A MODELING INVESTIGATION OF OBESITY AND BALANCE RECOVERY

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(ABSTRACT)

Obesity is associated with an increased risk of falls and subsequent injury. Previous studies have shown weight loss and strength training to be beneficial to balance, but knowing which is more beneficial will allow researchers to design interventions to maximize the benefits in terms of balance and reducing risk of falls. Therefore, the purpose of the first study was to evaluate the effects of weight loss and strength training on balance recovery using a combination of laboratory experiments and mathematical modeling. Nine male subjects with BMI 30.1 to 36.9 kg/m^2 were released from a forward lean and attempted to recover balance using an ankle strategy. Lean angle was increased until subjects required a step or hip flexion to recover balance. The maximum lean angle, θ_{max}, was used as the measure of balance recovery capability. Experimental data were used as inputs to an inverted pendulum model of balance recovery. Multiple simulations were used to determine the effects of strength (maximum ankle torque and ankle torque generation rate) and weight loss on θ_{max}. Changes in weight and strength were linearly related to changes in θ_{max}. A 6.6 ± 0.4% decrease in weight or 6.9 ± 0.9% increase in strength were estimated as required to improve (increase) θ_{max} by 1 degree. Based on these results, balance recovery using an ankle strategy can improve with either reductions in weight or increases in strength. In addition, weight loss may be a more effective intervention than strength gain at improving balance recovery capability. The purpose of the second study was to quantify changes in body segment inertial parameters (BSIPs) with weight loss. These data were needed to alter BSIPs in the first study to mimic changes with weight loss. Both before and after weight loss, magnetic resonance imaging scans were acquired along the length of the body and were used to calculate segment masses, COM positions, and radii of gyration. A number of significant changes in BSIPs occurred with weight loss.

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CHAPTER 1- INTRODUCTION

OVERWEIGHT AND OBESITY: A GLOBAL HEALTH CONCERN

Obesity is a major health concern in the United States and numerous countries across the world [1]. Over 1 billion people worldwide are considered overweight and of those, 300 million are considered obese [2]. In the United States and the United Kingdom, nearly two-thirds of adults are considered overweight or obese [3, 4]. The high prevalence of obesity is continually increasing. In the US between 1980 and 2002, obesity prevalence doubled among adults and overweight prevalence tripled among children [5]. More recently, the prevalence of obesity among US adult men increased from 27.5% in 1999 to 31.1% in 2004 [5].

Obesity is highly associated with numerous health conditions. For example, obesity is highly associated with hypertension, type 2 diabetes mellitus, cardiovascular disease, osteoarthritis (OA), and respiratory disease [1, 6]. In addition, the obese tend to have higher levels of functional limitation than the non-obese [7]. Overall, approximately 6%, or $92 billion, of total annual medical expenditures in the United States can be attributed to obesity [8].

OBESITY, FALLS, AND THE MUSCULOSKELETAL SYSTEM

In addition to its association with numerous health concerns, obesity is also associated with an increased risk of falls. Fjeldstad et al. [9] reported obese subjects to have a higher prevalence of falls (27% vs 15%) and ambulatory stumbling (32% vs 14%), or a loss of balance that was restored without falling, than their non-obese counterparts. Additionally, the overweight and obese are more likely to experience falls on the same level and falls to a lower level [10] as well as multiple falls [11]. Falls by themselves are not problematic, but an increased risk of falls increases the risk for injury. Falls were identified as the most common cause of injuries in the obese (~36% of all injuries) and the cause of a higher proportion of injury-related hospitalization in the obese compared to the non-obese [12]. Similarly, Finkelstein et al. [13] found the overweight and obese to have greater odds of sustaining a medically treated injury and even more likely to sustain a fall injury than the non-obese. Approximately 6% of falls result in a major injury such as fracture or dislocation [14]. As such, fractures have been reported to be the most common injury among the obese and obesity also increases the likeliness of sustaining a sprain/strain or dislocation [12].

Although the mechanism by which obesity contributes to falls has not been determined, the increased risk of falls in the obese is most likely caused by the negative impact of obesity on musculoskeletal function. Obesity has been highly associated with impaired musculoskeletal function, particularly in the lower limbs [15]. It is thought that excess weight increases the stress within the bones, joints, and soft tissues, resulting in impaired musculoskeletal function such as abnormal mechanics [16, 17] and posture [18, 19]. These impairments, such as impaired balance, gait, strength, sensory function, and neuromuscular function, have been identified as strong risk factors for falls [20].

Obesity causes alterations in gait that are associated with an increased risk of falls. Several studies have found preferred walking speed, step length, and step frequency to be significantly lower in the obese compared to the non-obese [21-23]. Additionally, the obese have a longer stance phase [21, 22] and greater period of double support [22]. Reduced walking speed, cadence, and step length as well as increased stance duration have been identified as strong risk factors for falls [20]. Spyropoulos et al. [22] have suggested that obesity requires an individual to walk slowly, take smaller strides, and remain in double support longer in order to maintain balance. Deviations from the obese gait pattern would result in instability and loss of balance. Spyropoulos et al. [22] also found the obese to have a larger step width during walking, which provides a wider base of support for balance. DeVita and Hortobagyi [21] found that obese adults tend to have a more erect posture while walking at a standard speed, compared to non-obese adults, as a result of reduced knee and hip flexion. It is possible this posture provides stability in the obese by counteracting an anterior displacement of the center of mass (COM) from the longitudinal axis of the body associated with obesity, reducing the amount of corrective torque needed to maintain balance.

The obese have also been shown to have impaired muscle strength, a strongly rated risk factor for falls [20]. More specifically, knee and ankle weakness have been identified as factors in poor balance and indicate a greater risk of falls [24, 25]. Although obese adults can generate higher absolute strength and power with the lower extremity, strength and power are significantly lower in obese adults when normalized to body weight. For example, Lafortuna et al. [26] investigated the effects of obesity on strength and power of the lower limbs muscles involved with anti-gravitational movements (quadriceps, gluteus, gastrocnemius, and soleus). Absolute lower limb strength was significantly higher in the obese subjects. However, during the push phase of a jump, the obese subjects had similar absolute lower limb power output and decreased lower limb output per unit body weight compared to the non-obese subjects. Other studies have also shown absolute knee strength to be higher and knee strength per body weight to be lower in the obese compared to the non-obese [27, 28]. These findings imply that although a greater absolute force is used by the obese to perform daily activities, their musculoskeletal system is impaired because they cannot produce equivalent force per weight compared to the non-obese. As such, when the obese are perturbed from a stable posture, they are unable to generate adequate force to recover from the perturbation and are likely to fall.

Increased weight bearing forces with obesity have also been shown to impair foot structure and function [29, 30], causing impaired balance and potentially increased risk of falls. In particular, plantar heel pain, characterized by a pain involving the insertion of the plantar fascia into the calcaneus, is five times more likely to develop in the obese than the non-obese [31]. Pain in the feet has been shown to increase postural oscillations, indicating an unstable posture and a decrease in balance [32]. Studies have considered the effects of obesity on foot structure to be a pathological source for heel pain. For example, contact area has been found to be significantly higher in the obese [33, 34], suggesting that obesity may inflict structural dysfunction such as a collapse of the longitudinal arch of the foot. However, Messier and Pittala found that arch structure was not able to discriminate between subjects with and without heel pain [35]. These conflicting reports indicate that obesity influences other factors in the development of heel pain. For instance, Ozdemir et al. [36] found obesity to be associated with a thinner heel fat pad, indicating higher susceptibility to heel pain and higher potential for falls. In contrast, Nass et al. [37] found that heel pad thickness increased as a function of BMI. It is also possible that increases in foot pressure alter balance in the obese. Foot mechanoreceptors and cutaneous sensation are contributors to balance control [38, 39]. Therefore, greater pressure and contact area may lead to reduced sensitivity of the foot mechanoreceptors and, consequently, impaired balance. Nass et al. [37] found plantar peak pressures to be significantly higher at the heel for the overweight group. Hennig et al. [40] found static plantar pressures to be minimally dependent on body weight in 111 non-obese adults, but concluded that changes in foot structure to increase foot dimensions allowed a redistribution of plantar loads from areas of high pressure to areas of low pressure. It was hypothesized that the obese will have higher plantar pressures under select portions of the foot when compared to the non-obese. As such, the mid foot and fore foot plantar pressures were found to be significantly higher with obesity [34, 41, 42] and were reduced with weight loss [43]. Though the exact mechanism remains unclear, a combination of reduced sensitivity and heel pain may compound to impair balance and increase the risk of falls in the obese.

