

**Effects of Obesity and Age on  
Muscle Strength, Gait, and Balance Recovery**

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## **ABSTRACT**

Obese and older adults are reported to have a higher rate of mobility limitation and are at a higher risk of fall compared to healthy-weight and young counterparts. To help identify potential mechanisms of these mobility limitations and higher risk of falls, the purpose of the research within this dissertation was to investigate the effects of obesity and age on muscle strength, gait, and balance recovery.

Three experimental studies were conducted. The purpose of the first study was to investigate the effects of obesity and age on extension and flexion strength at the hip, knee, and ankle. Absolute strength among obese participants was higher in dorsiflexion, knee extension, and hip flexion compared to healthy-weight participants. Strength relative to body mass was lower among obese participants in all joints/exertions. This lack of uniformity across the 6 exertions is likely due to the still unclear underlying biomechanical mechanism responsible for these strength differences, which may also be influenced by aging.

The purpose of the second study was to investigate the effects of obesity, age and, their interactions on relative effort at the hip, knee, and ankle during gait. The peak relative effort for each joint/exertion was expressed by peak NMM during gait as a percentage of the maximum available NMM. The relative effort in hip, knee, and ankle was higher among obese compared to healthy-weight participants. This higher relative effort in hip, knee, and especially in the ankle can be a contributing factor to compromised walking ability among obese individuals.

The purpose of the third study was to investigate the effects of age-related strength loss on non-stepping balance recovery capability after a perturbation while standing, without constraining the movements to ankle strategy. The balance recovery capability was quantified by the maximum recoverable platform displacement (MRPD) that was withstood without stepping. Two experiments were conducted. The first experiment involved human subjects and the results suggested that MRPD was lower among older participants compared to young participants. The second experiment involved a simulation study to manipulate muscle strength at hip, knee, and ankle. The results suggested that MRPD was reduced in cases of loss of strength in ankle plantar flexion and hip flexion compared to the young model and did not differ in rest of the cases. The finding suggested that plantar flexor strength plays a major role in capability to recover balance even though the movement was not constrained to the ankle.

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## **Chapter 1 - Overview**

### **1.1 Research Problem**

Limited mobility and a higher risk of falls are among the health concerns associated with obesity and aging. Limited mobility can adversely affect quality of life (Fjeldstad et al. 2008), and is a contributing factor to higher rate of mortality among older individuals (Hirvensalo et al. 2000). Falls are twice as common among individuals who are obese compared to their non-obese counterparts (27%-15%), and over one-third of older adults over the age of 65 falls each year (Hahn et al. 2014). Moreover, falls-related injuries are among the most common causes of hospitalization among individuals who are obese (Matter et al. 2007), and an estimated 1.64 million older adults were treated in emergency departments in 2001 for unintentional fall injuries (Stevens and Sogolow 2005).

The prevalence of obesity is very high and the population of older adults is increasing in the United States. Currently, over 30% of the United States population are considered obese (Ogden et al. 2014). Moreover, the population of older adults over the age of 65 is growing in the United States, and it is projected to double from 2010 to 2050 and reach over 80 million (Vincent and Velkoff 2010). The prevalence of obesity also increases with aging (Villareal et al. 2005). Compared to adults aged 20-39, adults over the age of 60 have a higher odds ratio for obesity (1.26 females and 1.35 males) (Flegal et al. 2010). This is problematic since obesity can exacerbate the health concern associated with aging (Vincent et al. 2010).

Altered muscle strength is associated with limited mobility (Jadelis et al. 2001; Stenholm et al. 2009; Visser et al. 2005) and higher risk of falls (Daubney and Culham 1999; Moreland et al. 2004). Obesity can alter lower extremity strength in that individuals who are obese exhibit higher absolute strength in knee extension (Capodaglio et al. 2009; Maffiuletti et al. 2007), but lower relative (to body mass) strength in knee extension and flexion (Capodaglio et al. 2009; Koushyar et al. 2013), ankle plantar flexion and dorsiflexion, and hip extension (Koushyar et al. 2013). Aging has also been associated with a decrease in absolute muscle strength (Dean et al. 2004; Savelberg and Meijer 2004). The effects of obesity on knee extensor strength has been fairly well characterized (Capodaglio et al. 2009; Hulens et al. 2001; Lafortuna et al. 2005; Maffiuletti et al. 2007; Miyatake et al. 2000). However, the effects of obesity on strength at the ankle and hip are not well characterized. In fact, a comprehensive evaluation of the effects of obesity, and the combined effects of obesity and age, on lower extremity strength is needed to better understand how these two increasingly important factors contribute to limited mobility and a higher risk of falls.

Altered lower extremity strength may help explain the kinematic and kinetic differences observed in gait between obese and healthy-weight individuals (Browning and Kram 2007; Lerner et al. 2014). A better understanding of the influence of altered strength on gait may be provided by determining the effect of obesity on relative effort during gait, or the percentage of available strength that is used while walking. Studying this would give insight into the underlying factors which contribute to the gait alterations associated with obesity, and could contribute to strategies to improve and maintain mobility among individuals who obese.

Altered muscle strength can also contribute to the higher risk of falls among older individuals (Daubney and Culham 1999; Moreland et al. 2004). Limited ability to recover balance from a postural perturbation due to reduced muscle strength (Barrett and Lichtwark 2008; Madigan and Lloyd 2005), may be one of the contributing factors to higher risk of falls. Therefore, studies are needed to better understand the biomechanical relationship between strength and balance recovery capability. The results from such studies would provide insight into the underlying factors which contribute to higher rate of falls among older individuals, and could help guide the development of strength training interventions to reduce the risk of falls among these individuals.

This proposal describes three studies that broadly investigate how strength is affected by obesity and age, and how these alterations in strength influence gait and balance recovery.

## **1.2 Document Organization**

This document is organized into 6 chapters. Chapter 2 is a literature review which addresses the health concerns associate with obesity and age, followed by the effects of obesity and age on muscle strength, gait and balance recovery. The last part of the literature review is a brief description of forward dynamics simulation which will be used in Chapter 5. Chapter 3 entitled *“Relative Strength at the Hip, Knee, and Ankle is lower among Young and Older Females Who are Obese”* presents the first study which measured the maximum voluntary strength at ankle, knee, and hip in human participants to more comprehensively characterize how obesity and age influence lower extremity strength. Chapter 4 entitled *“Relative Effort during Gait is higher among Individuals Who are Obese”* presents the second study in which the gait data collected from human participants was used, along with their strength measurements, to find the effect of

obesity and age and their interaction on relative effort during gait. The results could contribute to strategies to improve and maintain mobility among individuals who are obese. Chapter 5 entitles “Effect of Age-Related Strength Loss on Non-stepping Balance Recovery” presents the third study in which human subject and simulation studies were performed to investigate the relative effect of age-related loss of strength on balance recovery capability, therefore, the results could help guide the development of strength training interventions to reduce the risk of falls among older individuals. Finally, Chapter 6 summarizes the findings and present ideas for future work.

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## **Chapter 2 - Introduction**

### **2.1 Obesity and Aging: Health Concerns**

Obesity has become a major health concern in the United States. Obesity is typically defined as having a body mass index (BMI) over 30 kg/m<sup>2</sup> (Organization 2000). Over one-third of adults in the United States are considered obese (Ogden et al. 2014), and although the prevalence of obesity remained stable in general adult population between 2003/2004 and 2009/2010, the high prevalence is still alarming because obesity is associated with numerous maladies such as heart disease, stroke, hypertension, diabetes, and osteoarthritis (Kopelman 2000). Obesity is also linked to mobility limitation (Koster et al. 2007; Vincent et al. 2010), and a higher risk of falls (Fjeldstad et al. 2008).

Another major health concern in the United States is the growing number of older adults. The number of older adults over the age of 65 is expected to double from 40.2 million in 2010 to over 80 million in 2050 (Vincent and Velkoff 2010). Aging has also been associated with mobility limitation (Grabiner and Enoka 1995; Horak et al. 1989; Vandervoort 1992) and higher risk of falls (Rubenstein 2006). Moreover, many of the health concerns that are prevalent among individuals who are obese, such as heart disease, hypertension, osteoarthritis, are also common among older adults (Statistics 2012).

The prevalence of obesity also increases with aging (Flegal et al. 2010; Villareal et al. 2005). Compared to adults aged 20-39, adults over the age of 60 have a higher odds ratio for obesity (1.26 females and 1.35 males) (Flegal et al. 2010). Although the prevalence of obesity has

remained stable in general adult population, the prevalence of obesity in women over the age of 60 has increased from 31.5% in 2003/2004 to 38.1% in 2011/2013 (Ogden et al. 2014). This is a major health concern since obesity can exacerbate the health risks associated with aging (Dutil et al. 2013; Fjeldstad et al. 2008; Vincent et al. 2010). For example, the rate of Osteoarthritis is 50% among older individuals who are obese compared to just 31% among non-obese older individuals (Fjeldstad et al. 2008). BMI is also suggested to be a consistent predictor of the onset or worsening of mobility disability among older individuals (Vincent et al. 2010). Obesity also has a negative impact on postural control among older women, and can therefore be considered a potential contributing factor for falls.

## **2.2 Obesity and Aging: Limited Mobility and Higher Risk of Falls**

One of the main problems associated with obesity is mobility limitation (Hue et al. 2007; Koster et al. 2007; Vincent et al. 2010). This can include difficulty rising from a chair (Galli et al. 2000; Pataky et al. 2014), difficulty ascending or descending from stairs (Stickles et al. 2001), lower gait speed (Lai et al. 2008; Spyropoulos et al. 1991), impaired dynamic balance (Hue et al. 2007; Jadelis et al. 2001; Menegoni et al. 2009), and compromised walking ability (Angleman et al. 2006; Lamb et al. 2000; Stenholm et al. 2009). For example, Menegoni et al. (2009) investigated the gender-specific effect of obesity on balance. They found that among females, obese individuals exhibited larger mean postural sway in anterior-posterior axis ( $10.0 \pm 1.4$  mm) compared to normal-weight counterparts ( $6.2 \pm 1.6$  mm) (Menegoni et al. 2009). Moreover, individuals who were obese walked at a slower speed (1.09 m/s) compared to normal-weight individuals (1.64 m/s) (Spyropoulos et al. 1991), and needed more time (11.29 sec) to complete

five sit-to-stand movements compared to their normal-weight counterparts (8.28 sec) (Pataky et al. 2014).

Obesity effects the mobility limitation among older individuals (Davison et al. 2002; Dutil et al. 2013; Launer et al. 1994). Older obese women were twice as likely to report functional limitations compared to normal-weight counterparts, which included difficulty in walking a quarter mile and difficulty standing up from an armless chair (Davison et al. 2002). Older adults who were obese walked at significantly lower speed compared to normal-weight older adults (1.06 m/s vs. 1.2 m/s) (Ko et al. 2010). Moreover, older individuals who were obese exhibited higher center of pressure speed during quiet standing with eyes open, compared to normal-weight counterparts ( $0.99 \pm 0.29$  cm/s vs.  $0.70 \pm 0.16$  cm/s). Also, older obese adults exhibited greater postural sway compared to non-obese counterparts in both medial-lateral ( $0.96 \pm 0.46$  cm vs.  $0.82 \pm 0.37$  cm) and anterior-posterior ( $1.87 \pm 0.39$  cm vs.  $1.59 \pm 0.43$  cm) axes (Dutil et al. 2013).

Another major problem facing individuals who are obese is a higher risk of falls. It has been reported that obese individuals are twice as likely to fall compared to the healthy-weight adults (27% vs. 15%) (Fjeldstad et al. 2008). Moreover, the prevalence of fall-related injuries is higher among individuals who are obese (Finkelstein et al. 2007), and fall-related injuries are the number one cause of hospitalization among these individuals (Matter et al. 2007).

Aging has also been associated with a higher prevalence of falls and falls-related injuries. About 30% of older adults over the age of 65 fall each year (Rubenstein 2006). In 2001, an estimated

1.64 million older adults were treated in emergency departments for unintentional fall injuries (Stevens and Sogolow 2005). Women included 70.5% of the population included in previous study, and also had 1.8 times rate of hospitalization compared to men (Stevens and Sogolow 2005). Obesity can increase the risk of falls among older individual over the age of 65 (Himes and Reynolds 2012), with the odds ratio for risk of falls being 1.12 among class I obesity ( $30 \text{ kg/m}^2 < \text{BMI} < 34.9 \text{ kg/m}^2$ ) compared to normal-weight individuals.

### **2.3 Obesity and Aging: Muscle Strength**

Muscle strength is an important contributor to mobility and functional capacity. Moreover, muscle strength is an important component of maintaining balance and also a main contributing factor to risk of falls (Fukagawa et al. 1995; Moreland et al. 2004).

Individuals who are obese exhibit altered lower extremity strength compared to non-obese individuals (Capodaglio et al. 2009; Hulens et al. 2001; Lafortuna et al. 2005; Maffiuletti et al. 2007; Miyatake et al. 2000). Altered lower extremity strength is a likely underlying cause for mobility limitation among individuals who are obese (Jadelis et al. 2001; Stenholm et al. 2009; Visser et al. 2005). Lower extremity strength can be characterized in terms of “absolute strength”, which is the maximum force in Newtons or net muscle moment in Newton-meters generated at a joint. It can also be defined in terms of “relative strength”, which is the absolute strength divided by body mass (Capodaglio et al. 2009; Miyatake et al. 2000), or fat-free mass (Hulens et al. 2001; Maffiuletti et al. 2007). Absolute strength of the knee extensors is higher among individuals who are obese (Capodaglio et al. 2009; Hulens et al. 2001; Lafortuna et al. 2005; Maffiuletti et al. 2007), but relative strength of the knee extensors is lower among these

individuals (Capodaglio et al. 2009; Maffiuletti et al. 2007; Miyatake et al. 2000). Maffiuletti et al. (2007) investigated differences in isometric and isokinetic (at 60, 120, and 180 deg/sec) quadriceps strength between lean and obese subjects. Their results showed that individuals who were obese exhibited about 20% higher absolute torque, especially at higher angular velocities. When the absolute values were normalized to body mass, individuals who were obese exhibited approximately 32% lower torque across all angular velocities. The isometric data were consistent with isokinetic data in that individual who were obese had 16% higher absolute strength and 34% lower relative strength.

