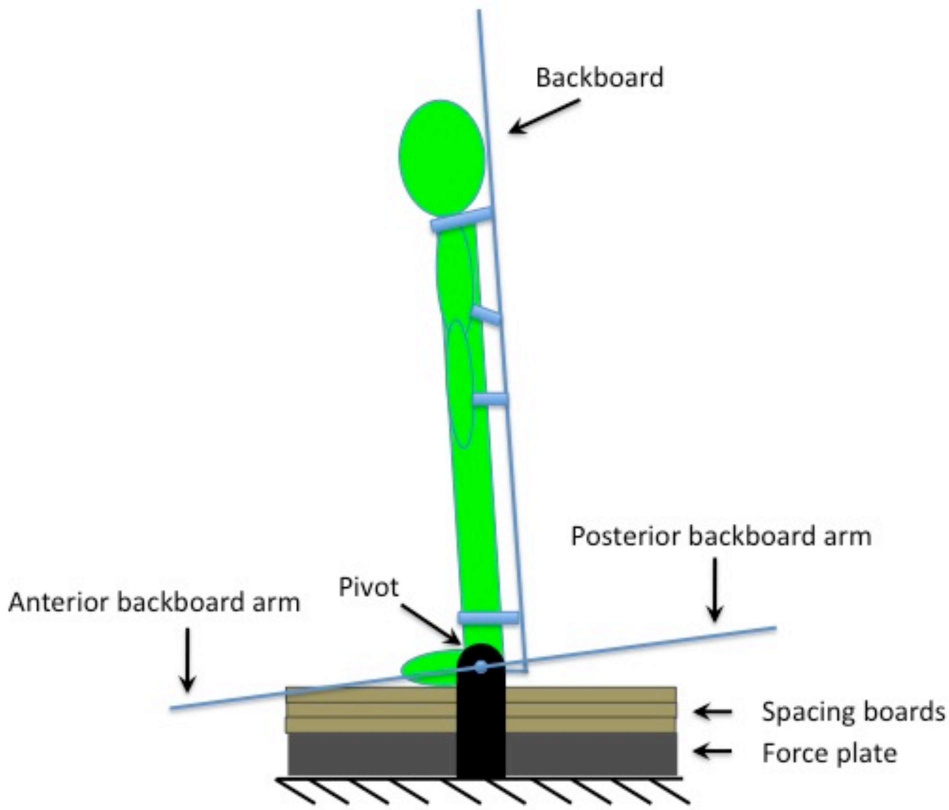

Figure 2: Backboard setup showing baseline (B) condition. The backboard and arms all rotate as one unit about the pivot point, aligned with the participant's lateral malleoli.

the participant's feet were used to align the participant's lateral malleoli with the pivot. Rigidly attached to the backboard were four "arms" (two on the participant's left and two on the participant's right) extending from the axis of rotation of the backboard. These arms rotated with the backboard, and were used to increase the participant's inertia and weight as described below.

To increase inertia of the backboard (and thus of the backboard/body system) without increasing weight, equal masses (plate weights used for resistance training) were placed on all four of the arms attached to the backboard (Fig. 3). These masses were all positioned an equal distance from the axis of rotation, thus they did not result in a net