Effects of Obesity on Balance and Balance Recovery

Epidemiological evidence linking obesity with an increased risk of falls suggests that obesity negatively effects balance. In fact, numerous studies have investigated the effects of obesity on postural sway during quiet standing.

Several studies have investigated the effects of obesity on balance in young populations [19, 44-46]. Goulding et al. [45] used several clinical balance tests, including the Bruininks-Oseretsky test, to observe static and dynamic balance in overweight and lean boys of age 10 to 21 years. The Bruininks-Oseretsky test of balance consisted of three static balance tasks (standing on one leg on the floor, standing on one leg on balance beam with and without eyes open) and five dynamic balance tests (walking on a line and balance beam, walking heel-to-toe on a line and balance beam, and stepping over a stick on the balance beam). Their results showed lower Bruininks-Oseretsky test scores in overweight boys. Balance can also be quantified using the center of pressure (COP) and can be defined as the neuromuscular response to the imbalances of the body's COM [47]. McGraw et al. [46] observed obese and non-obese boys of age 8 to 10 years during quiet standing with a normal (heels together) and tandum (dominant foot forward, heel-to-toe) stance and several visual conditions (full vision, dark, and visual conflict). Obese boys had greater maximum COP displacement, root mean square COP, and sway areas than non-obese boys in the anterior-posterior and medial-lateral direction. These results suggest that obese children have poorer balance than non-obese children, particularly in the medial-lateral direction.

In contrast, Bernard et al. [19] found obese adolescents of age 13 to 17 to have a greater COP path length during quiet standing only when standing on a foam surface, a more challenging balance task than standing on a hard surface. Similarly, Colne et al.[44] found no differences in COP displacement in the anterior-posterior direction between obese and non-obese adolescents (average age of 17 years) while standing on a hard surface. Balance was also assessed during a forward leaning posture in which subjects were to rotate forward about the ankles as far as possible without lifting the heels or toes from the ground. During a forward lean, obese adolescents had decreased COP displacements and similar sway areas compared to non-obese adolescents. These findings suggest that more challenging balance tasks are needed to observe differences in balance during quiet standing between obese and non-obese adolescents. However, measurements during a sustained forward lean were taken at each subject's maximum lean angle and these angles were not reported. It is possible that the smaller COP displacements in obese adolescents compared to lean adolescents could be due to a smaller maximum lean angle in the obese.

The effect of obesity on postural sway has also been investigated in adults. Some studies have found an association of adult obesity and postural sway parameters such as mean COP peak, the time instants in which the COP is relatively stable, and mean COP distance, the distance between the peaks. For example, Hue et al. [48] found that as mass increased, mean COP distance increased, corresponding to an increase in the distance between stable regions, and mean COP peak decreased, corresponding to an increase in time spent in a stable region. Additionally, mean COP speed increased as mass increased. These results suggest that an increase in body weight causes an individual to be less sensitive in the regulation of body sway oscillations. Other studies have reported similar results of increased mean COP speed with obesity [49, 50], also indicating that the obese have decreased stability and increased risk of falling. In contrast to younger populations, several studies have shown that obesity does not affect the anterior-posterior displacement of the COP in adults [34, 51]. It is possible that obesity will affect selection of motor strategies for balance [46, 52]. For example, obese individuals may lean backwards to compensate for an anterior displacement of COM. It is also possible that falling in obese subjects may be more related to dynamic measures of balance and require more demanding balance recovery tasks to ascertain differences between the obese and non-obese.

Only a small number of studies have investigated the effects of obesity on balance recovery after a postural perturbation. Miller et al. [53] investigated the effects of obesity on balance in response to small postural perturbations. A ballistic pendulum was used to apply small anteriorly-directed perturbations just inferior to the scapula. These perturbations were small enough so that no step was required to maintain balance. Contrary to expectations, obese subjects exhibited reduced peak COP displacement and velocity, suggesting that the obese subjects had better balance than the non-obese subjects. However, when responses were compared between groups at perturbations normalized to body weight, there were no differences in sway parameters between the obese and non-obese subjects. Other studies have investigated balance recovery using non-obese subjects wearing external weight to simulate obesity. Li and Aruin [54] conducted a perturbation study in which subjects stood with their arms extended forward and a 5-lb load place 10 cm above the subjects' hands. Subjects were required to catch the load while wearing external weight to simulate obesity. This study found that as mass increased, COP displacement increased. Similarly, Ledin and Odkvist [55] applied perturbations to non-obese subjects with external weight. Randomized perturbed posturography, or a randomly moving support surface, was used to apply three levels of perturbations in the anterior-posterior direction: no platform translation, low amplitude translation (2cm peak to peak, RMS amplitude 1.3cm), and high amplitude translation (4cm peak to peak, RMS amplitude 2.6cm). This study found that sway area increased with increased mass. However, it is possible that adaptations occur if the increased mass was worn day to day as in the obese. Experiments using non-obese subjects with external weight may not account for such adaptations. Berrigan et al. [56] examined the effect of a goal-oriented movement on postural stability in obese and non-obese subjects. Subjects, in a standing posture, were to point to a target as fast and as precisely as possible after an auditory signal. This rapid acceleration of the arm can be considered a small postural perturbation due to the momentum the movement exerts on the body. Obese subjects were found to have greater COP speed and displacement than their non-obese counterparts. During this study, the obese subjects moved their whole body forward while aiming at the target, as can be seen in a greater COP displacement. The authors suggested that by moving their COM closer to the target, the obese subjects were better positioned for correcting hand movements.

However, a forward displacement of the COM can result in an unstable posture due to a larger corrective torque needed to maintain balance.

INTERVENTIONS TO IMPROVE BALANCE IN THE OBESE

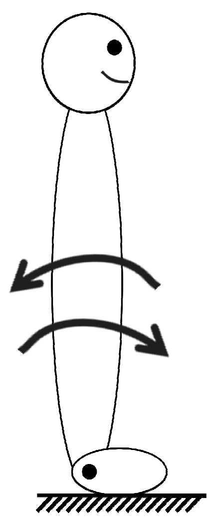
Differences in postural sway between obese and non-obese individuals suggest that balance may improve with weight loss. Teasdale et al. [50] imposed a hypocaloric diet and bariatric surgery on obese and morbidly obese subjects, respectively. Mean COP speed decreased linearly with weight-loss in both groups. Additionally, there were no differences in mean COP speed between the non-obese, obese, and morbidly obese after weight loss. Decreases in COP displacement in the anterior-posterior and medial-lateral directions were also seen with weight loss. These results indicate that balance improves with weight loss, and that improvement is linearly related to the magnitude of weight loss. Maffiuletti et al. [57] also investigated the effects of weight loss on balance during single limb stance. Morbidly obese subjects participated in a body weight reduction (BWR) program that included an energy-restricted diet, moderate exercise, and nutritional education. A portion of the morbidly obese subjects also received balance training in addition to the BWR program. Balance training consisted of repeated exposures to the balance task, i.e. single limb stance. It was found that trunk sway decreased and time of balance maintenance increased with weight-loss. However, these changes were found only with the combination of BWR and balance training. These studies suggest that weight loss interventions improve balance, but a combination of weight loss and balance training provides an even greater improvement.

Balance recovery after a postural perturbation has been shown to improve with increased strength [58, 59] and strength training has been commonly used as a fall-prevention intervention [60]. Lower extremity strength, specifically at the ankle and knee, can make a significant contribution to balance recovery [25]. However, lower limb function in the obese is impaired and not sufficient for balance recovery due to reduced lower limb strength and power per body weight compared to the non-obese. Therefore, improving lower limb strength should allow adequate force production to recover balance from a perturbation and reduce risk of falls. Although strength training has been encouraged as a fall prevention intervention in the obese

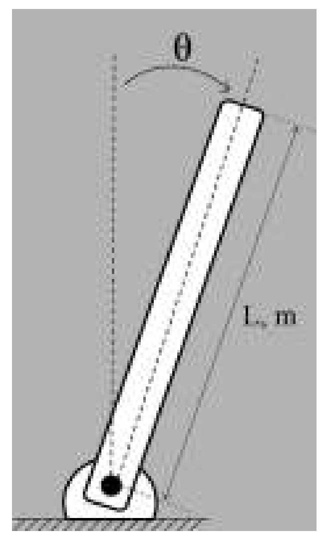
[61], no studies have investigated the effects of strength training on balance recovery in the obese.