The knee extensor muscle group plays a major role in supporting the body against gravity while standing and walking, and is essential for locomotion (Neder et al. 1999). Therefore, the higher absolute strength at the knee extensors among individuals who are obese is thought to be an adaptation to the chronic exposure to increased body mass (Capodaglio et al. 2009; Hulens et al. 2001). Consistent with this hypothesis, absolute strength of the knee flexors, which do not play a major role in supporting the body against gravity while standing or walking, is not higher among individuals who are obese (Capodaglio et al. 2009; Hulens et al. 2001). For example, Capodaglio et al. (2009) determined the knee flexor/extensor ratio from their strength measurements and reported a significantly lower ratio among individuals who are were compared to their healthy-weight counterparts, which supports this hypothesis. Despite the importance of ankle strength in postural stability (Jadelis et al. 2001) and balance recovery using an ankle strategy (Matrangola and Madigan 2009), and hip strength on producing the whole leg movement (Dean et al. 2004), the effects of obesity on ankle and hip strength has received little attention.

Aging has been associated with a decrease in muscle strength (Dean et al. 2004; Savelberg and Meijer 2004; Simoneau et al. 2007) which, in turn, can negatively impact the mobility of older individuals. Older individuals exhibited lower muscle strength in ankle plantar flexion (Simoneau et al. 2007), knee extension and flexion (Savelberg and Meijer 2004), and hip extension and flexion (Dean et al. 2004) compared to young adults. Lower extremity muscle strength is a good predictor of functional capability and mobility among older adults (Avlund et al. 1994; Hasselgren et al. 2011). In a study by Hasselgren et al. (2011) on patients in geriatric rehabilitation, lower extremity strength, especially knee extension and ankle dorsiflexion strength, were related to mobility including sitting from a supine position in bed and transferring from bed to chair. Rate of falls is also reported to be lower in individuals who have stronger lower extremities (Daubney and Culham 1999; Moreland et al. 2004). The older individuals who reported falls had lower ankle dorsiflexor and hip extensor strength (Daubney and Culham 1999).

Obesity also alters muscle strength among older individuals (Miyatake et al. 2000; Rolland et al. 2004). Despite young obese adults, which exhibited higher absolute knee extensor strength compared to young normal-weight adults, older obese adults did not exhibit higher absolute knee extensor strength compared to normal-weight older adults (Miyatake et al. 2000). The direction of change of strength among older individuals who are obese is not intuitive because of the loss of strength associated with aging (Dean et al. 2004; Hunter et al. 2000), and the increase in absolute strength associated with obesity (Capodaglio et al. 2009; Maffiuletti et al. 2007; Miyatake et al. 2000). Therefore, studies are needed to investigate the interaction of age and obesity on lower extremity strength.

## 2.4 Obesity and Aging: Gait

Obesity is associated with slower gait speed (Lai et al. 2008; Spyropoulos et al. 1991), shorter step lengths (Hulens et al. 2003; Page et al. 2014), longer stance phase duration, and longer double support phase (Browning and Kram 2007; DeVita and Hortobágyi 2003; Spyropoulos et al. 1991). Spyropoulos et al. (1991) reported that obese men walked at 1.09 m/sec which was slower than non-obese counterparts at 1.64 m/sec. They also observed that individuals who were obese had a shorter stride at 1.25 m compared to non-obese at 1.67m. Studies have also reported that individuals who are obese exhibited higher absolute hip and knee extensor torques (Browning and Kram 2007), no differences in absolute hip and knee extensor torques (DeVita and Hortobágyi 2003; Lai et al. 2008), and higher absolute ankle plantar flexor torque (Browning and Kram 2007; DeVita and Hortobágyi 2003). Browning and Kram (2007) showed that the absolute peak knee extensor torque was 51% greater for obese versus normal-weight subjects while walking at 1.5 m/sec. However, DeVita and Hortobagyi (2003) did not report any differences in peak knee extensor torque, and 61% higher torque in ankle plantar flexor torque between obese and lean participants at a standard speed of 1.5 m/sec. Similar to young individuals who were obese, older obese individuals walked with slower speed, but exhibited no differences in peak sagittal plane joint torques of lower extremity during gait (Ko et al. 2010).

Aging has also been associated with slower preferred gait speed (Kerrigan et al. 1998) and shorter step length (Anderson and Madigan 2014; DeVita and Hortobagyi 2000). Older adults walk with a smaller peak hip extension angle (Monaco et al. 2009; Yeadon et al. 2006) compared to young adults. Aging also causes a redistribution of joint torque which results in more usage

of hip extensors and less usage of knee extensors and ankle plantar flexors when walking speed is kept similar among older and young participants (DeVita and Hortobagyi 2000).

Reduced lower extremity strength may help explain the kinematic and kinetic differences observed during gait between obese and normal-weight individuals (Browning and Kram 2007; Lerner et al. 2014). In an effort to better understand the importance of lower extremity strength for mobility, biomechanics researchers have investigated relative effort (or joint torque expressed as a percentage of maximum available joint torque) during sit-to-stand (Bieryla et al. 2009), stair ascent and descent (Samuel et al. 2013), and gait (Anderson and Madigan 2014; Requião et al. 2005). Anderson and Madigan (2014), for example, reported ankle plantar flexor torque to be close to 100% among older adults aged 75 and over while walking fast, indicating little or no capacity to increase plantar flexor torque (Anderson and Madigan 2014). These findings support the hypothesis that reduced plantar flexor strength contributes to reduced plantar flexor torque among older adults during gait (DeVita and Hortobagyi 2000). Similar to how higher relative effort among older adults may lead to age-related differences in gait kinetics (or kinematics), higher relative effort among obese adults may lead to the obesity-related differences in gait kinetics, and help identify underlying factors contributing to altered gait among individuals who are obese.

## **2.5 Obesity and Aging: Balance Recovery**

Obesity has been associated with a limited ability to recover from postural perturbation using ankle strategy (Matrangola and Madigan 2011). Matrangola and Madigan (2011) examined the effect of obesity on balance recovery using an ankle strategy. Three types of postural

perturbations were tested which included an initial angular displacement, an initial angular velocity from natural stance, and an initial angular velocity from a predefined position. The position was imposed by a static lean angle, and the angular velocity was applied by an abrupt force impulse to the upper back using a ballistic pendulum. Recovery was quantified as the maximum recoverable initial angular position or angular velocity. They found that obesity only limits balance recovery if initial angular velocity is applied. The maximum recoverable initial velocity was 29.9% and 20.1% lower among individuals who were obese for the trials using initial angular velocity from natural stance, and using initial angular velocity from a predefined position, respectively.

Older adults commonly exhibit an impaired balance recovery capability after a postural perturbation while standing (Mackey and Robinovitch 2006; Madigan and Lloyd 2005; Mansfield and Maki 2009; Thelen et al. 2000; Wojcik et al. 1999). For example, Mackey and Robinovitch (2006) determined the largest static lean angle from which young and older women could recover balance upon release when using the so-called ankle strategy. Older women exhibited a 36% smaller maximum lean angle (13.1 deg) compared to young women (16.3 deg). Wojcik et al. (1999) determined the largest lean angle from which young and older females could recover by taking a single step. They found that older females exhibited 47% smaller maximum lean angle (16.2 deg) compared to young females (30.7 deg). Similarly, Madigan and Lloyd (2005) reported that older males exhibited 31% smaller maximum lean angle (20.5 deg) compared to young males (29.9 deg) while recovering balance by taking a single step.

Altered lower extremity muscle strength may contribute to the higher risk of falls among individuals who are obese, and is also one of the major contributors to higher risk of falls among older individuals (Daubney and Culham 1999; Moreland et al. 2004). Despite the fact that individuals who are obese exhibit higher absolute strength in knee extension (Capodaglio et al. 2009; Maffiuletti et al. 2007; Miyatake et al. 2000), relative strength (or strength divided by body mass) is lower in knee extension and flexion (Capodaglio et al. 2009; Koushyar et al. 2013), ankle plantar flexion and dorsiflexion, and hip extension and flexion (Koushyar et al. 2013) compared to healthy-weight adults. Older adults exhibit lower absolute, and consequently relative, muscle strength at ankle plantar flexion (Simoneau et al. 2007), knee extension and flexion (Savelberg and Meijer 2004), and hip extension and flexion (Dean et al. 2004) compared to young adults. Reduced relative strength in the lower extremity likely limits the ability to recover balance from a postural perturbation to avert a fall. Madigan and Lloyd (2005), for example, reported increasing net muscle moments in plantar flexion, knee extension, and hip extension as adults recovered balance with a single step after being released from progressively larger static lean angles. This indicates greater strength is needed as perturbation severity increases, therefore, individuals who are obese or older individuals may not be able to recover more severe perturbation due to their lower relative strength.

Balance recovery capability after a postural perturbation can be improved by increasing the muscle strength (Matrangola and Madigan 2009; Robinovitch et al. 2002). Robinovitch et al. (2002) used a combination of human subjects' data and mathematical modeling to investigate the effect of ankle torque magnitude and the rate of increase of ankle torque on balance recovery using an ankle strategy after release from a static forward lean. They found that increasing the

maximum ankle torque by 100% resulted in an increase in the maximum recoverable lean angle from 5.3 deg to 8.7 deg. Matrangola and Madigan (2009) investigated the relative effect of strength training and weight loss on balance recovery using ankle strategy. Among individuals who were obese, a 15.3% increase in ankle strength was required to add 1 degree to the maximum lean angle which could be recovered. While both of these studies provide useful insight on the importance of ankle plantar flexor strength on balance recovery without stepping, constraining the movement to an ankle strategy limits the ability to understand the role of knee and hip strength on balance recovery.

## **2.6 Forward Dynamic Simulation**

Two approaches are generally used to perform analysis for rigid-body models of human movement. The first approach is called inverse dynamics which utilizes the measured kinematics (e.g. angle and angular velocity) to determine the kinetics (e.g. joint torques) that contributed to the measured kinematics (Winter 2009). This approach can be generally thought to measure the “effect”, and calculate the “cause”, and is commonly used to help understand torques the body generates to control movement. The second approach is called forward dynamics simulation which utilizes torques to determine the angle and angular velocity (Winter 2009). In this approach, the kinematics are not required as an input. This approach can be generally thought to manipulate the “causes”, and determine the “effect”, and it is commonly used to investigate how changes in torques the body generate, affect model’s kinematics. Therefore, this approach has been utilized to answer many of the reasearch questions related to human movement.

Forward dynamics simulation of human motion has several benefits and limitations. First benefit of this approach is that it provides the opportunity to investigate variables which are difficult to manipulate in a human subject. As an example, the specific muscle parameters such as maximum isometric muscle strength can be manipulated using a model. Second benefit is that it gives the researcher an increased experimental control. For example, it is possible to change maximum isometric strength in a human subject, but other variables such as subject weight might vary as well, making it more difficult to isolate the effects of strength on an outcome measure. The third benefit is that it is possible to investigate a cause and effect relationship which might exist in a dynamic system. For example, Robinovitch et al. (2002) investigated the effect of magnitude and the speed of torque development on balance recovery using ankle strategy. They found that increasing the maximum ankle torque from 0.6 Nm/(kg.m) to 1.2 Nm/(kg.m), resulted in 64% increase in the maximum recoverable lean angle, which could be recovered after a tether release (Robinovitch et al. 2002). Because inverse dynamics measures the “effect” to calculate the “cause”, it is difficult to determine a cause and effect relationship. Despite the powerful benefits of forward dynamic simulation, it also has several limitations. First, forward dynamics frequently requires significant simplifications of a much more complicated system. It is easier to understand the effect of a parameter of interest (e.g. max isometric torque) using a simple model but the simple model may not be able to capture the true nature of the problem. Second, it is often difficult to select parameters required for modeling. Third, forward dynamics simulation requires complicated mathematical modeling which requires numerical approaches to be solved. This results in a significant computational time.

The following framework will be used to perform forward dynamic simulations as a part of the proposed work (Figure 2.1). First, joint activation is prescribed to scales the torque at each joint. Then, the prescribed activation is combined with the current state of the system (angle and angular velocity) to determine the amount of joint torque created at each instant in time. Afterward, the determined joint torques along with the external forces or momets are entered in to equation of motion to determine the segments' accelerations which are then used to find body's new velocity and position. Finally, the new state of the model is used to determine body's motion in the nex time increment.

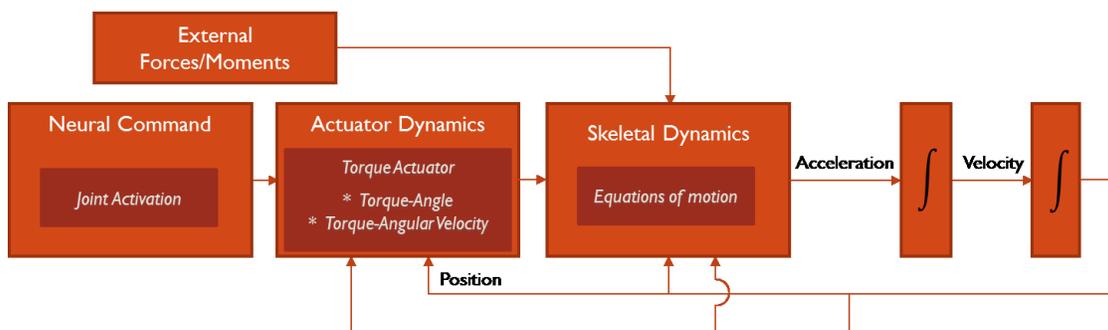


Figure 2.1 Framework of forward dynamic simulation

As an input to forward dynamics simulation, a time history of joint activation is needed. Due to the complexity of the musculoskeletal system, it is very challenging to find the activations which generate a realistic movement for the specified task. To solve for activation levels, optimization techniques such as simulated annealing are utilized to minimize or maximize an objective function. There are two common methods to define an objective function. The first one is called the “tracking method” which is based on the minimizing the error between the experimentally collected and simulation kinematics and/or kinetics. This method is preserving the experimental movement but cannot be utilized if the experimental data is not available. The second method is

called “performance based” objective function which is usually used when the task has specific goals such as minimizing metabolic energy used during gait (Crenshaw and Grabiner 2014). This method does not require the experimental data but defining the proper and realistic objective function can be very difficult.