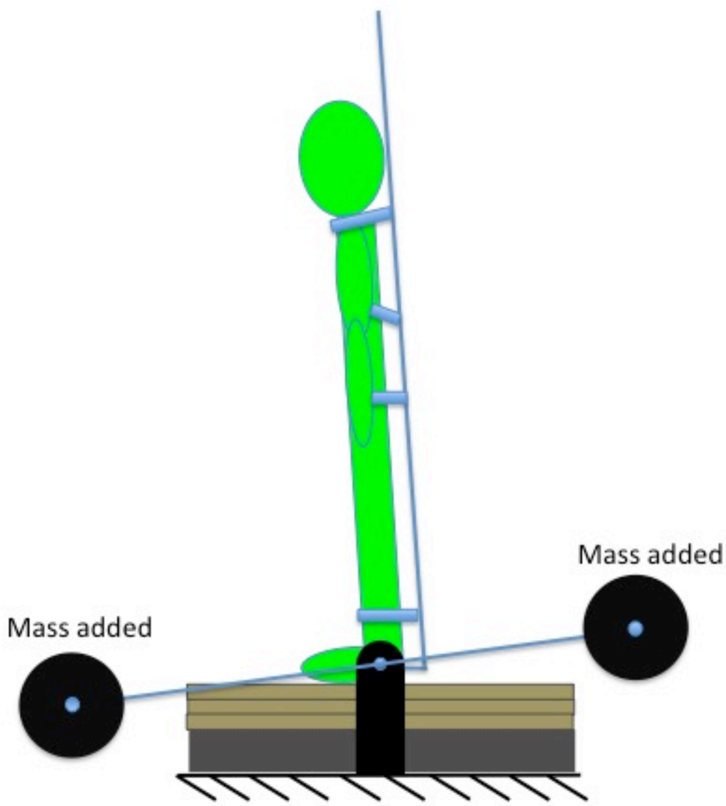

Figure 3: Backboard setup showing increased inertia (I) condition. Masses (plate weights used for resistance training) are added to the backboard arms.

moment about the axis of rotation. Additionally, participants were not required to support these added masses because the backboard and arms were supported by the ground. These masses only increased the mass moment of inertia about the axis of rotation of the backboard. Varying amounts of inertia could be added by adjusting either the amount of mass added to the arms or the distance between the masses and the axis of rotation.

To increase weight (i.e. moment due to gravity) of the backboard (and thus of the backboard/body system) without increasing inertia, a near constant anteriorly-directed force was applied to ends of the two posterior arms (Fig. 4). These forces caused a moment about the axis of rotation that tended to rotate the body/backboard system forward just as gravity tends to rotate the body/backboard system forward because the body COM is anterior to the lateral malleoli during quiet standing. The force was applied with stretched lightweight surgical tubing to avoid adding appreciable mass (and thus inertia) to the system. The change in length of the surgical tubing during testing was negligible, resulting in a near-constant force in the tubing. The moment about the axis of rotation due to this force could vary with backboard angle in the same way that the gravitational torque acting on participants varied with backboard angle. This was accomplished by adjusting the angle of the arms on the backboard such that they were perpendicular to a line between the participant's lateral malleoli and the combined backboard/body system COM. This resulted in the line of action of the applied force being through the middle of the axis of rotation of the backboard (and thus providing no moment about this axis) at the same angle of the backboard as when the backboard/body system COM was directly above the ankles (and thus providing no moment about the ankles). Varying amounts of weight could be added to the backboard/body system by adjusting the magnitude of the applied force. This was accomplished by changing the length of the surgical tubing using a winch positioned 5.2 meters in front of the backboard and measuring the magnitude of the applied force using an in-line load cell (Cooper Instruments and Systems, Warrenton, VA). The relatively large distance between the winch and the setup, compared to the minimal change in height of the tubing where connected to the posterior arms of the backboard resulted in negligible changes in the line of action of the force applied by the surgical tubing over the typical range of backboard angles seen during quiet standing. This meant that both the gravitational moment about an axis through the ankle joints due to the backboard/body COM, and the moment about the axis of rotation of the backboard due to the applied force, were both functions of $\sin(\alpha)$ where α is the angle from vertical to a line connecting the lateral malleoli and combined COM of the backboard/body system.