INVERTED PENDULUM MODELS OF BALANCE RECOVERY

Mathematical modeling of a balance recovery task allows researchers to investigate multiple variables associated with the task that could otherwise not be manipulated independently in human subjects. Depending on the balance recovery task the human body can be modeled in many different ways. An inverted pendulum model can be used if the body segments remain closely aligned and movement occurs mostly at the ankle. For example, anterior-posterior sway during quiet standing has been modeled as an inverted pendulum (Figure 1).



(a)



(b)

Figure 1a. Anterior-posterior sway of a human subject during quiet standing.

Figure 1b. Inverted pendulum model.

The equation of motion for the inverted pendulum is thus:

$$I\ddot{\theta}(t) = \frac{mgL}{2}\sin[\theta(t)] \tag{1}$$

where I is the moment of inertia of the rod about the pivot point, θ is the angle measured from upright position to the centerline of the rod, g is the gravity constant, and L and m are the height and mass of the subject, respectively. Consider a slightly more challenging task such as balance recovery from a forward lean. One can restrict the subject to recover balance by only contracting the muscles spanning the ankle, a technique called the "ankle strategy." For such a task, a torque applied about the pivot, or in this case, the ankle, can account for the muscle contraction across the ankle. The inverted pendulum equation of motion then becomes:

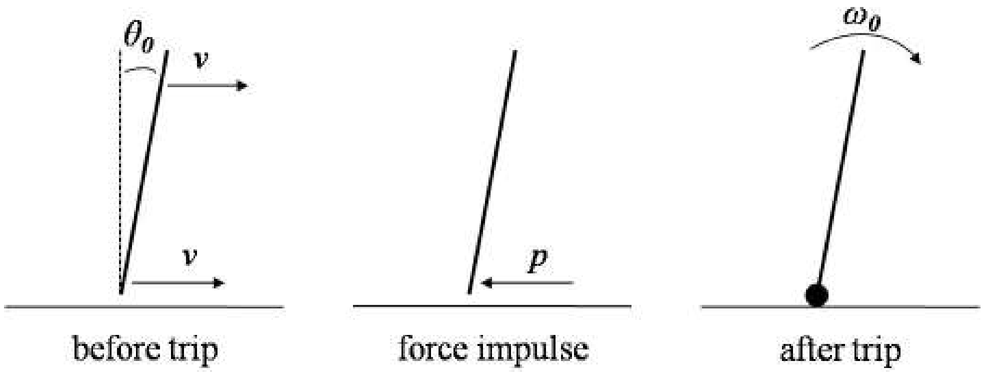
$$I\ddot{\theta}(t) = \frac{mgL}{2}\sin[\theta(t)] - T_a(t) \tag{2}$$

where T_a is the torque generated about the ankle.

Inverted pendulum models have been used in previous balance recovery studies but there has only been one study, to our knowledge, that utilized an inverted pendulum model to investigate the effects of obesity on balance [58]. However, models used to investigate balance recovery in other populations can provide valuable information to model development, analysis, and validation.

As an example of a balance recovery study that used an inverted pendulum model, Robinovitch et al. [59] investigated the effects of ankle strength and ankle torque development rate on balance recovery. During the experiment, subjects were leaned forward and instructed to hold the body straight and recover balance by using the ankle strategy. Subjects were then modeled as an inverted pendulum (Figure 1). A torque actuator was used to simulate the torque generated by the ankle muscles to recover balance. A linear approximation of the time history of torque during the experiment was used to model the torque actuator. Model validation was completed by comparing predicted maximum lean angle, θ_{max}, to the experimental mean. The model predicted θ_{max} values within one standard deviation of the experimental mean, leading the investigators to conclude that there was generally good agreement between the experimental and mathematical predictions. Properties of the torque history were manipulated within the model to determine the effects of isolated and combined variations of strength, such as maximum torque, and torque generation rate. Such properties are difficult, if not impossible, to manipulate independently in human subjects [62]. It was found that if ankle torque remains constant or declines at a small rate after reaching the maximum torque, θ_{max} always increases with increasing torque generation rate. These results imply that recovery limits depend on the capacity to quickly generate and maintain high magnitudes of ankle torque.

An inverted pendulum model has also been used to investigate the independent and interactive effects of walking velocity and response time on balance recovery from a forward trip in older adults [63]. Relationships between walking velocity and response time cannot be determined from human experiments due to adaptations and learning effects. With a mathematical model, one can manipulate these variables in numerous combinations and obtain an unbiased response. The subject was modeled as a rigid rod at a forward tilt angle, θ_0, and forward velocity, v prior to the trip (Figure 2a). The trip was simulated with a force impulse, p, at the lower end of the rigid rod (Figure 2b). The authors then assumed, after the trip, the subject rotated about a fixed axis at the stance foot ankle and the subject was modeled as an inverted pendulum (Figure 2c). Inertial parameters of the subject, initial orientation (θ_0), initial velocity (v), and the force impulse (p) were used to determine the initial angular velocity, ω_0, of the inverted pendulum after the trip. As a result, only walking velocity, body height, and initial angle were required for the inverted pendulum model to predict movement after the trip. Model validation was completed by examining the relationship between measured tilt angle and predicted tilt angle. Predicted tilt angles and measured tilt angles were highly correlated (Pearson product-moment correlation r=0.930), indicating that the model was a good predictor of the movement after the trip. Response time and walking speed were then manipulated and the investigators concluded that variations in response time are more important in determining the success of recovery than variations in walking speed.



(a) **(b)** **(c)**

Figure 2. van den Bogert et al. [63] inverted pendulum model of a tripping event.
Image is property of A.J. van den Bogert

Corbeil et al. [58] used an inverted pendulum model to investigate the effects of obesity on the stabilizing torque at the ankle joints. In particular, the study examined the effects of increased mass and an anterior displacement of the body COM on the stabilizing torque. An obese and lean "humanoid", or the inverted pendulum model of the subject, were developed using Hanavan's model [64]. Postural perturbations were simulated by imposing a sudden angular velocity on the humanoid. To stabilize the humanoid, both passive (stiffness and viscosity of the tissues) and active components (muscle contraction) of ankle torque were included in the model. Three torque parameters (ankle torque onset time, time to peak torque, and peak muscular ankle torque) were manipulated within the model. It was found that a nonlinear increase of torque was needed to stabilize the obese humanoid when the motor response had delayed temporal parameters. This effect was more pronounced with anterior displacements of the COM.

SUMMARY AND PURPOSE

Obesity is a growing health concern in the United States and around the world. Along with numerous medical conditions, evidence suggests that obesity also increases the risk of falls and subsequent injury.

Both weight loss and strength training can potentially improve balance and reduce the risk of falls. Previous studies have suggested that obesity impairs balance during quiet standing and balance recovery from a perturbation. Weight loss interventions can be used to improve balance [50, 57]. It has also been implied that an increase in corrective torque, such as through strength training, can improve balance [58, 59]. Though both appear to be beneficial, knowing which is more beneficial will allow researchers to design interventions to maximize the benefits in terms of balance and reducing risk of falls.

Inverted pendulum models such as those presented by Bogert et al.[63], Robinovitch et al.[59], and Corbeil et al.[58] can be utilized to examine the effects of obesity interventions on balance recovery. Mathematical modeling allows one to test the effects of various interventions on a shorter time scale. Studies using human subjects to investigate the effects of weight-loss on balance have used various programs that require several months [50, 65]. For example, Teasdale et al. [50] imposed a caloric restriction until subjects maintained a stable body weight for four consecutive weeks. Treatment duration ranged from 15 to 47 weeks. Mathematical modeling also allows one to investigate the effects of several variables on a given task. From these results, one can determine the most promising interventions to implement on human subjects for a longitudinal study. Therefore, the goal of this study (Chapter 2) was to investigate the effects of weight loss and strength training on balance recovery using an ankle strategy.

To accurately simulate a weight loss intervention, guidelines are needed to describe how body segment inertial parameters (BSIPs) change with weight loss. To our knowledge, such guidelines do not currently exist but are needed to realistically modify BSIPs within a mathematical model. Therefore, the goal of the second study (Chapter 3) was to quantify changes in BSIPs with weight loss.