The framework described above involves the torque actuation which is less complicated than the muscle actuation and require fewer parameters to define the actuator model since they represent the net effect of multiple muscles that cross a joint. Torque actuator uses the prescribed activation, the maximum torque available at each joint, and the current state of the system (angle and angular velocity) to determine the amount of joint torque created by the model at each instant in time. The defined joint torque along with the external torques and forces are then entered in to the skeletal dynamics (equations of motion) to find the acceleration and then integrated to find the velocity and position of the system at the next time step. The equations of motion can be derived by using Lagrange method. For complicated systems, computer software such as Autolev software (OnLine Dynamics, Inc., Sunnyvale, CA) can be used to find the equations of motion.

## **2.7 Summary**

The high prevalence of obesity and an increasing rate of older population in the United States poses some major concerns due to association of both conditions to serious health problems. Besides problems such as hypertension, diabetes, and heart disease, both obesity and aging are associated with limited mobility and higher risk of falls. The increasing prevalence of the obesity among older adults adds to the complications of the problem since obesity can exacerbate the

health issues related to aging. The broad goal of the present research was to investigate lower extremity strength as one of the main contributing factors to impaired mobility and higher risk of falls among individuals who are obese and older individuals. This was accomplished by investigating the effect of obesity and age on lower extremity strength (Chapter 3), the impact of obesity-related variation of muscle strength on gait (Chapter 4), and the influence of age-related variation of muscle strength on non-stepping balance recovery (Chapter 5). The results determined specific muscle weaknesses among obese and older individuals which would help in designing interventions to improve the mobility and decrease the risk of falls among these individuals.

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## **Chapter 3 - Relative Strength at the Hip, Knee, and Ankle is lower among Younger and Older Females Who Are Obese**

### **3.1 Abstract**

The mobility of individuals who are obese can be limited compared to their healthy-weight counterparts. Lower limb strength has been associated with mobility, and reduced strength may contribute to mobility limitation among individuals who are obese. However, our understanding of the effects of obesity on lower limb strength is limited. The purpose of this study was to investigate the effects of obesity and age on extension and flexion strength at the hip, knee, and ankle. Using a cross-sectional design, 10 younger (18-30 years) healthy-weight (body mass index 18-24.9 kg/m<sup>2</sup>), 10 younger obese (body mass index >30 kg/m<sup>2</sup>), 10 older (65-80 years) healthy-weight, and 10 older obese female participants performed isokinetic maximum voluntary contractions in ankle plantar flexion (PF), ankle dorsiflexion (DF), knee extension (KE), knee flexion (KF), hip extension (HE), and hip flexion (HF). Absolute strength among obese participants was 29% higher in DF ( $p=0.002$ ), 27% higher in KE ( $p=0.004$ ), and 23% higher in HF ( $p=0.001$ ) compared to healthy-weight participants. Strength relative to body mass among obese participants was 31% lower in PF ( $p<0.001$ ), 14% lower in DF ( $p=0.042$ ), 16% lower in KE ( $p=0.015$ ), 27% lower in KF ( $p<0.001$ ), 29% lower in HE ( $p<0.001$ ), and 19% lower in HF ( $p=0.001$ ). Obese females exhibited lower relative strength at the ankle and hip, similar to the lower relative strength exhibited at the knee. Obese females also exhibited higher absolute strength, but only for 3 of 6 lower limb exertions investigated. This lack of uniformity across the 6 exertions is likely due to the still unclear underlying biomechanical mechanism responsible for

these strength differences, which may also be influenced by aging. The effects of obesity on lower limb strength were also generally consistent between the 2 age groups investigated.

### **3.2 Introduction**

Over one-third of adults in the United States are obese (Ogden et al. 2014). One of the many problems associated with obesity is limited mobility (Hue et al. 2007; Koster et al. 2007; Vincent et al. 2010) which can include difficulty rising from a chair (Galli et al. 2000; Pataky et al. 2014), difficulty ascending or descending from stairs (Stickles et al. 2001), lower gait speed (Lai et al. 2008; Spyropoulos et al. 1991), and poorer balance (Dutil et al. 2013; Jadelis et al. 2001; Menegoni et al. 2009). This limited mobility has been related to having less lower limb strength relative to body mass, and higher lower limb strength demands due to having additional body mass (Capodaglio et al. 2010).

Individuals who are obese exhibit altered lower limb strength compared to healthy-weight individuals (Capodaglio et al. 2009; Hulens et al. 2001; Lafortuna et al. 2005; Maffiuletti et al. 2007; Miyatake et al. 2000). Lower limb strength can be characterized in terms of absolute strength and relative strength. Absolute strength can be defined as the maximum force or net muscle moment that can be generated at a joint, and is expressed in units of Newtons or Newton-meters. Relative strength is typically determined as absolute strength divided by body mass (Capodaglio et al. 2009; Miyatake et al. 2000) or fat-free mass (Hulens et al. 2001; Maffiuletti et al. 2007). Knee extensor strength among individuals who are obese is higher when expressed as absolute strength (Capodaglio et al. 2009; Hulens et al. 2001; Lafortuna et al. 2005; Maffiuletti et al. 2007), but lower when expressed as relative strength (Capodaglio et al. 2009; Maffiuletti et al.

2007). The higher absolute strength is thought to be a neuromuscular adaptation to the chronic exposure to increased body weight (Capodaglio et al. 2009; Hulens et al. 2001), since the knee extensors play a major role in supporting the body against gravity while standing and walking. Consistent with this hypothesis, absolute strength of the knee flexors, which do not play a major role in supporting the body against gravity while standing or walking, is not higher among individuals who are obese (Capodaglio et al. 2009; Hulens et al. 2001). Despite the importance of hip and ankle strength on (Barrett and Lichtwark 2008; Rantanen et al. 1998), no studies to our knowledge have investigated obesity-related differences in strength at these joints.

The prevalence of obesity among adults over the age of 60 has increased from 31% in 2003-2004 to 35% in 2011-2012 (Ogden et al. 2014). This growth in the older obese population provides motivation for understanding how obesity affects strength among older adults, particularly because of the association between lower limb strength and mobility (Payette et al. 1998). Although absolute knee extensor strength is higher among individuals who are obese, the typical loss of strength associated with aging (Dean et al. 2004; Hunter et al. 2000), may result in obesity-related differences in absolute strength being smaller among older, compared to young, adults. Although relative knee extensor strength is lower among individuals who are obese, the typical loss of strength associated with aging may result in obesity-related differences in relative strength being larger among older, compared to young, adults. Obesity is associated with higher absolute knee extensor strength (Rolland et al. 2004), and lower relative knee extensor strength among older adults (Miyatake et al. 2000), but no comparison of these effects were performed between young and older adults. Additional studies are needed to more fully understand the interaction of obesity and age on lower limb strength.

The purpose of this study was to investigate the effects of obesity and age on extension and flexion strength at the hip, knee, and ankle. These data will help elucidate whether the effects of obesity on strength differs between lower limb joints, between extension and flexion directions, and between younger and older adults. Such information may be useful in developing strength training interventions among obese/older adults for maintaining mobility. We focused on females because of their higher prevalence of obesity (Ogden et al. 2014), higher prevalence of obesity-related mobility limitation (Vincent et al. 2010), and higher rate of falls and fall-related injuries (Stevens and Sogolow 2005). The specific hypotheses investigated were that 1) absolute strength would be higher among obese compared to the healthy-weight females, 2) relative strength would be lower among obese compared to the healthy-weight females, 3) the effect of obesity on absolute strength would be smaller among older females compared to younger females, and 4) the effect of obesity on relative strength would be larger among older females compared to younger females.

### **3.3 Methods**

This cross-sectional study was performed in a university biomechanics research laboratory over a period of 11 months. Forty adult females completed the study, including 10 younger (18-30 years) and healthy-weight (body mass index, or BMI, 18-24.9 kg/m<sup>2</sup>), 10 younger and obese (BMI >30 kg/m<sup>2</sup>), 10 older (65-80 years) and healthy-weight, and 10 older and obese (Table 3.1). The number of participants in each group was based upon a sample size analysis using data from two studies investigating obesity-related differences in isokinetic knee extension strength among young adults (Capodaglio et al. 2009; Maffioletti et al. 2007). These analyses indicated 10 healthy-weight and 10 obese participants were needed to detect a main effect of obesity with

0.70 power. To be conservative, we recruited 20 healthy-weight and 20 obese participants (equally split between younger and older age groups). Participants were recruited from the university and local community using web and email announcements, flyers, and newspaper advertisements. Participants were required to pass a screening to exclude individuals with self-reported neurological, cardiac, or musculoskeletal conditions that might jeopardize their safety during testing. In addition, all participants were body mass stable ( $<2.3$  kg) for the prior 6 months, and had no obvious balance problems. Participants completed the Godin leisure-time exercise questionnaire (Godin and Shephard 1997), to quantify their habitual physical activity level, given that higher physical activity levels may influence lower limb strength. Participants who performed  $\geq 1$  hour of moderate exercise more than 3-4 times a week were excluded from the study. Body fat percentage was measured using a Lange skinfold caliper (Cambridge Scientific Industries, Cambridge, MA, USA) at 4 locations including: superficial to the triceps and biceps, inferior to the scapula, and superior to the lateral iliac crest. These measurements were summed, and body fat percentage was predicted from this sum according to the caliper manufacturer's specifications. The study was approved by the university Institutional Review Board, and written informed consent was obtained from all participants prior to participation.

Participants completed two sessions. Knee strength was measured during the first session, and ankle and hip strength were measured during the second session. All strength measurements were performed on the right lower limb, which was the preferred limb to kick a ball. The strength testing protocol was adapted from prior work (Anderson et al. 2007). Strength was measured as the maximum net muscle moment during concentric isokinetic maximum voluntary

contractions (MVCs). These MVCs were performed using a Biodex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, NY). The experimental set-up can be seen in Figure 3.1.

Table 3.1 Participant characteristics (median (IQR))

	YH	YO	OH	OO
<b>Age (years)</b>	20.5 (4.5)	22.0 (5.8)	69.0 (8.8)	68.5 (8.5)
<b>Height (cm)</b>	165 (5.7)	168.5(7.3)	161.0 (10.4)	161.8 (9.0)
<b>Mass (kg)</b>	62.1 (10.3)	94.1 (9.5)	59.0 (8.8)	87.8 (8.5)
<b>BMI (kg/m<sup>2</sup>)</b>	22.4 (3.2)	33.1 (4.3)	22.2 (5.8)	32.5 (4.2)
<b>Body fat (%)*</b>	23.4 (1.9)	35.3 (4.3)	33.8 (7.7)	42.6 (4.2)
<b>Godin score</b>	32.0 (19.8)	24.0(30.0)	23.5 (20.0)	25.5 (20.0)

YH = Young and Healthy-weight, YO = Young and Obese, OH = Old and Healthy-weight, OO = Old and Obese

\*determined from Lange skinfold caliper (Lange skinfold caliper; Cambridge Scientific Industries, Cambridge, Massachusetts, USA)

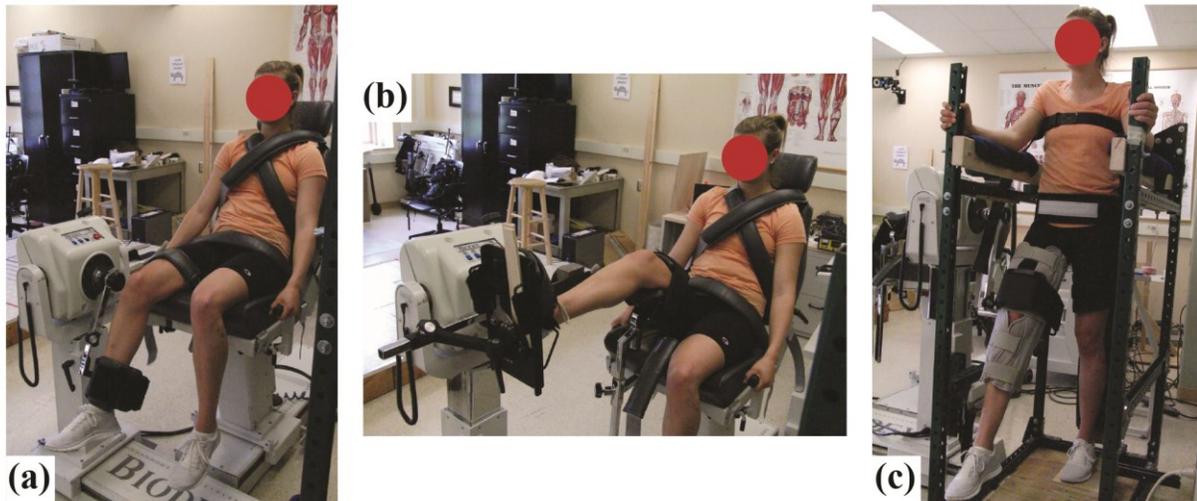


Figure 3.1 Experimental set-up for strength measurements at the (a) knee, (b) ankle, and (c) hip.

In the first session, knee strength was measured while participants were secured, using straps, in a seated posture with their hip flexed approximately 70 degrees (Anderson et al. 2007). Relaxed trials were performed first to measure the passive elastic/gravitational moment over the entire joint range of motion. Participants were instructed to remain relaxed while the Biodex attachment moved at 5 degrees/sec through the range of motion at least 3 times. Participants then performed concentric isokinetic MVCs in knee extension (KE) and knee flexion (KF) at 75 deg/sec. This velocity was chosen based upon prior studies (Anderson et al. 2007; Hulens et al. 2001; Maffiuletti et al. 2007). A total of 4 MVCs were completed for each exertion direction. Prior to data collection, participants performed 1 practice trial for each exertion.

In the second session, ankle and hip strength were measured in this order. The general testing protocol was similar to that used for the knee, but with different body positions and isokinetic velocities. Concentric isokinetic MVCs in ankle plantar flexion (PF) and dorsiflexion (DF) were performed at 60 degrees/sec in a seated position, while the knee and hip were flexed 50 degrees and 80 degrees, respectively. Concentric isokinetic MVCs in hip extension (HE) and hip flexion (HF) were performed at 60 degrees/sec in a standing position, while the knee was held in a near fully extended position. These joint angles and angular velocities were chosen based upon prior studies (Anderson et al. 2007; Perry et al. 2007).