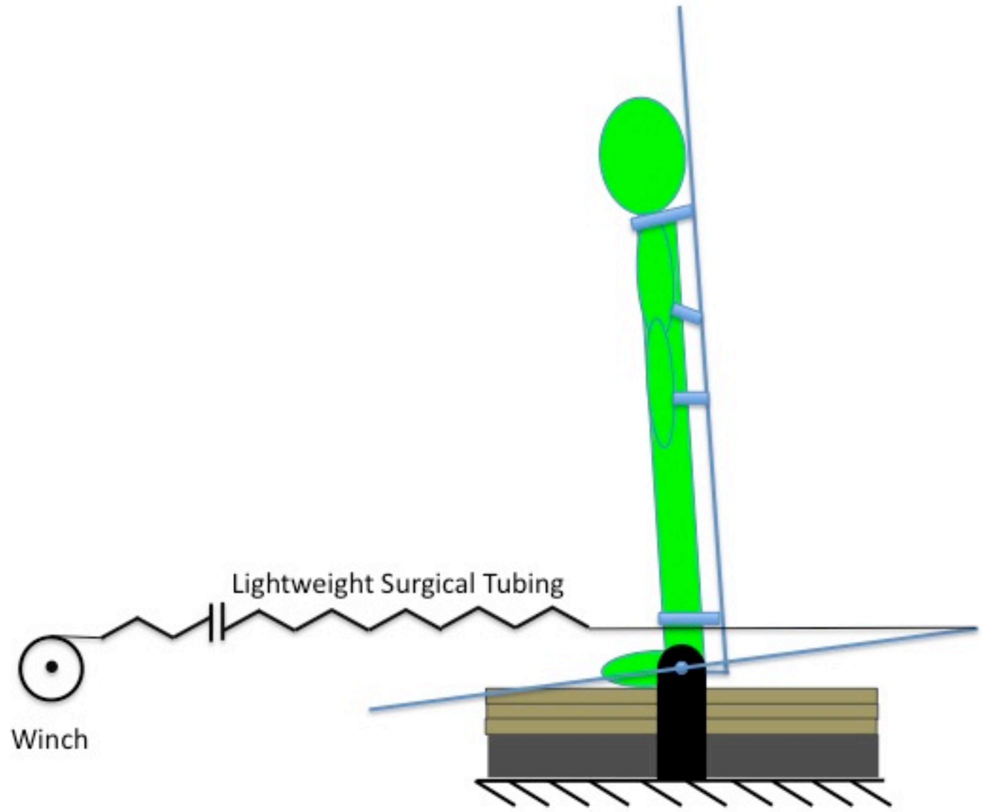

Figure 4: Backboard setup showing increased weight (W) condition. A force is applied by stretching lightweight surgical tubing across the room to a winch, causing a moment about the pivot.

The participant's superior-inferior COM position was determined using the reaction board method [13]. The participant's anterior-posterior (AP) COM position was determined knowing that the mean AP position of the COM is equal to the mean AP position of the center of pressure (COP) during quiet standing over an extended period of time [13]. During three 30-second quiet standing trials on a force plate (Bertec Corp., Columbus, OH) the mean angle of the backboard was measured with a linear potentiometer (Unimeasure, Corvallis, OR), and the mean AP position of the COP relative to the center of rotation of the backboard/body system was measured. The mean angle of the backboard was used to make a change of coordinate system and calculate the AP position of the backboard/body COM relative to the backboard.

A 30% increase in inertia and/or weight was induced for the I, W, and IW experimental conditions. This was an increase of 30% of the participants' initial body inertia and/or weight (and did not include the inertia and/or weight of the backboard). All participants initially had a BMI in the healthy range. Increasing weight by 30% would cause the participants to increase from a healthy BMI to the borderline between having an overweight BMI and having an obese BMI. Baseline inertia of participants was calculated using the relationship between the time period of a swinging pendulum and its inertia:

$$I = (T^2 * m * g * d_{COM}) / (4 \pi^2) \qquad \text{(Equation 1)}$$

where I is the inertia of the swing/body about the axis of rotation, T is the period, m is the combined mass of the swing/body, and d_{COM} is distance from the axis of rotation to the combined COM of the swing/body. Equation 1 comes from using a small angle approximation and assuming that there is little effect of friction on the pendulum's movement. Participants stood on a custom-built rigid wooden swing (Fig. 5) with their arms at their sides and head facing forward. The axis of rotation of the swing was an axis over the head of the participants that was parallel to an axis passing through both lateral malleoli. The trajectory of the swing following release from an initial angle of approximately 8° from vertical was measured using a Vicon 460 motion analysis system (Vicon, Lake Forest, CA) with two markers. Five trials were completed for each participant. The mean time period was determined by averaging the time between consecutive instances of zero angular velocity (of the angle formed by the Vicon markers) for each of the first five periods of each trial. Because the values for mass and distance to the COM used in Equation 1 were for the combined COM of the swing/body system, the inertia of the swing itself (77.9 kg*m^2 about the swing's pivot) was first subtracted, and then the parallel axis theorem was used to obtain the inertia of participants about a transverse axis through the lateral malleoli. The mass of the swing was 30.0 kg.

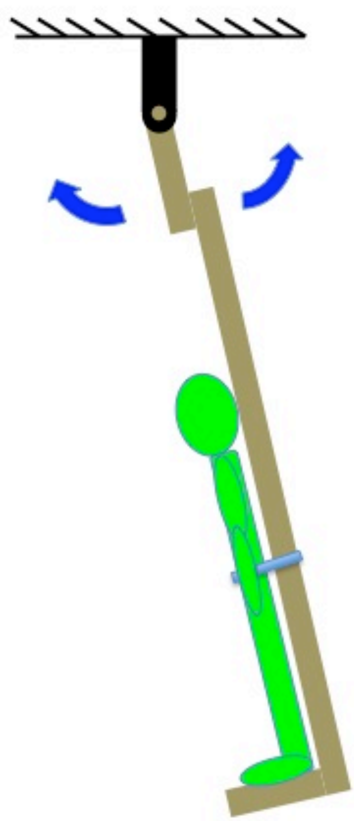

Figure 5: Rigid swing used to measure the participant's mass moment of inertia.

At the beginning of the experiment, height, mass, ankle height, foot length, and ankle-to-heel distance of the participant were measured. The reaction board method was then used to calculate the superior-inferior position of the body COM. The swing was used to calculate inertia of the participant, and quiet standing was performed to determine the anterior-posterior COM. Participants then performed quiet standing trials under each of the four experimental conditions (B, I, W, and IW). Participants were strapped to the backboard at the chest, mid-thigh, and mid-calf. They were instructed to stand as still as possible with their eyes closed (to force participants to rely more on their vestibular and proprioceptive feedback systems), arms at their sides, and head facing forward for 30 seconds. Forceplate data, backboard angle data, moment arm data (using a rotational potentiometer, Unimeasure, Corvallis, OR), and load cell data were sampled at 500 Hz. Five trials were performed under each experimental condition. All data were low-pass filtered at 10 Hz (4^{th}-order Butterworth zero-phase-lag filter). Electromyography of the left tibialis anterior was monitored during testing to ensure participants did not exhibit obvious increases in lower leg co-contraction above the level observed during quiet standing without the backboard. Participants wore athletic clothing without shoes during the entire testing session.

Balance was quantified by calculating measures of postural sway using both the COP and backboard angle. Specific measures included mean position, range of position, mean speed, and range of speed of both the COP and backboard angle. In addition, centroidal frequency of the COP position and the root mean square value of the difference between the COP and COM positions (COP-COM) [14] were calculated. The first trial of each experimental condition was considered to be a practice and removed from the statistical analysis. Therefore, trials 2 through 5 were used for further analysis. A two-way analysis of variance was conducted for each balance measurement using trial (2-5) and condition (B, I, W, IW) as independent variables. When significance was found for either independent variable or their interaction, post-hoc Tukey honestly significant difference tests were performed to determine differences between conditions. A statistical significance level of $p \leq 0.05$ was used for all analyses. Logarithmic transforms of the data were performed for range of COP speed, range of backboard angular position, range of backboard angular speed, and centroidal frequency to correct for non-normal distributions.