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CHAPTER 2 – A MODELING INVESTIGATION OF OBESITY AND BALANCE RECOVERY

INTRODUCTION

Obesity is a major health concern in the United States and around the world [1]. Over 60% of adult men in the US and over one billion people worldwide are considered overweight or obese [2, 3]. Not only is the prevalence high, but it is continually increasing. Between 1980 and 2002, the prevalence of obesity among US adults doubled [4]. One of the many concerns with this high prevalence of obesity is its association with an increased risk of falls. Fjeldstad et al. [5] reported obese subjects to have a higher prevalence of falls (27% vs 15%) and ambulatory stumbling (32% vs 14%) than their non-obese counterparts. An increased risk of falls increases the risk for injury. Falls were identified as the most common cause of injuries in the obese (~36% of all injuries) and the cause of a higher proportion of injury-related hospitalization in the obese compared to the non-obese [6].

Weight loss has the potential to improve balance and reduce the risk of falls in the obese. Hue et al. [7] reported an increase in mean center of pressure (COP) speed, an index of the activity required to maintain balance, as body weight increased. These results suggest that an increase in body weight causes an individual to be less sensitive to body sway oscillations and require more activity to maintain a stable posture. In an intervention study, Teasdale et al. [8] investigated the effects of weight loss on balance in obese and morbidly obese subjects. Mean COP speed decreased linearly with weight-loss in both groups. Additionally, there were no differences in mean COP speed between the either group after weight loss and non-obese controls. In another intervention study, Maffiuletti et al. [9] investigated the effects of weight loss on balance during single limb stance. Morbidly obese subjects participated in a body weight reduction program that included an energy-restricted diet, moderate exercise, and nutritional education. Trunk sway, defined as the deviation of the trunk from the medial-lateral axis, decreased and time of balance maintenance, defined as the longest period a subject could stand without help, increased with weight-loss. These studies indicate that balance during quiet standing is influenced by obesity and that balance can be improved with weight loss.

Strength training also has the potential to reduce risks of falls in the obese [10]. Lower extremity strength, specifically at the ankle and knee, can make a significant contribution to balance recovery [11]. Balance recovery after a postural perturbation has also been shown to improve with increased strength [12, 13]. For example, Robinovitch et al. [13] found that recovery limits depend on the capacity to quickly generate and maintain high magnitudes of ankle torque. Although obese adults exhibit increased strength and power producing capability compared to non-obese, strength and power are significantly lower in obese adults when normalized to body weight [14-16]. As such, improving lower limb strength would increase strength relative to body weight.

While evidence suggests both weight loss and strength training can improve balance and reduce the risk of falls in the obese, knowing which is more beneficial will allow researchers to design interventions to maximize the benefits in terms of balance and reducing risk of falls. Therefore, the goal of this study was to investigate the effects of weight loss and strength training on balance recovery using an ankle strategy. Manipulating weight loss and strength independently in human subjects would be difficult and time-consuming. As such, we employed a combination of experimental testing and forward dynamic simulations to achieve our goal. The selection of balance recovery using an ankle strategy as our task of interest was guided by the desire to investigate a dynamic task in the context of fall prevention that could be modeled relatively easily.

METHODS

Nine male subjects aged 23.3 ± 4.5 years (mean ± standard deviation) were used in this study. Subjects had height of 178.9 ± 6.2 cm, mass of 107.2 ± 10.0 kg, and BMI of 33.5 ± 2.3 kg/m^2. The study was approved by the Virginia Tech Institutional Review Board, and written consent was obtained from all participants.

Experimental testing was performed to determine inputs to the mathematical model of balance recovery using an ankle strategy. While harnessed at the waist, subjects were leaned forward with their arms held behind their back and body segments aligned (Figure 1). In this position, subjects were instructed to keep their ankles relaxed and recover balance upon release of the harness rope using an ankle strategy (defined as contraction of only the muscles spanning the ankle and returning to an upright posture while keeping the body straight and without stepping). The body angle relative to vertical prior to release was defined as the lean angle (θ). Upon release, the subjects' heels were allowed to rise from the ground and only slight heel raise was used during most trials. The lean angle was systematically increased by 0.5° until the subject failed three times at a given angle. A failed recovery was defined as stepping or bending at the hip to recover balance. The maximum lean angle (θ_{max}) was used to quantify balance recovery capability.

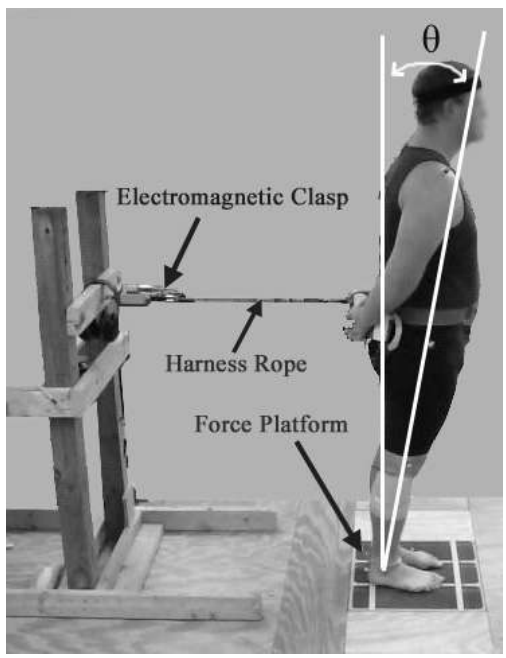


Figure 1. Balance recovery experiment.
Photo taken by Sara L. Matrangola

Body position was sampled at 100 Hz using a Vicon 460 motion analysis system (Vicon, Lake Forest, CA, USA). Data were collected from both sides of the body and averaged to determine position in the sagittal plane. Ground reaction force was sampled at 1000 Hz using a force platform (Bertec Corporation, Columbus, OH). Voltage applied to the electromagnetic clasp that held the harness rope was sampled at 1000 Hz and was used to determine the time of release. Body position, ground reaction force, and clasp voltage data were low-pass filtered at 5, 7, and

20 Hz, respectively (eighth order zero-phase-shift Butterworth filter). Body position data of the lateral maleolus and greater trochanter were used to determine θ_{max}. Additionally, body position data of the lateral maleolus, ground reaction force data, and time of release data were used to determine the torque generated about the ankle using an inverse dynamics analysis.

A mathematical model of the balance recovery task was used to evaluate the relative effects of weight loss and strength training. The model included a slender rod rotating about a hinge joint to represent the human body rotating about the ankle joint with a torque actuator at the ankle (Figure 2). Consistent with an earlier study employing the same model [13], the center of mass (COM) position was assumed to be located at 50% of the subject's height from the ankles.

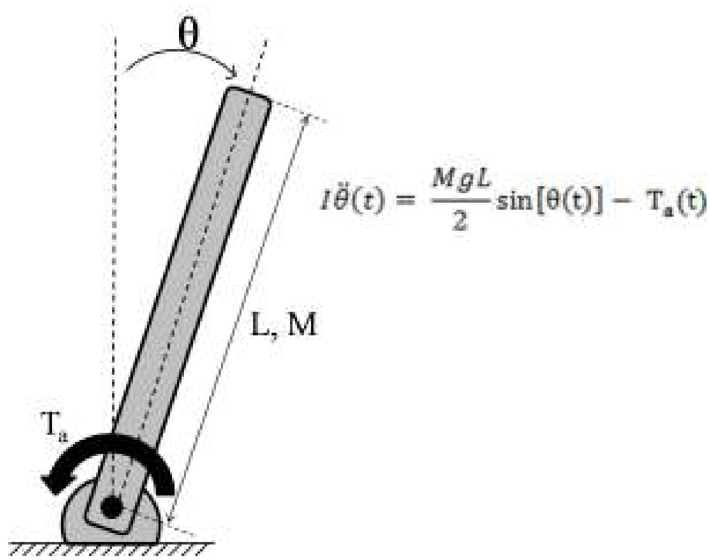
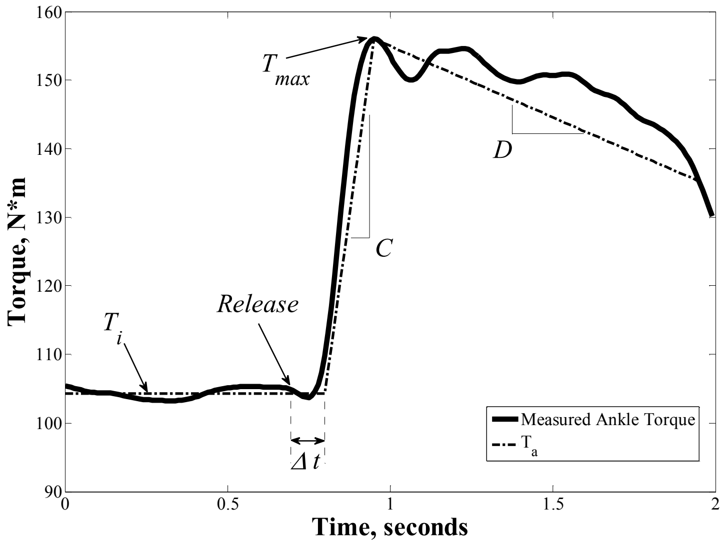


Figure 2. Inverted pendulum model of balance recovery where M, L, and I are the mass, height, and moment of inertia about the ankle of the subject, respectively, g is the gravitational constant (9.81 m/s^2) and T_a is the torque generated about the ankle.