Joint angle, angular velocity, and moment were sampled from the dynamometer at 200 Hz and low-pass filtered at 5 Hz (4<sup>th</sup>-order Butterworth filter). The passive elastic/gravitational moment was estimated by fitting a curvilinear line (least square) to moment data from relaxed trials throughout the range of motion, and was subtracted from each MVC trial (Anderson et al. 2007).

The isokinetic region was identified for each trial, defined by where the acceleration was negligible, and the peak moment in that region was determined. The maximum moment across the 4 trials for each joint/exertion direction was used as the absolute strength for that joint/exertion direction. Relative strength was determined by normalizing this strength measurement to body mass. All post processing was performed in Matlab (The MathWorks Inc., Natick, MA, USA).

Separate 2-way analyses of covariance were used to determine the effects of obesity (healthy-weight or obese), age (younger or older), and their interaction on each strength measurement. Strength measurements included absolute and relative strength in PF, DF, KE, KF, HE, and HF. Godin score was used as a covariate. The first and second hypotheses were tested using the main effect of obesity. The main effects of age are also reported, but these were not a major focus since they have been reported in previous studies. The third and fourth hypotheses were tested using the age x obesity interaction. In the event of a significant obesity x age interaction simple effects testing was used to assess the effects of obesity within each age group, and the effects of age within each obesity group. No data was missing during the analyses. Effect sizes were reported using the partial eta squared ( $\eta_p^2$ ). Percent differences and absolute differences reported in the Results were least squares means to account for the influence of the Godin score on strength measurements. Statistical analyses was performed using JMP Pro 10 (SAS Institute, Inc., Cary, NC, USA) with a significance level of  $p < 0.05$ .

### 3.4 Results

Absolute and relative strength among younger and older groups are shown in Figure 3.2 and Figure 3.3. Several main effects of obesity were observed on both absolute and relative strength (Table 3.2). No strength variables exhibited a significant obesity by age interaction. Absolute strength among obese participants was 29% (4.7 Nm) higher in DF ( $p = 0.002$ ), 27% (24.1 Nm) higher in KE ( $p = 0.004$ ), and 23% (18.5 Nm) higher in HF ( $p < 0.001$ ) compared to healthy-weight participants. Relative strength among obese participants was 31% (0.26 Nm/kg) lower in PF ( $p < 0.001$ ), 14% (0.04 Nm/kg) lower in DF ( $p = 0.042$ ), 16% (0.24 Nm/kg) lower in KE ( $p = 0.015$ ), 27% (0.21 Nm/kg) lower in KF ( $p < 0.001$ ), 29% (0.48 Nm/kg) lower in HE ( $p < 0.001$ ), and 19% (0.25 Nm/kg) lower in HF ( $p = 0.001$ ) compared to healthy-weight participants.

Table 3.2 Main effect of obesity, age and obesity by age interaction on different exertions ( $p$ -value (effect size))

	Absolute Strength (N·m)			Relative Strength (N·m/kg)		
	Obese vs. healthy-weight	Older vs. young adults	Obesity × age	Obese vs. healthy-weight	Older vs. young adults	Obesity × age
<i>Plantarflexor</i>	0.54 (0.011)	0.19 (0.048)	0.13 (0.065)	<0.01 (0.292)	0.83 (0.001)	0.26 (0.036)
<i>Dorsiflexor</i>	<0.01 (0.252)	<0.01 (0.401)	0.35 (0.025)	0.04 (0.113)	<0.01 (0.324)	0.82 (0.001)
<i>Knee Extensor</i>	<0.01 (0.214)	<0.01 (0.429)	0.32 (0.029)	0.01 (0.158)	<0.01 (0.383)	0.80 (0.002)
<i>Knee Flexor</i>	0.087 (0.082)	<0.01 (0.570)	0.36 (0.025)	<0.01 (0.390)	<0.01 (0.467)	0.47 (0.015)
<i>Hip Extensor</i>	0.45 (0.016)	0.05 (0.103)	0.44 (0.017)	<0.01 (0.299)	0.21 (0.045)	0.88 (0.001)
<i>Hip Flexor</i>	<0.01 (0.275)	<0.01 (0.367)	0.07 (0.089)	<0.01 (0.265)	<0.01 (0.228)	0.6 (0.008)

Several main effects of age were also observed on both absolute and relative strength (Table 3.2). Absolute strength among older participants was 30% (6.6 Nm) lower in DF ( $p < 0.001$ ), 33% (39.7 Nm) lower in KE ( $p < 0.001$ ), 35% (20.9 Nm) lower in KF ( $p < 0.001$ ), and 22% (22.8 Nm) lower in HF ( $p < 0.001$ ) compared to younger participants. Relative strength among older participants was 25% (0.07 Nm/kg) lower in DF ( $p < 0.001$ ), 27% (0.44 Nm/kg) lower in KE ( $p < 0.001$ ), 31% (0.25 Nm/kg) lower in KF ( $p < 0.001$ ), and 17% (0.23 Nm/kg) lower in HF ( $p = 0.003$ ) compared to younger participants.

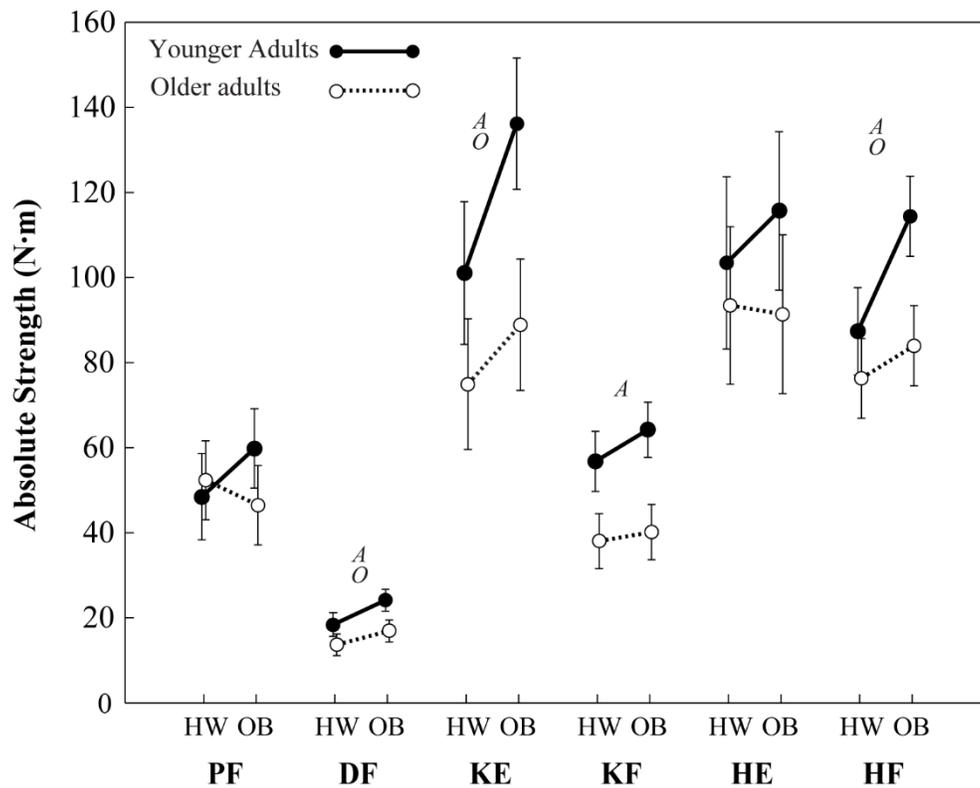


Figure 3.2 Least squares means of absolute strength (error bars indicates upper and lower 95<sup>th</sup> confidence intervals). Note: HW=Healthy-Weight; OB=Obese; A = main effect of age; O = main effect of obesity; ( $p < 0.05$ ). PF = plantar flexors; DF = dorsiflexors; KE = knee extensors; KF = knee flexors; HE = hip extensors; and HF = hip flexors.

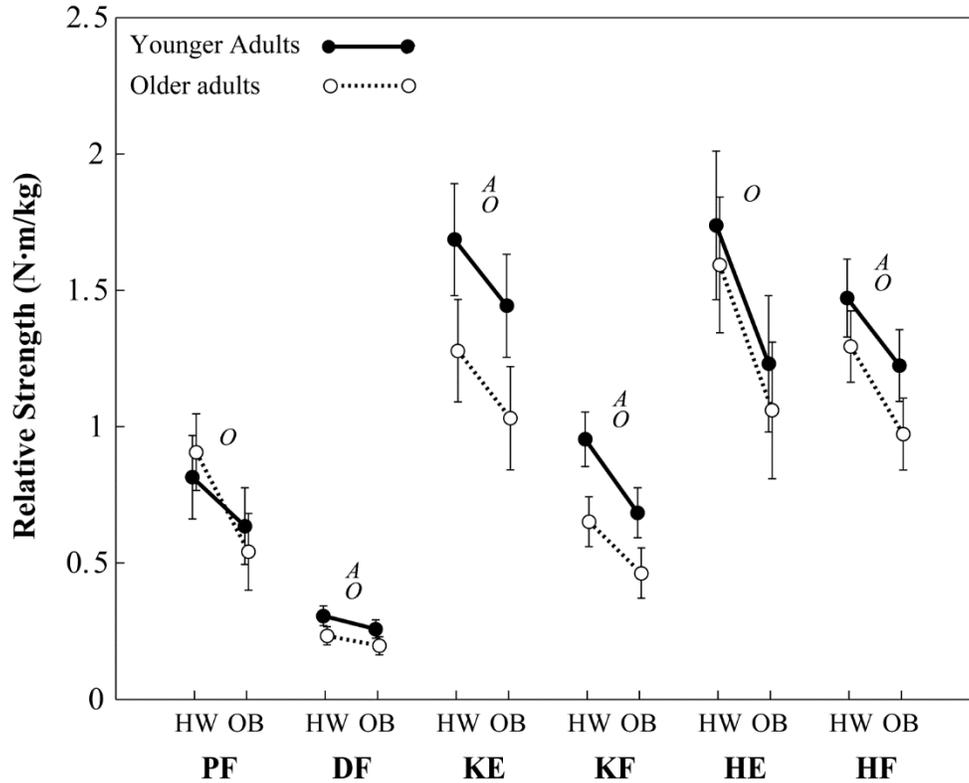


Figure 3.3 Least squares means of relative strength (error bars indicates upper and lower 95<sup>th</sup> confidence intervals). Note: HW=Healthy-Weight; OB=Obese; A = main effect of age; O = main effect of obesity; ( $p < 0.05$ ). PF = plantar flexors; DF = dorsiflexors; KE = knee extensors; KF = knee flexors; HE = hip extensors; and HF = hip flexors.

### 3.5 Discussion

The purpose of this study was to investigate the effects of obesity and age on extension and flexion strength at the hip, knee, and ankle. Our first hypothesis was that absolute strength would be higher among obese compared to health-weight females. This hypothesis was supported for DF, KE, and HF, since all 3 were higher among obese participants. Our second hypothesis was that relative strength would be lower among obese compared to healthy-weight females. This hypothesis was supported for all 6 joint/extension-flexion combinations since relative strength

was lower among obese participants. Our third hypothesis was that the effect of obesity on absolute strength would be smaller among older females compared to younger females. This hypothesis was not supported for any joint/extension-flexion combination since no age x obesity interaction effect was observed. Our fourth hypothesis was that the effect of obesity on relative strength would be larger among older females compared to younger females. This hypothesis was also not supported. These results provide 3 general findings. First, obese females exhibited lower relative strength at the ankle (PF and DF) and hip (HE and HF) similar to the lower relative strength they exhibited at the knee (KE and KF) found here and reported elsewhere. Second, the magnitude of these differences in relative strength, and the existence of differences in absolute strength, differed between the 6 joints/extension-flexion combinations investigated. Third, the effects of obesity on lower limb strength were generally consistent between the 2 age groups investigated.

Strength differences at the knee found here were consistent with previous reports, but differed in magnitude. Absolute KE strength was 27% higher among obese vs. healthy-weight females here, which was slightly larger than 12-20% differences reported elsewhere (Capodaglio et al. 2009; Hulens et al. 2001; Maffiuletti et al. 2007; Miyatake et al. 2000). Relative KE strength was 16% lower among obese vs. healthy-weight females here, which is moderate compared to 32% and 7% differences reported elsewhere (Capodaglio et al. 2009; Maffiuletti et al. 2007; Miyatake et al. 2000). Absolute KF strength did not differ between obese and healthy-weight females here, or between obese and healthy-weight participants elsewhere (Capodaglio et al. 2009; Hulens et al. 2001). Relative KF strength was 27% lower among obese vs. healthy-weight females here, which was somewhat smaller than the 41% difference reported elsewhere (Capodaglio et al. 2009).

There are at least 3 factors that may contribute to the differences in magnitude between the current and prior studies. The first factor is a difference in age of participants between these studies. General age ranges in the noted prior studies included exclusively younger adults (e.g. 20-40 years) (Capodaglio et al. 2009; Maffiuletti et al. 2007), or essentially the entire adult age range (e.g. 20 to 79 years) (Hulens et al. 2001; Miyatake et al. 2000). The second factor is a difference in BMI of participants between these studies. Prior studies have included obese participants with BMIs of 31-43 (Capodaglio et al. 2009), over 35 (Maffiuletti et al. 2007), and over 26.4 (Miyatake et al. 2000), whereas the present study included obese participants with a range in BMI of 30-37. The third factor is a difference in gender of participants between these studies. Prior studies have included only female participants (Capodaglio et al. 2009; Hulens et al. 2001), only male participants (Maffiuletti et al. 2007), and both genders (Miyatake et al. 2000), so the effects of obesity on lower limb strength may be influenced by gender. Despite these discrepancies in participant characteristics and the magnitude of strength differences between studies, the differences in absolute and relative KE and KF between obese and healthy-weight participants appears relatively consistent across studies.

The underlying cause for the higher absolute strength among obese participants remains unclear. The higher KE absolute strength among obese participants has been attributed to a neuromuscular training effect from chronic exposure to higher body weight (Capodaglio et al. 2009; Hulens et al. 2001). This seems reasonable given that the KE muscle group plays a major role in supporting the body against gravity, for example, while standing and walking. However, DF and HF also exhibited higher absolute strength among obese participants, but these exertions are not typically associated with body support during standing or gait (Sadeghi et al. 2001).