Results

No main effects of trial or trial × condition were found. As such, the results focus on the effects of condition.

The mean position of the COP differed across the four conditions (p=0.001), and was more anterior in both the W and IW conditions than in the I condition (Fig. 6a). The range of the COP position also differed across the four conditions (p<0.001), and was larger in both the W and IW conditions than in the B condition. The range of the COP was also larger in the W condition than in the I condition (Fig. 6b). The mean speed of the COP was higher in both the W and IW conditions than both the B and I conditions (p<0.001) (Fig. 6c). Range of COP speed did not differ across conditions (p=0.212) (Fig.

6d). Centroidal frequency of COP position was also affected by condition (p=0.007) and was higher in the I condition than in the W condition (Fig. 7).

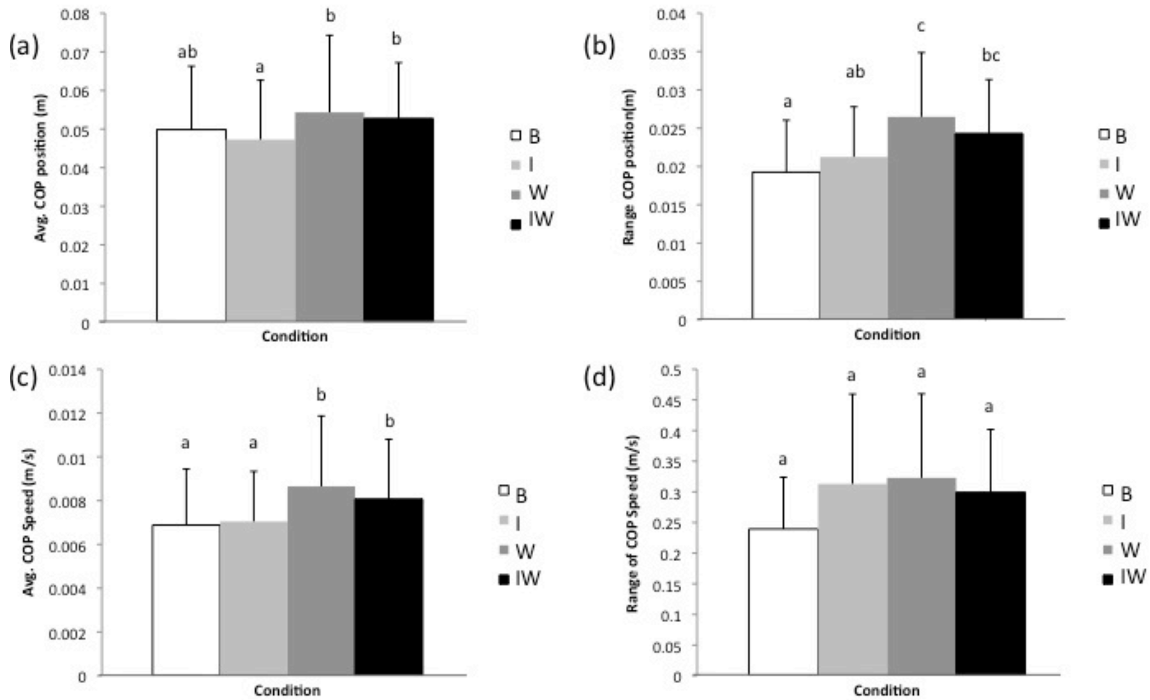

Figure 6: Anterior-posterior center of pressure position and speed variables. (a) Mean anterior position of the COP (b) range of COP position (c) mean speed of the COP and (d) range of speed of the COP. B – baseline, I – increased inertia only, W – increased weight only, IW – increased inertia and weight. Conditions not connected by the same letter are statistically different. Error bars represent standard deviations.