A linear approximation of ankle torque history collected during the experimental trials was applied to the model (Figure 3). Initial ankle torque (T_i), response time (Δt), maximum ankle torque (T_{max}), ankle torque generation rate (C), and ankle torque decline rate (D) were calculated using a similar approach as Robinovitch et al.[13]. The response time was determined to be to the time from release until ankle torque exceed T_i by 2.5 Nm, rather than 5 Nm, due to smaller amplitude of fluctuations of ankle torque prior to release. T_i, T_{max}, C, and D were normalized to the product of body mass (kg) and height (m). Experimental values for θ_{max} and ankle torque properties are given in Figure 3.



Parameter	Mean ± S.D.
θ_{max} (deg)	9.79 ± 1.26
T_i (Nm)*	114. 35 ± 19.12 (0.60 ± 0.11)
Δt (ms)	150 ± 15
C (Nm/s)*†	514.39 ± 204.71 (2.66 ± 1.01)
D (Nm/s)*	91.98 ± 41.46 (0.48 ± 0.21)
T_{max} (Nm)*†	192.47 ± 26.44 (1.00 ± 0.06)

* Values given in parentheses are normalized parameter values = parameter value / {body mass (kg) × height (m)}
† Maximum values across all lean recovery trials

Figure 3. Experimental values and linear approximation (T_a) of ankle torque history during recovery.

Passive torque due to stiffness of the muscles and tendons about the ankle is a component of initial torque and varied with release angle. As such, normalized initial torque during recovery from θ_{max} was found to be linearly related to release angle (Figure 4). This relationship was then used within the model to calculate initial torque for each simulation.

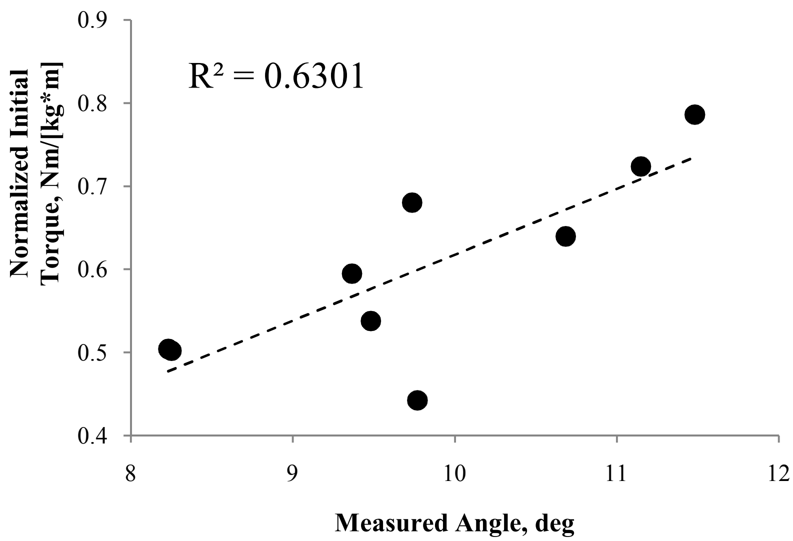


Figure 4. Initial ankle torque measured from subjects prior to release increased as lean angle increased. As a result of this relationship, a regression line fit to these data was used to determine the initial torque for simulations.

The model was validated by comparing the predicted θ_{max} and with measured experimental values. Body mass, height, and ankle torque properties of each subject, with the exception of T_i, were entered into the model to determine the predicted θ_{max}. Recovery was defined by the occurrence of $\dot{\theta}<0$ while $\theta<90°$.

Multiple simulations (using MATLAB, The MathWorks, Natick, MA) were used to determine the effects of altering maximum ankle torque and mass on θ_{max}. Strength training to increase maximum ankle torque also results in an increase in torque generation rate [17]. As such, a second condition for altering maximum ankle torque was investigated that included a concomitant increase in torque generation rate [17]. This condition was termed strength training for the purpose of this study. To simulate weight loss, COM position and radius of gyration (r_{COM}) were varied simultaneously with changes in mass according to Chapter 3. The study presented in Chapter 3 examined changes in body segment inertial parameters with approximately 14% weight loss. For the current study, COM position and r_{COM} were assumed to change linearly with weight loss and were interpolated for various amounts of weight loss.

RESULTS

The balance recovery model reproduced the measured θ_{max} values with moderate accuracy (Figure 5). The root mean squared error (RMSE) was 1.94° and the coefficient of determination between predicted and measured θ_{max} values was $r^2 = 0.4820$.

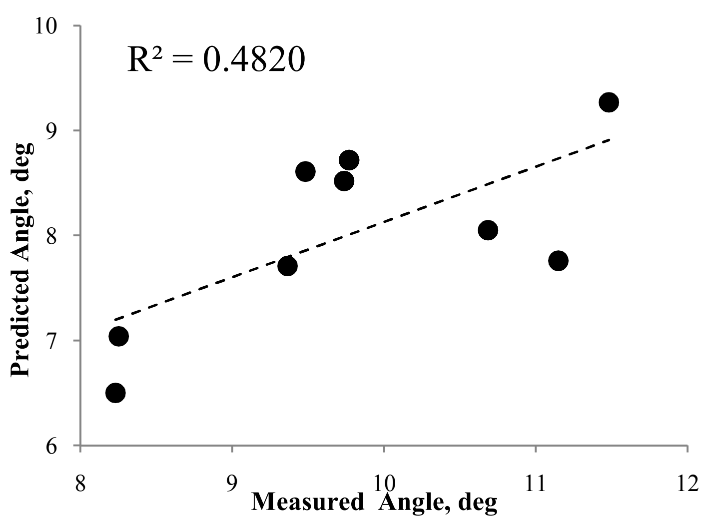


Figure 5. Balance recovery model predicted values versus measured values of θ_{max}

A contour plot was created (Figure 6) to illustrate the relative effects of strength training and weight loss on balance recovery for a representative subject (weight=94.8 kg, height=1.72 m, and BMI=32.0 kg/m^2). Nearly equal changes in strength and weight loss were required to increase θ_{max} by 1 degree. If strength training and weight loss are applied simultaneously, similar changes were required to achieve the same increase in θ_{max} .