Instead, PF and HE have been associated with body support during gait (Honeine et al. 2013; Sadeghi et al. 2001) but did not exhibit higher absolute strength among obese participants. Peak net muscle moments in HE and HF are fairly similar during gait (Eng and Winter 1995), suggesting that any increase in strength, if mediated by net muscle moments during gait, would not seem to differ between these 2 exertions. This seems inconsistent with our results that absolute strength increased for HF, but not HE. One possible explanation for these findings at the hip is that the higher HF absolute strength among obese participants could be a secondary effect of the higher KE absolute strength since the rectus femoris muscle contributes to both KE and HF. However, no such biarticular muscle can help to explain the unexpected increase in DF and not PF. Peak net muscle moments in DF during standing and gait are typically much smaller than in PF, and not traditionally thought to play a major role in supporting the body against gravity. It is also interesting to note that the 3 exertions that exhibited higher absolute strength among obese participants (HF, KE, DF) also exhibited smaller effect sizes for relative strength when compared to their respective antagonist muscle groups at the same joint (HE, KF, PF). Although no obesity by age interaction was found to be statistically significant, our mean results (Figure 3.2) suggest a greater likelihood of an obesity by age interaction for PF and HE as indicated by lower mean absolute strength among older obese participants vs. older healthy-weight participants. This opposite trend compared to that seen among young participants may have moderated the overall main effect of obesity for these 2 exertions. Indeed, prior work has shown older adults to exhibit a neuromuscular adaptations during gait that deemphasize the use of PF net muscle moments and emphasize the use of HE net muscle moments (DeVita and Hortobagyi 2000). In light of these results, chronic exposure to higher body weight may still play a major role in explaining the higher absolute strength among individuals who are obese, but that

neuromuscular changes associated with aging may modify these effects among older adults. Additional research is needed to better understand the underlying biomechanical mechanisms responsible for the higher absolute strength associated with obesity.

A novel contribution of this study is the finding that lower limb relative strength was lower among obese vs. healthy-weight, not only at the knee (KE and KF), but also at the hip (HE and HF) and ankle (PF and DF). These findings are important because ankle and hip strength, in addition to knee strength, are important for mobility (Barrett and Lichtwark 2008; Rantanen et al. 1998), and suggest that strength training interventions aimed at improving or maintaining mobility with obesity should include not only the knee based upon prior reports, but also the hip and ankle. These data also provide normative strength values for young and older obese and healthy-weight adults that could be used to identify individuals with potentially more drastic strength differences. However, the relative importance of the joint/exertions for improving or maintaining mobility among individuals who are obese remains unclear, and would be useful to know to focus strength training interventions on the joint/exertions that are most critical for mobility.

Age-related differences in strength were in general agreement with prior work. Harbo et al. (2012) reported isokinetic strength at the hip, knee, and ankle for different age and gender groups. Absolute strength of healthy-weight females of age (mean  $\pm$  standard deviation) 63 $\pm$ 3 years was, compared to healthy-weight females of age 25 $\pm$ 4 years, 24% lower in PF, 18% lower in DF, 26% lower in KE, 35% lower in KF, 16% lower in HE, and 14% lower in HF. These differences in absolute strength compare favorably to the knee and hip data reported here when

comparing older healthy-weight participants to younger healthy-weight participants. For example, differences here in KE (30%), KF (32%), HE (11%), and HF (15%) exhibited a similar magnitude to those by Harbo et al. (2012). Differences here in PF (2%) and DF (28%) between older and younger participants did not agree as well between studies, and may be due to differences in participant characteristics and strength protocols.

This study had a few limitations that should be acknowledged. First, and as with all cross-sectional experimental designs, the effects of obesity that we report may not be solely attributed to obesity. Other factors, for example, may include differences in motivation level to generate maximum contractions, or lower limb muscle mass secondary to differences in body shape/mass distribution. Second, the number of subjects in each group was based upon a sample size analysis using knee extensor strength data from younger adults. Because the effects of obesity on some joint/direction combinations may have been smaller, we may not have had sufficient statistical power to detect all effects that we investigated. Third, the PF strength measurement set-up, while the standard set-up recommended by the dynamometer manufacturer, was susceptible to influence by KE effort, although participants were instructed to only use their ankle muscles. Fourth, the results presented here do not necessarily generalize to other populations, or other types of muscle contractions, besides those tested. Fifth, only sagittal plane strength was measured, and the findings may differ for other movements.

### **3.6 Conclusions**

Obese females exhibited lower relative strength at the ankle and hip, similar to the lower relative strength exhibited at the knee. Obese females also exhibited higher absolute strength, but only

for 3 of 6 lower limb exertions investigated. This lack of uniformity across the 6 exertions is likely due to the still unclear underlying biomechanical mechanism responsible for these strength differences, which may also be influenced by aging. The effects of obesity on lower limb strength were also generally consistent between the 2 age groups investigated. These findings provide a better understanding of how obesity influences lower limb strength, and lends insight on the causes of limited mobility among individuals who are obese.

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## **Chapter 4 - Relative Effort during Gait is higher among Individuals Who Are Obese**

### **4.1 Abstract**

Obese and older individuals exhibit limited mobility during activities of daily living possibly due to greater effort required to perform a task relative to their maximum available capacity. Since walking is a major activity performed daily, the purpose of this was to investigate the effects of obesity, age and their interaction on relative effort at the hip, knee, and ankle during gait. Four groups of participants 10 younger (18-30 years) healthy-weight (body mass index 18-24.9 kg/m<sup>2</sup>), 10 younger obese (body mass index >30 kg/m<sup>2</sup>), 10 older (65-80 years) healthy-weight, and 9 older obese female participated in the study which included measurement of maximum available net muscle moment (NMM) of lower extremity and gait trials under self-selected and controlled conditions (1.5 m/sec gait speed and 0.65 m step length). The peak relative effort for each joint/exertion was expressed by peak NMM during gait as a percentage of the maximum available NMM. During controlled gait, peak relative effort exhibited obesity-related differences including 11.6% higher in knee extension ( $p=0.009$ ) and 35% higher in ankle plantar flexion ( $p=0.008$ ). During controlled gait, peak relative effort exhibited obesity-related differences including 12.5% higher in hip flexion ( $p=0.043$ ), 16.4% higher in knee flexion ( $p=0.003$ ), and 36% higher in ankle plantar flexion ( $p=0.003$ ). This higher relative effort in hip, knee, and especially in the ankle can be a contributing factor to compromised walking ability among obese individuals. The findings can help in developing targeted interventions to maintain and improve mobility among obese individuals.

## 4.2 Introduction

About one-third of the adult population in the United States is obese (Ogden et al. 2014). Obesity is linked to mobility limitations (Koster et al. 2007; Vincent et al. 2010) which can include difficulty rising from a chair (Galli et al. 2000; Pataky et al. 2014), difficulty ascending or descending from stairs (Stickles et al. 2001), impaired dynamic balance (Dutil et al. 2013; Jadelis et al. 2001; Menegoni et al. 2009), and compromised walking ability (Angleman et al. 2006; Lamb et al. 2000; Stenholm et al. 2009). Compromised walking ability includes slower gait speed (Lai et al. 2008; Spyropoulos et al. 1991), self-reported difficulty in walking long (500 m) distances (Stenholm et al. 2007), shorter step lengths (Hulens et al. 2003; Page et al. 2014), longer stance phase duration, and longer double support phase (Browning and Kram 2007; DeVita and Hortobágyi 2003; Spyropoulos et al. 1991). Despite the noted obesity-related changes in gait, obesity did not impact the joint kinematics at the hip and knee in sagittal plane during stance (Browning and Kram 2007; Lai et al. 2008; Spyropoulos et al. 1991), but it was associated with reduced ankle plantar flexion throughout the stance phase (Spyropoulos et al. 1991).

Reduced lower extremity strength has been identified as an underlying factor for mobility limitations (Jadelis et al. 2001; Stenholm et al. 2009; Visser et al. 2005). Although absolute lower extremity strength has been reported to be higher among obese individuals in knee extension (Capodaglio et al. 2009; Hulens et al. 2001; Koushyar et al. 2016; Lafortuna et al. 2005), hip flexion, and ankle dorsiflexion (Koushyar et al. 2016) compared to healthy-weight individuals, relative lower extremity strength (absolute strength normalized by body mass) has been reported to be lower among obese in flexion and extension at the hip, knee, and ankle

compared to healthy-weight individuals (Capodaglio et al. 2009; Koushyar et al. 2013; Maffiuletti et al. 2007; Miyatake et al. 2000).

Reduced relative lower extremity strength among obese individuals may help explain obesity-related compromised walking ability (Browning and Kram 2007; Lerner et al. 2014). In an effort to better understand the importance, and potentially limiting effect, of lower extremity strength on mobility, net muscle moment (NMM) have been expressed as a percentage of maximum available NMM (measured during strength testing) during sit-to-stand (Bieryla et al. 2009; Hortobágyi et al. 2003; Hughes et al. 1996; Ioannis et al. 2014), stair ascent and descent (Hortobágyi et al. 2003; Samuel et al. 2013), and gait (Anderson and Madigan 2014; Requião et al. 2005; Samuel et al. 2013). NMM expressed as a percentage of maximum available NMM estimates the percentage of overall strength capacity utilized, and has been referred to as relative effort (Hortobágyi et al. 2003; Ioannis et al. 2014). Given that obese individuals exhibit both higher lower extremity NMM during gait (Browning and Kram 2007) and higher absolute maximum available strength (Capodaglio et al. 2009; Hulens et al. 2001; Koushyar et al. 2013), it is unclear if obese individuals use greater relative effort during gait than healthy-weight individuals. Identifying such differences could help clarify the underlying factors by which obesity compromised walking ability and limits mobility.

Aging has also been associated with mobility limitations, represented by a decline in the ability to perform the activities of daily living (Grabiner and Enoka 1995; Horak et al. 1989; Vandervoort 1992). During gait, for example, aging is associated with slower gait speed, shorter step length and less range of motion at hip, knee, and ankle joint (Elble et al. 1991; Hageman and

Blanke 1986; JudgeRoy et al. 1996; Kerrigan et al. 1998). One of the reasons for these mobility limitations can be that older adults require a higher relative effort to perform activities of daily living (Anderson and Madigan 2014; Hortobágyi et al. 2003; Nadeau et al. 1999; Requião et al. 2005). Hortobágyi et al. (2003), for example, reported higher relative effort among older adults during chair rise, and while ascending and descending stairs compared to young adults. Anderson and Madigan (2014) reported relative effort of the ankle plantar flexors to be close to 100% among older adults while walking at a hurried speed, indicating near full effort was being exerted, and little or no capacity to further increase plantar flexor NMM. Older adults who are obese may use a particularly high relative effort during activities of daily living due to age-related loss of muscle strength and higher NMM as a result of increased body mass.

The purpose of this study was to investigate the effects of obesity, age and their interaction on relative effort at the ankle, knee, and hip during gait. We focused on females because of their higher prevalence of obesity (Ogden et al. 2014) and higher prevalence of obesity-related mobility limitation (Vincent et al. 2010). We hypothesized 1) peak relative effort would be higher at the ankle, knee and hip among obese females, and 2) the effect of obesity on relative effort would be larger among older females compared to young females. The results from this study will provide insight into the underlying factors which contribute to the mobility limitation associated with obesity, and could contribute to strategies to improve and maintain mobility among obese individuals.

### 4.3 Methods

Thirty nine adult females completed the study including 10 young (18-30 years) healthy-weight (body mass index 18-24.9 kg/m<sup>2</sup>), 10 young obese (body mass index >30 kg/m<sup>2</sup>), 10 older (65-80 years) healthy-weight, and 10 older obese (Table 4.1). Participants were recruited from the university and local community using web and email announcements, flyers, and newspaper advertisements. All participants were required to pass a medical screening to exclude individuals with self-reported neurological, cardiac, or musculoskeletal conditions, balance problems, or more than 2.3 kg change in body mass over the past six months. Participants also completed the Godin leisure-time exercise questionnaire (Godin and Shephard 1997) to quantify their physical activity level. The study was approved by the university Institutional Review Board, and written informed consent was obtained from all participants prior to participation.

Table 4.1 Subject characteristics (mean  $\pm$  SD)

<b>Group</b>	<b>YH</b>	<b>YO</b>	<b>OH</b>	<b>OO</b>
<b>Age (years)</b>	21.7 $\pm$ 3.3	23.1 $\pm$ 3.8	68.8 $\pm$ 5.5	68.2 $\pm$ 4.2
<b>Height (cm)</b>	164.6 $\pm$ 4.3	166.4 $\pm$ 6.1	161.5 $\pm$ 6.3	163.1 $\pm$ 5.0
<b>Mass (kg)</b>	60.9 $\pm$ 5.7	93.3 $\pm$ 8.9	59.2 $\pm$ 7.9	86.4 $\pm$ 5.4
<b>BMI (kg/m<sup>2</sup>)</b>	22.5 $\pm$ 1.8	33.7 $\pm$ 2.9	22.7 $\pm$ 3.0	32.5 $\pm$ 2.0
<b>Godin score</b>	36.8 $\pm$ 16.9	33.5 $\pm$ 20.6	27.2 $\pm$ 15.2	27.2 $\pm$ 13.8

YH = Young and Healthy-weight, YO = Young and Obese, OH = Old and Healthy-weight, OO = Old and Obese

Participants completed two experimental sessions. In the first session, participants performed all gait trials as described below, and performed strength measurements at the knee. In the second session (completed within one week of the first session), participants performed strength measurements at the ankle and the hip.