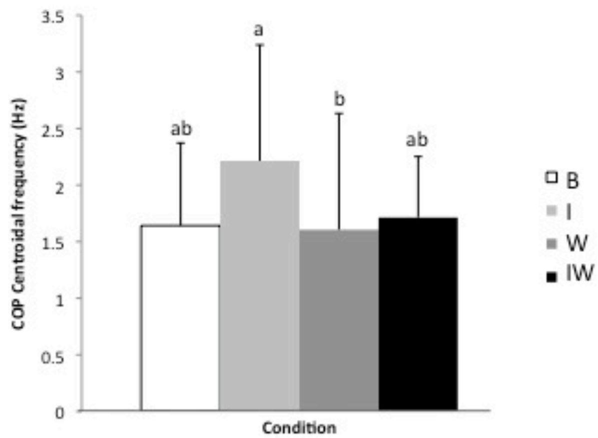

Figure 7: Centroidal frequency of the anterior-posterior center of pressure.

The mean angular position of the backboard did not differ across conditions (p=0.074) (Fig. 8a), but the range of backboard angular position did differ across conditions (p=0.032) with a larger range in the W condition than in the B condition (Fig. 8b). Mean angular speed of the backboard also differed across conditions (p<0.001), and was higher in the W condition than any of the other three conditions (Fig. 8c). The range of angular speed of the backboard was not different for different conditions (p=0.430) (Fig. 8d).

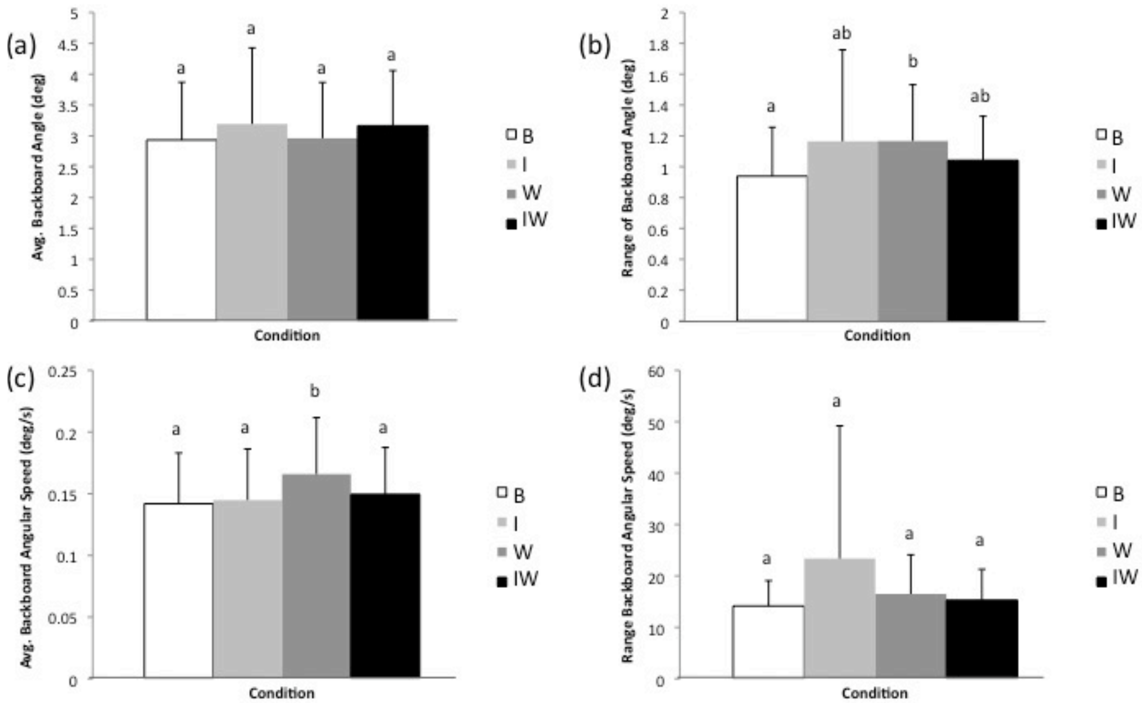

Figure 8: Backboard angular position and speed variables. (a) Mean angular position of the backboard (b) range of backboard angular position (c) mean angular speed of the backboard and (d) range of angular speed of the backboard.