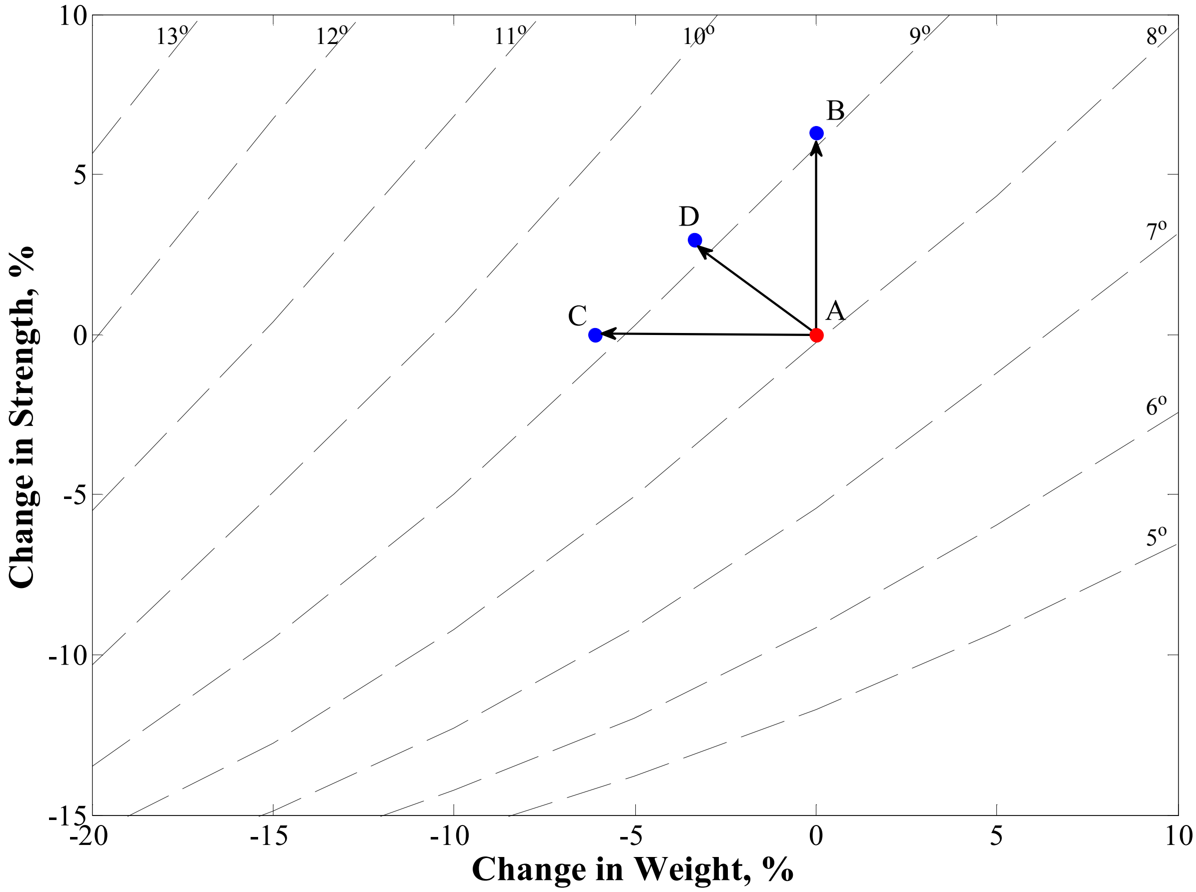
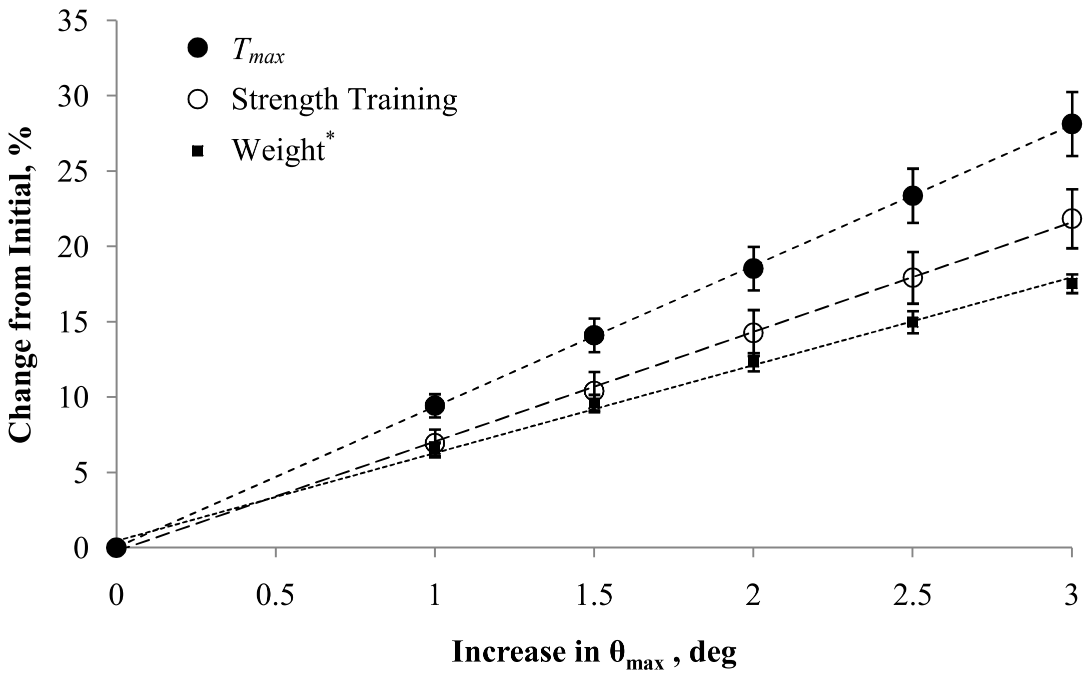


Figure 6. Contour plot balance recovery illustrating the effects of strength and weight on balance. Dashed lines represent recovery at a given angle. Note that the change in strength on the y-axis includes changes in torque generation rate. At baseline (Point A), the model indicates the representative subject will have a θ_{max} of 8.1°. To increase θ_{max} by 1° , the subject would need to increase strength by 6.2% (Point B) or lose 6.1% of original body weight (Point C). If the interventions were applied simultaneously, the smallest magnitude change required to increase θ_{max} by 1° would be a 3.3% decrease in weight and a 3.0% increase in strength (Point D).

Using results averaged across all subjects, changes in weight, maximum ankle torque, and strength were linearly related to changes in θ_{max} (Figure 7). Weight loss consistently required the smallest change among all three interventions to achieve a desired increase in θ_{max}, and increases in maximum ankle torque alone consistently required the largest change. As larger increases in θ_{max} were achieved in our simulations, the differences in the efficacy of the three interventions became more apparent.



* Note: Changes in weight reflect a decrease in weight. Changes in T_{max} and strength reflect increases in these properties.

Figure 7. Average changes in weight and ankle torque parameters to improve θ_{max}. Differences in the three interventions investigated increased with larger increases in θ_{max}. To achieve a one degree increase in θ_{max}, a 6.6 ± 0.4% decrease in weight, 9.4 ± 0.8% increase in T_{max}, or a 6.9 ± 0.9% increase in strength were required.

DISCUSSION

The purpose of this study was to evaluate the effects of weight loss and strength training on balance recovery using an ankle strategy. Our results indicate that both interventions can improve balance, as measured by θ_{max}. In addition, a smaller amount of weight loss was needed compared to increases in strength to achieve equivalent improvement in balance. The difference between these two interventions became more apparent as larger improvements in balance were targeted. These data suggest that weight loss is a more effective intervention than strength training in regards to balance recovery using an ankle strategy. It is unclear at this point if the differences between these interventions are clinically meaningful.

The balance recovery model predicted θ_{max} with moderate accuracy, and the error between predicted and measured values of θ_{max} could be attributed to several factors. First, the balance recovery model does not include feedback control and recovery limits are limited by the temporal constraints imposed by Δt, C, and T_{max}. Once T_{max} is reached (Figure 3), torque will begin to decline, regardless if recovery has been achieved or not. It is plausible that feedback control would maintain a maximum torque in order to achieve balance recovery in some circumstances. This also implies that the model is sensitive to the torque decline rate, D. If we altered our ankle torque histories such that T_a remained constant after T_{max} was reached, the RMSE between measured and predicted θ_{max} was reduced to 1.10°. Robinovitch et al. [13] found similar results in that θ_{max} will slightly increase when ankle torque remains constant after T_{max} is reached. Second, a portion of the error seen in this study can also be attributed to the estimation of the COM position and the r_{COM} of our model of the human body. By using a slender rod as a model, as in the study conducted by Robinovitch et al. [13], we assume that the subject's mass was uniformly distributed the subject's height. Corbeil et al. [12], who conducted modeling investigation of the effects of obesity on the stabilizing torque at the ankle joints, used the Hanavan geometric model [18], a commonly used model to estimate obese body segment inertial parameters. Four subjects from the current study were selected at random to determine the COM location using the Hanavan model. By using this COM location (54.68 ± 2.24% of the subject's height) in the model, the RMSE increases to 3.49°, almost double the RMSE when using a COM location at 50% of the height. However, we feel that our results pertaining to the relative benefits of the interventions investigated were largely insensitive to COM location. Third, θ_{max} was used as a surrogate measure of balance recovery capability and the relation between θ_{max} and risk of falls has not been determined. Fourth, it is unclear if the results of this study using an ankle strategy will generalize to other balance recovery strategies and/or movements.