In the first session, participants first donned spandex shorts and were given the same brand of standardized walking shoes. Participants then completed gait trials on a 10-meter walkway with self-selected and controlled gait. The controlled trials were included in an attempt to control for the confounding effects of gait speed and step length on gait kinetics (Allet et al. 2011; Kerrigan et al. 1998). During self-selected trials, participants were instructed to walk naturally and look straight ahead. Up to seven practice trials were performed to acclimatize participants to the experimental surroundings, and to identify the proper starting position along the walkway so their right foot naturally and consistently landed on a force platform integrated in the middle of the walkway (all but one young healthy-weight and one older healthy-weight participant were right foot dominant). Five trials were then performed with proper foot placement on the force platform. During controlled trials, speed was constrained to  $1.5 \pm 0.05$  m/s (via verbal feedback after each trial using a passive reflective marker placed on the participant's back), and the step length was constrained to 0.65 m by asking participants to step on markings spaced this distance along the length of the walkway (Figure 4.1). The controlled speed and step length were representative of self-selected speed and step length among young adults (JudgeRoy et al. 1996; Kerrigan et al. 1998; Silder et al. 2008). Five trials were then performed.



Figure 4.1 Walking under controlled gait

After completing the gait trials, strength measurements were collected in knee extension (KE) and knee flexion (KF) on a Biodex system 3 dynamometer (Biodex Medical Systems, Inc., Shirley, NY). Measurements were collected from the right knee while in a seated position with the hip flexed 70 deg. Relaxed trials were first performed to measure the passive elastic/gravitational moment over the entire range of motion. Participants were instructed to remain relaxed while the Biodex attachment moved at 5 deg/sec throughout the joint range of motion at least three times. Next, participants performed isometric maximum voluntary contractions (MVCs) in KE and KF. One practice trial was performed for each exertion prior to data collection. The isometric MVCs were performed at four different angles distributed evenly throughout the knee range of motion. Each measurement was repeated three times to make sure about the consistency of the data. Moreover, isokinetic (concentric and eccentric) MVCs were performed using a procedure described in a previous study (Anderson et al. 2007).

In the second session, strength measurements were performed in ankle plantar flexion (PF) and dorsiflexion (DF), followed by hip extension (HE), and hip flexion (HF). The testing protocol was similar to knee strength measurements with a few differences in participants' position and isokinetic velocities. Ankle measurements were taken while in a seated position with the knee flexed 50 deg and the hip flexed 80 deg. Hip measurements were taken while in a standing position, using a costume-built set up (Anderson et al. 2007), with the knee held in a near fully extended position by a knee immobilizer. A ten minute break was given between ankle and hip measurements.

During gait trials, ground reaction force was sampled at 1000 Hz using a force platform (Bertec Corporation, Columbus, OH) embedded in the middle of the walkway, and low-pass filtered at 40 Hz (4th order Butterworth filter). Body position was sampled at 100 Hz using a six-camera motion analysis system (MX-T10, Vicon Motion Systems Inc., L.A, CA) and low-pass filtered at 7 Hz (4th order Butterworth filter). Markers were placed bilaterally at the acromion process, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, lateral femoral epicondyle, lateral malleolus, calcaneus, and head of the 5<sup>th</sup> metatarsal. In addition, clusters of three markers placed on mid shank and thigh. During strength measurements, Biodex attachment angle, angular velocity, and moment were sampled at 200 Hz and low-pass filtered at 5 Hz (4th order Butterworth filter). The passive elastic/gravitational moment was estimated by fitting a line (least square) to torque data from relaxed trials throughout the range of motion, and was subtracted from each MVC trial (Anderson et al. 2010) to find the active component of strength. Both active and passive components of the strength were used to obtain a subject specific model of maximum available NMM (Anderson et al. 2007). The parameters of the model were found

by least square fitting the model to each subject's isometric, isokinetic, and passive strength measurements. The model outputs the maximum isometric NMM at each joint/exertion with the moderating effects of joint angle and angular velocity.

A four segment model of each participant was developed which included the upper body, thigh, shank, and foot. Segmental inertial parameters including segments mass, the location of center of mass, and segments moment of inertia were estimated using the method of Pavol et al. (2002). Hip and knee joint centers were identified by a functional method (Piazza et al.). The ankle joint center was identified as the midpoint between the lateral and medial malleoli markers. For each participant, a trial which best matched the requirement for each self-selected and controlled gait was used as the representative of the gait for that participant. The selection of the best trial for self-selected gait was based on the deviation of the subject's speed from her own average self-selected speed. The model was then used to perform a sagittal plane inverse dynamics analysis, using custom-written Matlab 2013a (The Mathworks Inc., Natick, MA) code, to estimate the NMM at the ankle, knee, and hip during gait.

Relative effort (RE) was determined using the method adapted from Anderson and Madigan (2014). At each instant in time during the gait trial, relative effort was calculated using:

$$RE = \frac{NMM}{NMM_{iso\_max}} \times 100$$

where  $NMM$  was obtained by inverse dynamics analysis, and  $NMM_{iso\_max}$  is the maximum available isometric NMM predicted by the model (Anderson et al. 2007; Anderson et al. 2010).

A two-way analysis of covariance was used to determine the effects of obesity group (healthy-weight or obese), age group (young or older), and their interaction on maximum available NMM, peak NMM, and peak relative effort during the stance phase of self-selected and controlled gait. Godin score was used as a potential covariate. The first hypothesis was tested using the main effect of obesity. The second hypothesis was tested using the obesity x age interaction. In the event of a significant obesity x age interaction, simple effects testing was used to assess the effects of obesity within each age group, and the effects of age group within each obesity group. Statistical analyses was performed using JMP Pro 10 (SAS Institute, Inc., Cary, NC) with a significance level of  $p < 0.05$ .

#### **4.4 Results**

Several obesity and age-related differences were observed in maximum available isometric NMM. Maximum available NMM differed between obesity groups in that it was 16.0 Nm higher in HF ( $p=0.006$ ), 23.8 Nm higher in KE ( $p=0.038$ ), and 5.1 Nm higher in ADF ( $p=0.006$ ) among obese subjects, and did not differ between obesity groups in HE ( $p=0.078$ ), KF ( $p=0.151$ ), and APF ( $p=0.420$ ; Table 5.2). Maximum available NMM differed between age groups in that it was 27.4 Nm lower in HF ( $p < 0.001$ ), 24.3 Nm lower in KE ( $p=0.039$ ), 18 Nm lower in KF ( $p < 0.001$ ), and 5 Nm lower in ADF ( $p=0.007$ ) among older subjects, and did not differ between age groups in HE ( $p=0.319$ ), and APF ( $p=0.299$ ).

Table 4.2 Maximum available isometric NMM for healthy-weight and obese groups (Least squares means (SE)), ° indicates main effect of obesity, and <sup>a</sup> demonstrate the main effect of age,  $p < 0.05$ . HW=healthy-weight, OB=obese, Y=young, and O=older.

<i>Muscle group</i>	<b>Maximum available isometric NMM (Nm)</b>			
	HW	OB	Y	O
<b><i>HE</i></b>	119.8(6.8)	137.8 (7.2)	133.9 (7.0)	123.7 (7.0)
<b><i>HF</i></b>	89.2 (3.7)	105.2 (3.9) °	110.9 (3.8)	83.5 (3.8) <sup>a</sup>
<b><i>KE</i></b>	121.7 (7.7)	145.6 (7.9) °	145.8 (7.8)	121.5 (8.0) <sup>a</sup>
<b><i>KF</i></b>	60.2 (3.1)	66.8 (3.2)	72.4 (3.2)	54.6 (3.3) <sup>a</sup>
<b><i>APF</i></b>	82.5 (5.2)	88.6 (5.4)	89.7 (5.3)	81.5 (5.5)
<b><i>ADF</i></b>	26.9 (1.2)	31.9 (1.2) °	31.9 (1.2)	26.8 (1.3) <sup>a</sup>

NMM and relative effort during the stance phase of self-selected and controlled gait exhibited similar trends across obesity and age groups with two local maxima/minima at the hip (HE1 and HF1), four at the knee (KF1, KE1, KF2, KE2), and two at the ankle (ADF1, APF1; Figure 4.2 and Figure 4.3).

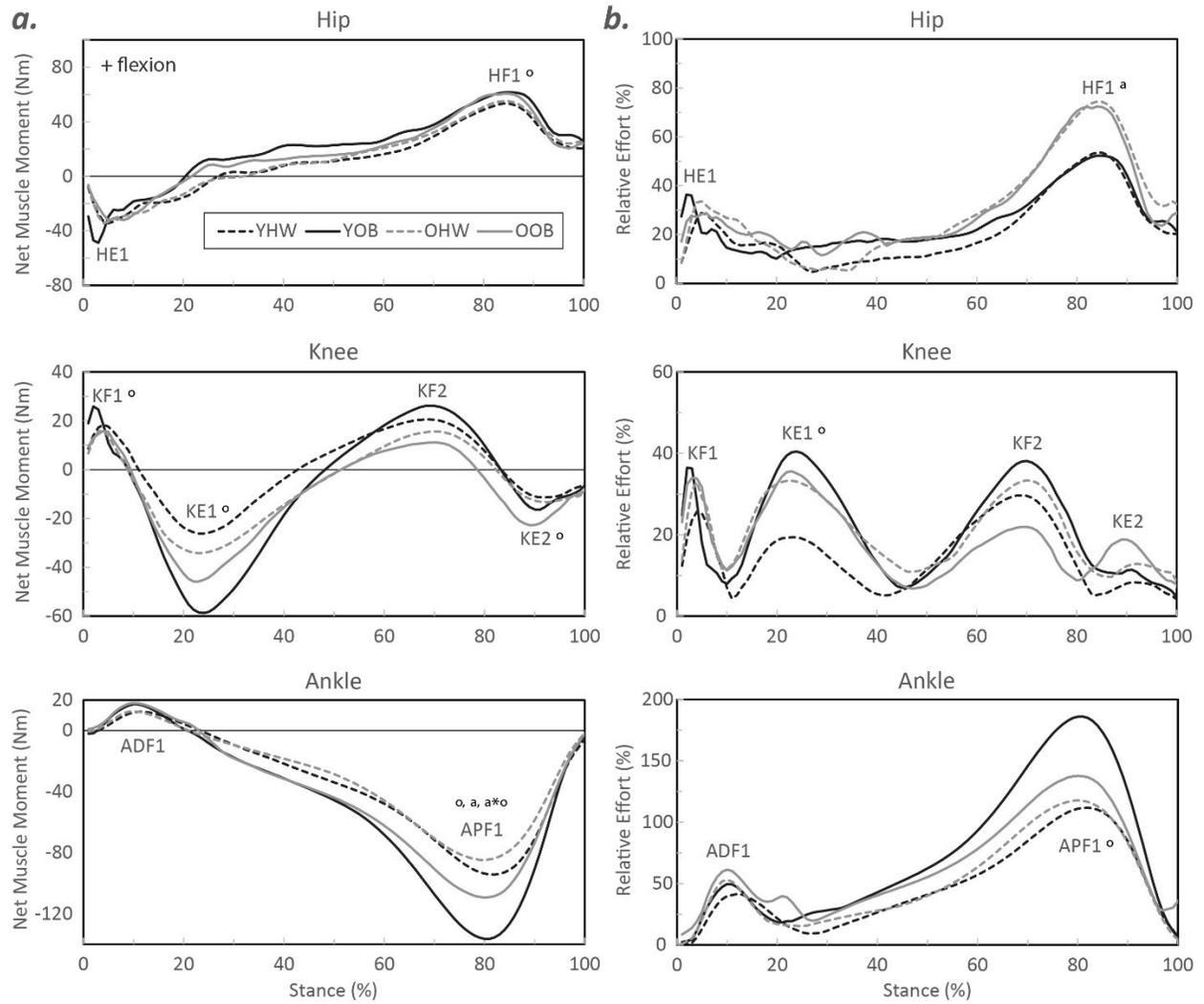


Figure 4.2 During self-selected gait, group means of a) NMMs b) relative effort over stance.

Positive NMMs are flexor dominant, and negative NMMs are extensor dominant. ° indicates

main effect of obesity, and <sup>a</sup> indicates main effect of age,  $p < 0.05$ .

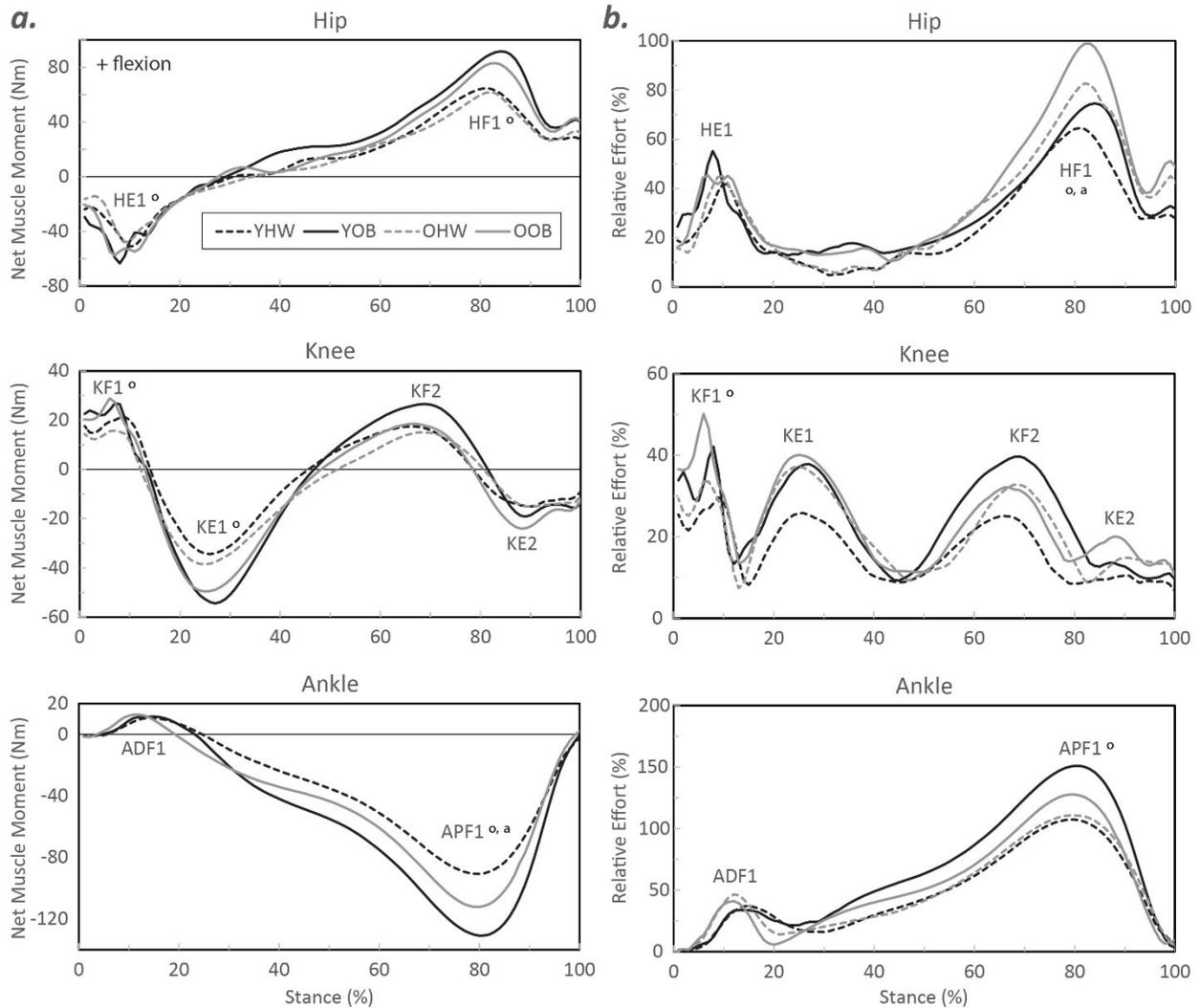


Figure 4.3 During controlled gait, group means of a) NMM b) relative effort over stance.