COP-COM differed across conditions (p<0.001) with COP-COM of both the W and IW conditions higher than both the B and I conditions (Fig. 9).

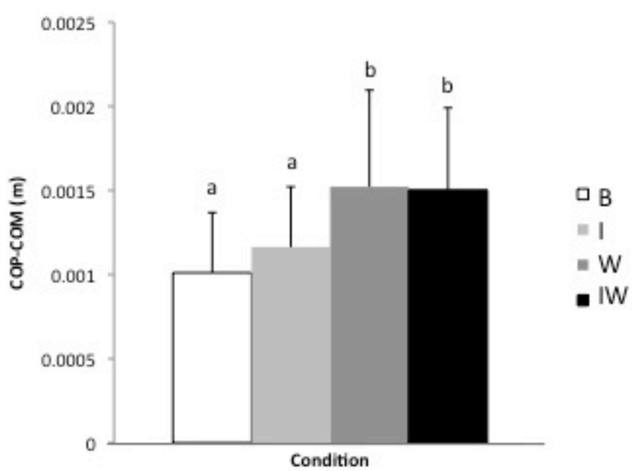

Figure 9: Root mean square of anterior-posterior center pressure minus center of mass.

Discussion

The goal of the present study was to investigate the effects of altering the dynamics of the human body – specifically increasing inertia or weight – on human standing balance. Multiple COP- and COM-based sway measures were affected by the altered dynamics.

The values of COP variables reported in the present study for the B condition are similar to those found in the literature [15]. Prieto *et al.* report a mean COP speed of 0.0089 m/s, a COP range of 0.018 m, and a centroidal frequency of the COP of 0.57 Hz for young healthy adults with eyes closed, which are comparable to the results of the present study under the B condition.

No differences were found between the I and B conditions for any of the sway measures investigated in this study. COP speed is typically interpreted as how hard the body has to work to maintain an upright posture (i.e. the number and frequency of corrections to body position) [16-18]. Therefore, the results of this study suggest that it was not any easier or more difficult to maintain balance with increased inertia. Additionally, ankle torque is proportional to COP position and thus the lack of difference in COP range between the I and B conditions may indicate that no additional ankle torque was needed to control balance during the I condition. Therefore, during standing it appears that increased inertia does not necessitate greater ankle torques be used to control increased inertial forces.

During the W condition, participants exhibited a 37.5% larger COP range and 25.7% larger mean COP speed, a 24.3% larger backboard angular range and 16.9% larger mean backboard angular speed, and a 50.5% larger COP-COM than in the B condition. The backboard angular range and speed reflect the degree to which the body, and hence the COM, moved. Increased backboard angular range and speed indicate the body was moving faster and over a larger distance. Thus it is logical that the difficulty level required to maintain upright posture increased, as reflected in the change in mean COP speed. Additionally, it appears that increased ankle torque was necessary to control balance during the W condition as evidenced by the increased range of COP in the W condition as compared to the B condition. Thus, increased weight may prove problematic for individuals with strength deficits.

COP-COM gives information about the error signal responsible for controlling COM position, and is directly related to the angular acceleration of the participant [13]. Thus, a larger value for COP-COM in the W condition means that larger angular accelerations were occurring because there was a greater error in the COP position controlling the movement of the COM. This is consistent with the other results for increased weight, notably the mean COP speed, in suggesting that it was more difficult to control the COM movement with increased weight.