This study quantified changes in weight and strength needed to improve balance, but these results do not give insight on which intervention may be easier for individuals to achieve. For example, the contour plot (Figure 6) indicates that a ~6% decrease in weight or a ~6% increase in strength would both improve θ_{max} one degree. However, it may be easier for some individuals to achieve a 6% increase in strength rather than a 6% decrease in weight. It is likely that the difficulty of achieving these changes would be highly dependent upon the specific interventions employed and differences between individuals. Results from literature, however, can be used to begin to address this research question. Sartorio et al. [19] completed a study investigating the effects of a weight loss intervention on muscle strength and power. The subjects participated in a short-term (3 week) program that combined exercise (aerobic and strength training) and energy-restricted diet. Aerobic exercise using the lower limbs consisted of bicycling and walking. BMI of the male subjects decreased from 41.3 ± 4.0 kg/m^2 to 39.1 ± 3.7 kg/m^2, corresponding to 5.1 ± 0.9% weight loss, and lower limb strength increased by 32.3 ± 24.5 %. θ_{max} of our representative subject would increase by 4.4 , 0.8, and 5.4 degrees if the subject were to experience similar changes in strength only, weight only, and both strength and weight, respectively. This implies that increases in strength from a similar exercise regiment would provide more benefit to balance than only reduction in weight. Strength training alone can result in weight loss [20], but weight loss alone (i.e. diet only, bariatric surgery) does not typically result in increased strength [21, 22]. Therefore, interventions targeted at strength training may provide more benefit to balance recovery because of increases in strength and decreases in weight.

In conclusion, both reductions in weight and increases in strength can improve balance recovery using an ankle strategy in the obese. In addition, equivalent improvements in balance recovery were achieved with smaller amounts of weight loss than strength training (based on percentages). This suggests that weight loss may be a more effective intervention for reducing risk of falls, but further experimental research is needed to validate these findings.

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CHAPTER 3 - A PRELIMINARY STUDY OF CHANGES IN BODY SEGMENT INERTIAL PARAMETERS OF OBESE INDIVIDUALS WITH WEIGHT LOSS

INTRODUCTION

Obesity is a major and growing health concern in the United States (US). In 2004, 31.1% of adult men in the US were considered obese [1]. The prevalence of obesity (BMI > 30 kg/m^2) continues to rise and has doubled between 1980 and 2002 [1]. Approximately 6%, or $92 billion, of total annual medical expenditures in the US can be attributed to obesity [2]. Several health conditions are also associated with being overweight or obese including type-2 diabetes mellitus, high blood pressure, high cholesterol, asthma, and osteoarthritis [3, 4].

Perhaps the most direct intervention for avoiding or mitigating obesity-related health conditions is weight loss. Several experimental studies have investigated the effects of weight loss on musculoskeletal function and lower extremity kinetics [5-7]. These studies provide quantitative evidence on the benefits of weight loss in terms of function and prevention of the development and/or progression of obesity-related health conditions. For example, Messier et al. [5] reported a 4-fold reduction in knee loads during gait with each pound of weight loss in the obese. An alternative approach to these experimental studies investigating the effects of weight loss is to conduct virtual experiments using forward dynamic simulations of human movement. While they should ultimately be validated with experimental studies, forward dynamics simulations provide a cost-effective initial approach that offers a greater level of experimental control compared to human subject testing, and potentially greater understanding of the underlying biomechanical relationships. Moreover, they can be used to evaluate a variety of different interventions for efficacy prior to expensive and time-consuming clinical trials.

The biomechanical models used in forward dynamic simulations require accurate estimations of body segment inertial parameters (BSIPs) including body segment mass, center of mass (COM) position within each segment, and segment mass moment of inertia. To study weight loss, guidelines describing how BSIPs change with weight loss are necessary so that BSIPs in the biomechanical model can be modified realistically. However, no such guidelines are currently available. Based on this need, the goal of this study was to quantify three-dimensional changes in BSIPs with weight loss. Magnetic resonance imaging (MRI) was used since it is a non-invasive method of determining BSIPs, and has been used previously to estimate BSIPs in children [8] and adults [9-12].

METHODS

MRI images from 19 Caucasian males who participated in a separate study on weight loss [13] were used for the present study. These subjects had a mean (± standard deviation) age of 43.6 ± 7.5 years, and height 177.3 ± 6.9 cm. Body mass and BMI prior to weight loss were 102.7 ± 13.6 kg and 32.6 ± 3.2 kg/m^2, respectively (three of the subjects had a BMI below 30, but above 27.5, prior to weight loss). Subjects were assigned to one of three weight loss programs: seven subjects were assigned to a diet only (DO) program, five to a diet and aerobic exercise (DA) program, and seven to a diet and resistance exercise (DR) program. Details of the diet and exercise programs are reported elsewhere [13]. The weight loss study was conducted in accordance with the ethical guidelines of Queen's University, and all subjects provided written consent before participation.

MRI images both before and after weight loss were acquired using a Philips Gyroscan 1.5-T whole body scanner with a spin-echo sequence, a 500-ms repetition time, and a 20-ms echo time [14]. Subjects laid in a supine position with arms placed straight above the head. Transverse images of 10mm thickness were acquired from the fourth (L_4) and fifth (L_5) lumbar vertebrae to the ankle with a spacing of 50mm between image centers. Subjects were then required to exit the magnet and re-enter head first to acquire images from L_4-L_5 to the wrist.

Tissue discrimination was performed using commercially available medical imaging software (sliceOmatic v4.3, Montreal, Quebec, Canada). Optimal threshold values of pixel brightness for adipose tissue, muscle, organs, and bone were determined using automated procedures with manual correction of obvious artifacts. For each image, tissues were color-coded, and images were exported for subsequent analysis using customized programs in MATLAB (Mathworks, Natick, MA).

A total of six segments, including the whole body, were investigated (Table 1) with segment endpoints and coordinate systems defined by Dumas et al. [15], with the exception of the trunk. For all segment coordinate systems, the x direction corresponds to the anterior-posterior direction, the y axis to the superior-inferior direction, and the z axis to the medial-lateral direction of the segment. BSIPs of appendicular segments were calculated as the average of the left and right sides. The head, neck, hands, and feet were not included based on the expectation that changes in BSIPs of these segments would be considerably less than those segments investigated.

Table 1: Segment Origin Definitions

Segment	Proximal Endpoint	Distal Endpoint
Forearm	Elbow Joint Center	Wrist Joint Center
Upper Arm	Shoulder Joint Center	Elbow Joint Center
Trunk*	Mid-way between Shoulder Joint Centers	Mid-way between Hip Joint Centers
Thigh	Hip Joint Center	Knee Joint Center
Shank	Knee Joint Center	Ankle Joint Center
Whole Body†	Mid-way between Ankle Joint Centers	Mid-way between Shoulder Joint Centers

Notes: The origins of local coordinate systems were located at the proximal endpoint.

* Segment endpoints and coordinate system in this study differed from Dumas et al. [15]. The y axis runs from the proximal to the distal endpoint. The x axis is normal to a plane containing the shoulder joint centers and the hip joint centers, pointing anteriorly. The z axis is the cross product of the x and y axis.

† Whole body y axis runs from the proximal to the distal endpoint. The x axis is normal to a plane containing the shoulder joint centers and mid-way between ankle joint centers, pointing anteriorly. The z axis is the cross product of the x and y axis.

Segment masses were calculated using tissue densities of 1.178 g/cm^3, 1.705 g/cm^3, 1.158 g/cm^3, and 0.563 g/cm^3 for muscle, bone, liver, and lung tissues, respectively [16]. All other tissues were defined as lean tissue with a density of 1.1379 g/cm^3, which was an average of densities for blood, vasculature, tendon, and organs other than the lung and liver. Fat density was approximated as 0.947 g/cm^3, which was an average of the densities of the colon, off-peritoneum, and subcutaneous fat.

The segment COM position was calculated using a two-step procedure. First, the COM position within each scan was calculated as:

$$CM_{xi} = \frac{\sum[\rho_k \sum v_p(x_{ijk} - x_{i0})]}{p_k v_p n_{ik}} \tag{1}$$

where CM_{xi} is the x coordintate of the COM of the i^{th} scan, ρ_k is the density of the kth tissue, v_p is the volume of a 10mm thick pixel, x_{ijk} is the x coordinate of the jth pixel, ith scan, and kth tissue, x_{i0} is the origin of the ith scan, and n_{ik} is the number of pixels of the ith scan and kth tissue. The mass, COM, and I of the intervals between scan planes (40mm) were determined using methods similar to Pearsall et al. [12]. Four 10mm transverse "virtual scans" were created to account for the space between MRI scans and the mass, COM, and I of these scans were linearly interpolated from the bounding scan planes. The segment COM in the x direction can then be calculated as:

$$COM_x = \frac{\sum CM_{xi} m_i}{\sum m_i} \tag{2}$$

where m_i is the mass of the ith scan. Segment COM was calculated relative to the segment origin (Table 1).