Positive NMMs are flexor dominant, and negative NMMs are extensor dominant. ° indicates

main effect of obesity, and <sup>a</sup> indicates main effect of age,  $p < 0.05$ .

During self-selected gait, peak NMM exhibited several obesity and age-related differences. Gait speed was 0.11 m/sec slower ( $p=0.005$ ) among obese subjects (1.25 m/sec vs. 1.36 m/sec), while step length (mean = 0.7 m) did not differ between obesity groups ( $p=0.583$ ). Gait speed and step length did not differ between age groups. The only obesity x age interaction for peak NMM was observed for APF1. Among younger subjects, APF1 was 45.6 Nm ( $p < 0.001$ ) higher among

obese subjects, and among older subjects was 25.6 Nm ( $p<0.001$ ) higher among obese subjects. Peak NMM differed between obesity groups in that it was 13 Nm higher in HF1 ( $p=0.004$ ), 5.2 Nm higher in KF1 ( $p=0.036$ ), 21.4 Nm higher in KE1 ( $p<0.001$ ), and 13.5 Nm higher in KE2 ( $p=0.006$ ) among obese subjects, and did not differ between obesity groups at HE1 ( $p=0.061$ ), KF2 ( $p=0.203$ ), or ADF1 ( $p=0.116$ ; Table 4.3). Peak NMM differed between age groups in that it was 19.1 Nm lower in APF1 ( $p<0.001$ ) among older subjects, and did not differ between age groups in HE1 ( $p=0.767$ ), HF1 ( $p=0.439$ ), KF1 ( $p=0.300$ ), KE1 ( $p=0.840$ ), KF2 ( $p=0.163$ ), KE2 ( $p=0.164$ ), and ADF1 ( $p=0.577$ ).

Table 4.3 Peak NMM for healthy-weight and obese groups (Least squares means (SE)),

° indicates main effect of obesity,  $p<0.05$ .

	Self-selected gait peak NMM (Nm)		Controlled gait peak NMM (Nm)	
	HW	OB	HW	OB
<i>HE1</i>	48.5 (3.5)	58.2 (3.6)	68.6 (4.6)	88.8 (4.7) °
<i>HF1</i>	55.3 (2.9)	68.3 (3.0) °	64.9 (3.7)	89.9 (3.8) °
<i>KF1</i>	23.8 (1.7)	29.1 (1.7) °	26.7 (1.9)	38.9 (2.0) °
<i>KE1</i>	33.4 (3.7)	54.7 (3.9) °	37.7 (4.1)	51.7 (4.1) °
<i>KF2</i>	19.1 (2.9)	24.5 (3.0)	16.3 (2.7)	23.9 (2.8)
<i>KE2</i>	13.5 (1.8)	20.9 (1.8) °	13.9 (4.0)	20.7 (4.1)
<i>ADF1</i>	13.4 (2.8)	19.8 (2.8)	12.3 (1.4)	14.9 (1.5)
<i>APF1</i>	89.7 (3.1)	125.4 (3.1) °	86.6 (3.2)	122.1 (3.3) °

Peak relative effort during self-selected gait exhibited several obesity and age-related differences. Peak relative effort differed between obesity groups in that it was 41% higher in KE1 ( $p=0.009$ ), and 30% higher at APF1 ( $p=0.008$ ) among obese subjects, and did not differ between obesity groups in HE1 ( $p=0.461$ ), HF1, ( $p=0.579$ ), KF1 ( $p=0.533$ ), KF2 ( $p=0.918$ ), KE2 ( $p=0.097$ ), or

ADF1 ( $p=0.404$ ; Figure 4.4). Peak relative effort differed between age groups in that it was 22% higher in HF1 ( $p<0.001$ ) among older subjects, and did not differ between age groups in HE1 ( $p=0.429$ ), KF1 ( $p=0.052$ ), KE1 ( $p=0.418$ ), KF2 ( $p=0.353$ ), KE2 ( $p=0.147$ ), ADF1 ( $p=0.237$ ), and APF1 ( $p=0.395$ ).

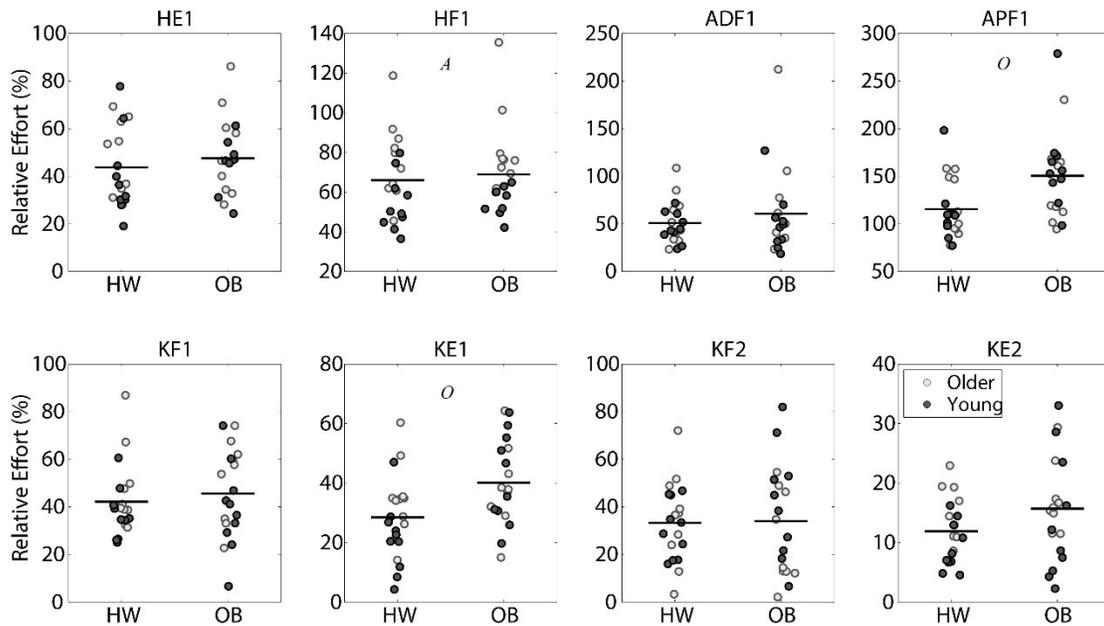


Figure 4.4 During self-selected gait; least square means of peak relative effort along with individual data points. Note: HW=Healthy-Weight; OB=Obese; *A* = main effect of age; *O* = main effect of obesity;  $p<0.05$ .

During controlled gait, peak NMM exhibited several obesity and age-related differences. Gait speed (mean = 1.49 m/sec) and step length (mean = 0.65) did not differ between obesity and age groups. Peak NMM differed between obesity groups in that it was 20.3 Nm higher in HE1 ( $p=0.004$ ), 25.1 Nm higher in HF1 ( $p<0.001$ ), 12.6 Nm higher in KF1 ( $p<0.001$ ), 14.0 Nm higher in KE1 ( $p=0.021$ ), and 35.5 Nm higher in APF1 among obese subjects, and did not differ between obesity groups in KF2 ( $p=0.057$ ), KE2 ( $p=0.249$ ), and ADF1 ( $p=0.207$ ; Table 4.3). Peak NMM differed between age groups in that it was 15.4 Nm lower in APF1 ( $p=0.002$ ) among older

subjects, and did not differ between age groups in HE1 ( $p=0.116$ ), HF1 ( $p=0.305$ ), KF1 ( $p=0.090$ ), KE1 ( $p=0.912$ ), KF2 ( $p=0.135$ ), KE2 ( $p=0.482$ ), and ADF1 ( $p=0.879$ ).

Peak relative effort during controlled gait exhibited several obesity and age-related differences. Peak relative effort differed between obesity groups in that it was 16% higher in HF1 ( $p=0.043$ ), 36% higher in KF1 ( $p=0.003$ ), and 33% higher in APF1 ( $p=0.003$ ) among obese subjects, and did not differ between obesity groups in HE1 ( $p=0.261$ ), KE1 ( $p=0.144$ ), KF2 ( $p=0.293$ ), KE2 ( $p=0.453$ ), and ADF1 ( $p=0.893$ ; Figure 4.5). Peak relative effort differed between age groups in that it was 19% higher in HF1 ( $p=0.003$ ) among older subjects, and did not differ between age groups in HE1 ( $p=0.628$ ), KF1 ( $p=0.478$ ), KE1 ( $p=0.415$ ), KF2 ( $p=0.483$ ), KE2 ( $p=0.389$ ), ADF1 ( $p=0.902$ ), and APF1 ( $p=0.670$ ).

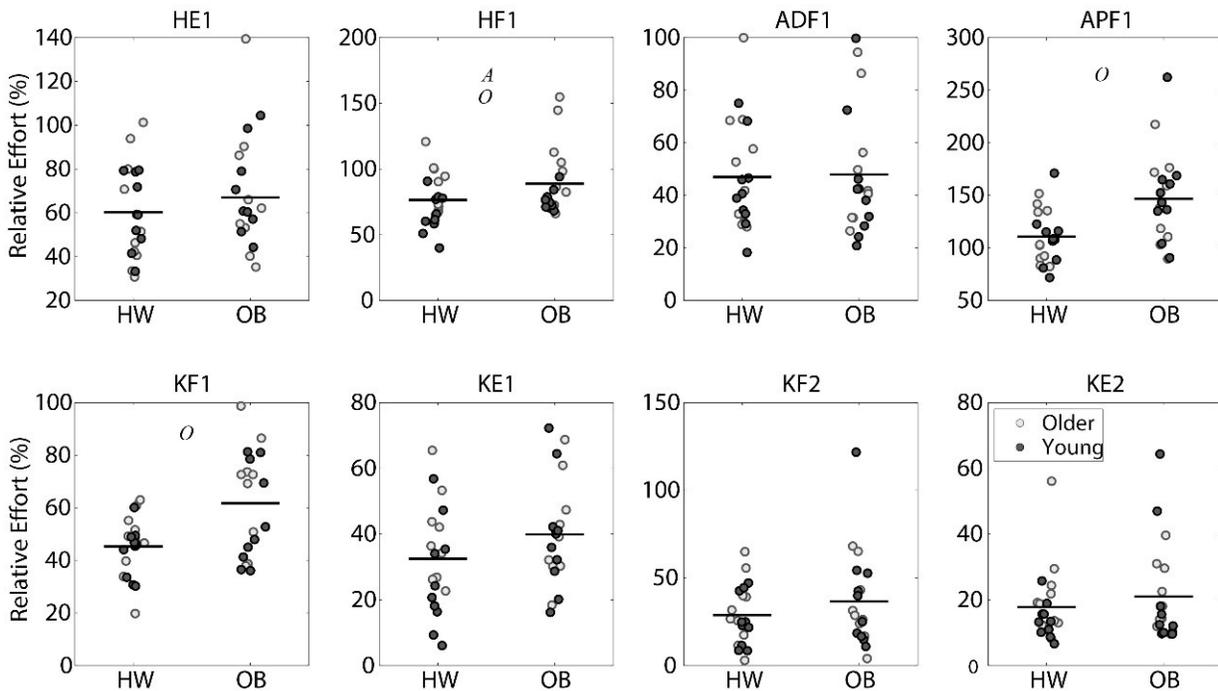


Figure 4.5 During controlled gait; least square means of peak relative effort along with individual data points. Note: HW=Healthy-Weight; OB=Obese; *A* = main effect of age; *O* = main effect of obesity;  $p<0.05$ .

An overview of the findings is shown in Table 4.4 to summarize the direction of obesity and age-related differences in maximum available NMM, peak NMM during gait, and peak relative during gait.

Table 4.4 Overview direction of obesity and age-related differences in maximum available NMM, peak NMM during gait, and peak relative during gait, *ss*=self-selected, *con*=controlled,  $\uparrow$  = higher ( $p < 0.05$ ) among obese or older group (as labeled),  $\downarrow$  = lower ( $p < 0.05$ ) among obese or older group (as labeled).