When participants experienced both increased inertia and increased weight in the IW condition, they had a 26.2% larger range of COP position, 17.3% larger mean COP speed, and a 48.6% larger COP-COM than in the B condition. Similar to the W condition, a larger degree of corrections was required in order to maintain upright posture, as evidenced by the larger COP speed, and the participants used increased ankle torques, as evidenced by the larger range of COP position, in the IW condition as compared to the B condition. Additionally, because COP-COM was increased in the IW condition as compared to the B condition, there again was greater angular acceleration of the body than in the B condition. However, in the IW condition, the participants did not experience the larger COM movement as was seen in the increased W condition (large range of backboard angle and angular speed). Due to the lack of differences in COM movement in the IW versus the B condition, the W condition seems to cause the most extreme changes as compared to the B condition. Several results support the trend that adding increased inertia to the W condition (i.e., the IW condition) tended to mediate some of the extreme effects caused by added weight.

The data for the IW condition is consistent with that of three studies involving load carriage. Nussbaum & Qu [19] found increased AP COP speed with increased load worn in a belt placed at the participant's COM. In this study, inertia and weight were both increased by increasing the mass of the backpack load. Similarly, a study by Zultowski *et al.* [21], investigated the effects of backpack load on standing balance and found that COP speed increased with increasing load. Sako *et al.* [20] investigated the effects of backpack load placement on standing balance and found that placing a load higher on the back increases COP speed. Placing the load higher increases inertia and weight by altering the position of the center of mass of the backpack/body system. The results of the present study are consistent with these studies in indicating that COP speed is increased with increased inertia and weight.

In comparing the increased inertia only to the increased weight only condition, there was a difference in centroidal frequency of the COP position with increased inertia having a larger centroidal frequency. Frequency variables can sometimes provide insight into underlying neural or sensorimotor differences [22], and thus it is possible that, while neither the I or W conditions were different from the B condition in terms of sensory differences, there were differences in the amount or type of sensory information being relayed to the brain in the I versus W conditions. As mentioned previously, the increased weight may have increased mechanoreceptor sensation on the plantar surface, while the increased inertia could not (because the weight of the added masses is supported by the ground). The differences in plantar sensation between the two conditions may explain some of the difference in centroidal frequency.

There were several limitations to the present study. First, movement was limited to an ankle strategy. While using mainly ankle musculature to control AP balance during standing is common [23, 24], it is unclear how these results would generalize to other balance tasks not limited to an ankle strategy. Second, only male participants were used. This was done to ensure a more homogenous participant population.

In conclusion, altering the dynamics of the human body during quiet standing did have some detrimental effects on balance. Increasing weight by 30% led to increases in body movement (increased backboard angular range and speed) with greater error in the COP position (increased COP-COM), which required greater corrections to body position (increased COP speed) using increased ankle torques (increased range of COP position). While there were no effects of increasing inertia by 30% on balance, increasing both inertia and weight simultaneously also increased the error in COP position, necessitating greater corrections to body position using increased torques. However, increasing inertia and weight did not have the subsequent increases in body movement seen when only weight was increased. Thus, increased inertia may have some mediating action on the extreme effects caused by increased weight.

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CHAPTER 4 – SUMMARY AND CONCLUSIONS

The goal of the present study was to investigate the effects of altering the dynamics of the human body – specifically increasing inertia or weight – on human standing balance. This was done using a custom device that allowed separate manipulation of these two aspects of body dynamics. Multiple COP- and COM-based sway measures were affected by the altered dynamics. Increasing weight by 30% led to increases in body movement (increased backboard angular range and speed) with greater COM acceleration (increased COP-COM), which required greater corrections to body position (increased COP speed) using increased ankle torques (increased range of COP position). While there were no effects of increasing inertia by 30% on balance, increasing both inertia and weight simultaneously also increased the acceleration of the COM, necessitating greater corrections to body position using increased torques. However, increasing inertia and weight did not have the subsequent increases in body movement seen when only weight was increased. Thus, increased inertia may have some mediating action on the extreme effects caused by increased weight. This study gives insight into the effects of altered dynamics, namely increased inertia and weight, on standing balance. Improved understanding of human balance may lead to improvements in fall prevention in individuals with conditions such as obesity, where both sensory feedback and body dynamics are altered from normal.