The segment radii of gyration were calculated using a three-step procedure. First, the moment of inertia for each scan was calculated as:

$$I_{xi} = \sum \left[\rho_k \sum v_p \left\{\left(y_{ijk} - CM_{yi}\right)^2 + \left(z_{ijk} - CM_{zi}\right)^2\right\}\right] \tag{3}$$

where y_{ijk} and z_{ijk} are y and z coordinate of the jth pixel, ith scan, and kth tissue, respectively, and CM_{yi} and CM_{zi} are the y and z coordinates of the COM of the ith scan, respectively. The segment mass moment of inertia was calculated as:

$$I_x = \sum I_{xi} + m_i \left\{\left(CM_{yi} - COM_y\right)^2 + (CM_{zi} - COM_z)^2\right\} \tag{4}$$

where COM_y and COM_z are the segment COM in the y and z direction, respectively. The segment radius of gyration about the segment COM was then determined by:

$$r_x = \sqrt{\frac{I_x}{\sum m_i}} \tag{5}$$

A Wilcoxon Signed-Rank test was used to analyze changes in BSIPs with weight loss. The overall Type 1 error rate for the 42 tests performed (6 segments × 7 parameters) was controlled using an approach termed false discovery rate control. This approach is an alternative to the commonly used Bonferroni procedure and controls the proportion of significant results that are in fact Type 1 errors [17]. Additionally, false discovery rate control does not suffer from the undesirable lack of statistical power associated with the Bonferonni procedure with an increasing number of tests [18, 19]. The trunk and whole body BSIPs of one subject were excluded from the analysis due to unequal segment lengths before and after weight loss. Statistical analysis was conducted using JMP v6 (Cary, North Carolina, USA).

RESULTS

Subjects lost 14.2 ± 3.4 kg or 13.8 ± 2.4 % of initial body weight with weight loss. This resulted in a 4.5 ± 1.0 unit decrease in BMI (four subjects still had a BMI > 30 kg/m^2 after weight loss). The MRI method estimated that subjects lost 13.9 ± 3.0% body weight, yielding a root mean squared error of 1.9% from actual weight loss.

Average changes in segment mass, COM position, and *r* with weight loss are given in Table 2. Effect sizes and p-values are given in Table 3. Segment mass decreased significantly in all segments. COM position moved distally for the thigh and upper arm, superiorly for the trunk, and inferiorly for the whole body. In general, segment *r* decreased in all segments.

Table 2: Percent Changes in Body Segment Inertial Parameters

Segment	Δ Mass	Δ COM_x	Δ COM_y	Δ COM_z	Δ r_x	Δ r_y	Δ r_z
Forearm	-9.53 ± 5.99*	-0.05 ± 1.42	-0.33 ± 1.25	0.68 ± 1.21*	-0.08 ± 1.07	-0.65 ± 1.94*	-0.14 ± 1.19
Upper Arm	-15.68 ± 6.50*	-0.15 ± 1.02	-0.60 ± 1.01*	0.32 ± 1.86	-0.13 ± 0.71	-1.38 ± 1.18*	0.00 ± 0.76
Trunk	-15.25 ± 3.99*	0.36 ± 1.13	0.63 ± 0.88*	0.03 ± 1.49	-0.54 ± 0.75*	-1.87 ± 1.25*	-0.26 ± 0.63*
Thigh	-13.06 ± 2.97*	-0.20 ± 0.95	-0.75 ± 0.43*	-0.06 ± 0.70	-0.72 ± 0.93*	-1.80 ± 1.25*	-0.86 ± 0.97*
Shank	-7.85 ± 3.01*	-0.50 ± 0.67*	-0.08 ± 0.62	0.41 ± 2.43	-0.06 ± 0.33	-0.89 ± 0.79*	-0.03 ± 0.36
Whole Body[&]	-13.91 ± 3.04*	-0.06 ± 0.77	-0.97 ± 1.77*	0.01 ± 0.46	0.63 ± 0.90*	-0.44 ± 0.19*	0.71 ± 0.92*

Notes: Changes in mass are given as percent change from pre weight loss mass. Changes in COM and r are given as percent of segment length. For appendicular segments, a positive ΔCOM_x, ΔCOM_y, and ΔCOM_z indicates the COM moved anteriorly, proximally, and medially, respectively. For the whole body and trunk, a positive ΔCOM_y indicates the COM moved superiorly.
[&] Whole body does not include head, neck, hands, and feet. * denotes a significant change with weight loss.

Table 3: Effect Sizes and p-Values for Changes in Body Segment Inertial Parameters

Segment	Δ Mass	Δ COM_x	Δ COM_y	Δ COM_z	Δ r_x	Δ r_y	Δ r_z
Forearm	0.2799 (<0.0001)*	-0.0389 (0.9217)	-0.2634 (0.3955)	0.2102 (0.0108)*	-0.0734 (0.6226)	-0.2499 (0.0181)*	-0.1134 (0.5949)
Upper Arm	-0.2367 (<0.0001)*	-0.1267 (0.5153)	-0.4827 (0.0141)*	0.2031 (0.9843)	-0.2413 (0.4413)	-0.9927 (0.0004)*	0.0066 (0.6794)
Trunk	-0.3711 (<0.0001)*	0.2756 (0.2121)	0.4315 (0.0016)*	0.0232 (0.7019)	-0.7106 (0.0047)*	-1.3137 (<0.0001)*	-0.4361 (0.0342)*
Thigh	0.1637 (<0.0001)*	-0.1616 (0.3124)	-0.7631 (<0.0001)*	-0.0569 (0.6507)	-0.5644 (0.0003)*	-1.0115 (<0.0001)*	-0.6608 (<0.0001)*
Shank	0.8081 (<0.0001)*	-0.6024 (0.0071)*	-0.0913 (0.3525)	0.0940 (0.4413)	-0.1052 (0.6794)	-0.7499 (<0.0001)*	-0.0612 (0.8906)
Whole Body	-1.0433 (<0.0001)*	-0.1109 (0.4683)	-0.4148 (0.0432)*	0.0269 (1.000)	0.0888 (0.0003)*	-1.4102 (<0.0001)*	0.6821 (<0.0001)*

Notes: p-values are given in parentheses. Effect size was calculated using Cohen's *d*, the mean value before weight loss subtracted from the mean value after weight loss, divided by the standard deviation (before weight loss).
* denotes a significant change with weight loss.

DISCUSSION

The goal of this study was to quantify changes in BSIPs with weight loss. Weight loss resulted in changes in many BSIPs, primarily due to changes in segment mass and changes in the distribution of mass along the segment's longitudinal axis.

Several limitations to the current study warrant discussion. First, the spacing between adjacent MRI image centers was 50mm, which was larger than the 8-25mm used in other studies that derived BSIPs with MRI [8-11]. This spacing could contribute to systematic errors in BSIPs primarily through inaccuracies in identification of segment endpoints. If a segment endpoint was between images, for example, then the segment was extended to the next image and would include mass from the adjacent segment. However, this study was interested in calculating changes in BSIPs, and such error would be systematic. Additionally, we do not think that this was a significant contribution to the overall error. A second limitation was that subjects laid in a supine position during imaging. This position likely contributed to some soft tissue deformation compared to the vertical position, which is likely the position of the body in most modeling studies. A third limitation was that the head, neck, hands, and feet were not included in this study. However, previous studies have shown that adipose tissue in the head and neck does not significantly change with weight loss [20] or age [21]. In addition, the MRI method calculated weight loss with a root mean squared error of 1.9% from actual weight loss. This indicates that the excluded segments were not a significant source of weight loss compared to the segments investigated in this study. Lastly, data were only collected from Caucasian male subjects, and therefore results may not be generalizable to other populations.

In conclusion, changes in BSIPs with weight loss were quantified in obese Caucasian males using MRI. These data can be used to investigate the biomechanical effects of weight loss using a biomechanical model and forward dynamic simulations.

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