	<b>Maximum available NMM</b>		<b>Peak NMM during gait</b>				<b>Peak relative effort</b>			
	Obese	Older	Obese		Older		Obese		Older	
			<i>ss</i>	<i>con</i>	<i>ss</i>	<i>con</i>	<i>ss</i>	<i>con</i>	<i>ss</i>	<i>con</i>
<b><i>HE1</i></b>	-	-	-	29% $\uparrow$	-	-	-	-	-	-
<b><i>HF1</i></b>	18% $\uparrow$	25% $\downarrow$	24% $\uparrow$	39% $\uparrow$	-	-	-	16% $\uparrow$	22% $\uparrow$	19% $\uparrow$
<b><i>KF1</i></b>	-	25% $\downarrow$	22% $\uparrow$	46% $\uparrow$	-	-	-	36% $\uparrow$	-	-
<b><i>KE1</i></b>	20% $\uparrow$	17% $\downarrow$	64% $\uparrow$	37% $\uparrow$	-	-	41% $\uparrow$	-	-	-
<b><i>KF2</i></b>	-	25% $\downarrow$	-	-	-	-	-	-	-	-
<b><i>KE2</i></b>	20% $\uparrow$	17% $\downarrow$	55% $\uparrow$	-	-	-	-	-	-	-
<b><i>ADF1</i></b>	19% $\uparrow$	16% $\downarrow$	-	-	-	-	-	-	-	-
<b><i>APF1</i></b>	-	-	40% $\uparrow$	41% $\uparrow$	16% $\downarrow$	14% $\downarrow$	30% $\uparrow$	33% $\uparrow$	-	-

#### 4.5 Discussion

The purpose of this study was to investigate the effects of obesity, age and their interaction on relative effort at the ankle, knee, and hip during gait. Our first hypothesis was that peak relative effort would be higher at the ankle, knee and hip among obese females. This hypothesis was

supported. Peak relative effort was 30-33% higher in APF1 among obese females during self-selected and controlled gait. This difference was attributed to the 40-41% higher APF1 NMM among obese females since maximum available NMM in APF did not differ between obese and healthy-weight females. Peak relative effort at APF1 exceeded 100% for both healthy-weight and obese groups, which implies that both groups walked at or near their maximum effort in APF. Ankle plantar flexors provide support and propulsion during gait (Kepple et al. 1997; Sadeghi et al. 2001), and produce approximately 73% of the overall sagittal plane work generated by the lower extremities during gait (DeVita and Hortobágyi 2003). Therefore, the higher relative effort among obese females in APF1, and that they had little/no capacity to further increase NMM, could be one of the main contributing factors to their compromised walking ability. Peak relative effort at KE1 was 41% higher among obese females during self-selected gait, and was attributed to a 64% higher NMM and 20% higher maximum available NMM at KE among obese females. The knee extensors help in resisting knee flexion during body support (Winter 1984). Therefore, it is possible that obese females required higher relative effort at KE1 for body support due to their increased body mass.

Obese females exhibited 16% higher relative effort at HF1 during controlled gait. This higher relative effort at HF1 was attributed to a 39% higher NMM and 18% higher maximum available NMM in HF. In addition to the ankle plantar flexors, the hip flexors during late stance and hip extensors during early stance are involved in energy generation that maintain or increase the forward velocity of the body (Chen et al. 1997; Requião et al. 2005; Winter 1991). Therefore, it is possible that obese females had to generate higher relative effort at HF1 during late stance in order to swing the leg forward more quickly since they were maximally using their ankle plantar

flexors. The high relative effort in HF1 (89%) implies that obese females used their hip flexors near their maximum capacity during late stance during controlled gait. Obese females also exhibited a 36% higher relative effort at KF1 during controlled gait. This higher relative effort in KF1 was attributed to a 46% higher NMM since maximum available NMM at KF did not differ between obesity groups. The NMM in KF1, which appeared near heel strike, showed a momentary flexor pattern that may be associated with HE1 via hamstring contraction. As such the higher values of relative effort in KF1 may indicate higher level of hamstring contraction.

Obesity-related differences in peak relative effort varied between gait conditions in that relative effort among obese females was higher in KE1, APF1 during self-selected gait, and higher in HF1, KF1, and APF1 during controlled gait. Since step length did not exhibit any obesity-related differences during self-selected gait (mean = 0.7 m), it is reasonable to assume the differences between self-selected and controlled gait are likely due to differences in gait speed (mean of 1.3 m/sec in self-selected and 1.5 m/sec in controlled gait). Requião et al. (2005) suggested that, by increasing the cadence in healthy subjects, the relative effort increases at both the ankle and hip during the period of energy generation, but with more pronounced increases for the hip flexor and hip extensor compared to the ankle plantar flexor. It was hypothesized that this increased role of the hip musculature with cadence is a strategy to attenuate fatigue and share the effort across more and larger muscle groups (Requião et al. 2005). Therefore, the higher relative effort in HF1 among obese females during controlled gait is possibly due to the fact that they had to increase their gait speed by 20% from their self-selected gait speed (1.25 m/sec vs. 1.5 m/sec) compared to 10% increase required by healthy-weight females (1.36 m/sec vs. 1.5 m/sec). Although we did not see an obesity-related difference in relative effort in HE1 during controlled

gait, the higher relative effort in KF1, which may be associated with HE1 via hamstring contraction, implied higher demand on hamstring among obese females during a faster gait speed. The lack of obesity-related increase in KE1 during controlled gait needs further investigation, but one possible explanation is that we did not have sufficient statistical power to detect such differences if they were small.

Peak relative effort was higher among obese females in KE1, KF1, and HF1. Although these higher values of relative effort did not exceed 90% and therefore suggested some available NMM reserve, walking at higher relative effort may still compromise walking ability. First, walking at higher relative effort can result in premature fatigue (Hortobágyi et al. 2003; Lerner et al. 2014), which can explain the lower gait speed and self-reported difficulty in walking longer distances (500 m) among obese individuals (Lai et al. 2008; Spyropoulos et al. 1991; Stenholm et al. 2007). Second, the increased level of muscle force exertion can result in an increase of force variability (Carlton and Newell 1993; Christou et al. 2002; Sherwood and Schmidt 1980) and such variability was considered a noise in neuromuscular system (Sherwood et al. 1988) which consequently can lead to movement error (Meyer et al. 1982; Schmidt et al. 1978). Finally, since walking with higher relative effort leaves a lower reserve capacity to utilize in case of unexpected circumstances (Samuel et al. 2013), it is possible that obese individuals alter their spatiotemporal gait characteristics, e.g. increase period of double support phase, to help in maintaining their dynamic balance (Hills et al. 2002; Spyropoulos et al. 1991) in case of such accidents.

Peak relative effort among healthy-weight females was 44% in HE1, 66% in HF1, and 115% in APF1 during self-selected gait. These values were slightly to moderately higher than previously reported ranges of 27%-38% for HE, 34%-45% for HF, and 57%-92% for APF (Anderson and Madigan 2014; Requião et al. 2005). There are at least two factors that may contribute to these differences in magnitude. The first factor is difference in the gender makeup of subjects between studies. In current study, we only included female participants while other studies included both genders (Anderson and Madigan 2014; Requião et al. 2005). It has been reported that females expend greater propulsive energy during push off (Kerrigan et al. 1998). This, accompanied by reduced lower extremity strength among females compared to males (Frontera et al. 1991; Harbo et al. 2012), may explain higher relative effort among females. The second factor is a difference in method of assessing maximum available NMM including different approaches to account for joint angle and angular velocity during gait. In present study we utilized the isometric available NMM to determine the relative effort but the joint angle and angular velocity can impact the maximum available NMM. For example, maximum available eccentric NMM can be higher than maximum available isometric NMM which may explain the higher relative effort if the maximum available isometric NMM are used instead of maximum available eccentric NMM. The noted previous studies accounted for the effect of angle and angular velocity to obtain the maximum available NMM, which may explain the discrepancies in magnitude of relative effort between studies.

Our second hypothesis was that the effect of obesity on relative effort would be larger among older females compared to young females. This hypothesis was not supported because peak relative effort did not exhibit any obesity by age interaction effects. One possible explanation for

this lack of interaction could be that we did not have sufficient statistical power to detect interactive effects, if they are small, however, this needs to be further investigated. In general, these results suggest that obesity does not differentially affect relative effort among young and older females.

Regarding age-related differences in relative effort, only HF1 differed between age groups and was 18-22% higher among older females, and had mean values of 78% and 92% during self-selected and controlled gait, respectively. This higher relative effort among older females was attributed to a 25% lower maximum available NMM among older females because HF1 NMM did not differ between age groups. It has been suggested that older adults compensate for reduced plantar flexor torque by increased hip extensor work and power (DeVita and Hortobagyi 2000; Monaco et al. 2009) or alternatively with increased hip flexor power and work in late stance (Cofré et al. 2011; JudgeRoy et al. 1996). The findings in present study seems to be consistent with the latter hypothesis in that relative effort was higher among older females in hip flexion not the hip extension. In other words, when ankle plantar flexor torque is reduced, older adults may rely more on the hip flexors to propel the leg into swing (Cofré et al. 2011).

There are several limitations to this study that warrant mention. The first limitation is that peak relative effort exceeded 100% in some cases (particularly at APF1) which suggests higher NMM than possible. Three possible explanations can contribute to these higher than expected values. First, it is possible that subjects did not generate their maximum NMM during strength measurements. Second, it is also possible that the maximum available NMM obtained by dynamometer were influenced by bi-articular muscles since the gastrocnemius muscle is a bi-

articular and sensitive to both ankle and knee angles and angular velocities. Third, the model used to determine maximum available NMM is a fit to NMM data during strength measurements, and may underestimate the available strength at some joint angles (Anderson et al. 2007). The second limitation is that we only utilized the estimated maximum available isometric NMM to calculate the relative effort and did not directly account for the effect of angular velocity on strength capacity. Accounting for angular velocity can impact the magnitude of relative effort since maximum available NMM can be higher in eccentric and lower in concentric exertions compared to isometric exertions. We did not account for angular velocity in present study because the model to estimate the maximum available NMM (Anderson 2007) appeared to be overly sensitive to small changes in angular velocity, and increased the variability in findings. The third limitation is that the skin movement artifact tissue is exacerbated among obese individuals due to adipose which can cause error in kinematics data. Finally, we only included female subjects, and the findings may not be generalizable to males.

#### **4.6 Conclusions**

Obese females exhibited higher relative effort at the hip, knee, and ankle during gait. The higher relative effort, especially in ankle plantar flexion where little/no reserve in strength was available, may be a major contributing factor to compromised walking ability among obese individuals. The findings may help in developing targeted interventions to maintain and improve mobility among obese individuals.

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## Chapter 5 - Effects of Age-related Loss of Strength on Non-stepping Balance

### Recovery

#### 5.1 Abstract

Aging has been associated with higher risk of falls. One of the factors contributing to this higher risk is the impaired ability to recover balance after a postural perturbation. Reduced lower limb strength likely limits the ability to recover balance from a postural perturbation to avert a fall. The effect of aging on balance recovery using ankle strategy has been extensively studied but such effects have not been investigated when the movement is not constrained to ankle. The purpose of this study was to investigate the effects of age-related strength loss on non-stepping balance recovery capability after a perturbation while standing, without constraining the movements to ankle strategy. Two experiments were conducted. In the first experiment five young adults (ages 20-30) and six community-dwelling older adults (ages 70-80) recovered their balance without stepping from a backward displacement of a support surface. The balance recovery capability was quantified by the maximum recoverable platform displacement (MRPD) that subjects could withstand without stepping. The MRPD was 27% smaller ( $p=0.014$ ) among the older group ( $11.8\pm 2.1$  cm) compared to the young group ( $16.2\pm 2.6$  cm). In the second experiment, forward dynamic simulations of a two segment rigid-body model were used to investigate the effects of manipulating muscle strength at hip extensor ( $HE_{35\%}, HE_{20\%}$ ), hip flexor ( $HF_{25\%}, HF_{20\%}$ ), ankle plantar flexor ( $PF_{35\%}, PF_{20\%}$ ), and ankle dorsiflexor ( $DF_{16\%}, DF_{20\%}$ ) on MRPD where the subscript represents loss of muscle strength compared to the young model, considering typical reductions in muscle strength associated with aging and when considering equal reductions. The MRPD was reduced 37% for  $PF_{35\%}$ , 5% for  $HF_{25\%}$ , 21% for  $PF_{20\%}$

compared to the young model and did not differ in rest of the cases. The finding suggested that plantar flexor strength plays a major role in capability to recover balance even though the movement was not constrained to the ankle. However, the body may choose different magnitude of the hip and ankle motion to recover balance depending on which joint/exertion endured loss of muscle strength. The findings can help guide the development of targeted strength training interventions to improve non-stepping balance recovery capability among older adults.

## **5.2 Introduction**

Aging is also associated with a higher prevalence of falls and falls-related injuries. About 40% of adults age 65 and older fall each year (Rubenstein 2006). In 2001, an estimated 1.64 million older adults were treated in emergency departments for unintentional fall-related injuries (Stevens and Sogolow 2005). Knowing that the number of adults age 65 and older is expected to double from 40 million in 2010 to over 80 million in 2050 (Vincent and Velkoff 2010), the high prevalence of falls among this age group is expected to grow substantially.

Older adults commonly exhibit an impaired balance recovery capability after a postural perturbation while standing (Mackey and Robinovitch 2006; Madigan and Lloyd 2005; Mansfield and Maki 2009; Thelen et al. 2000; Wojcik et al. 1999). For example, Mackey and Robinovitch (2006) determined the largest static lean angle from which young and older women could recover balance upon release when using the so-called ankle strategy. Older women exhibited a 36% smaller maximum lean angle (13.1 deg) compared to young women (16.3 deg). Wojcik et al. (1999) determined the largest lean angle from which young and older females could recover by taking a single step. They found that older females exhibited 47% smaller maximum

lean angle (16.2 deg) compared to young females (30.7 deg). Similarly, Madigan and Lloyd (2005) reported that older males exhibited 31% smaller maximum lean angle (20.5 deg) compared to young males (29.9 deg) while recovering balance by taking a single step.

Reduced lower limb strength is one of the major contributors to the higher risk of falls among older adults (Graafmans et al. 1996; Moreland et al. 2004), which likely limits balance recovery capability after a postural perturbation while standing (Barrett and Lichtwark 2008; Madigan and Lloyd 2005). Madigan and Lloyd (2005), for example, reported increasing net muscle moments in plantar flexion, knee extension, and hip extension as adults recovered balance with a single step after being released from progressively larger static lean angles. This indicates greater strength is needed as perturbation severity increases, and older adults may not be able to recover from more severe perturbations due to their reduced strength. Barrett and Lichtwark (2008) performed a simulation study to investigate the effect of age-related changes in neural, muscular, and tendinous parameters on balance recovery using ankle strategy. They reported a 19% decrease in the maximum recoverable lean angle from 7.2 deg to 5.8 deg after reducing isometric ankle plantar flexor force by 20% from 7200 N to 5760 N. Similarly, balance recovery capability when using an ankle strategy can be improved by increasing muscle strength (Matrangola and Madigan 2009; Robinovitch et al. 2002). Robinovitch et al. (2002) performed a simulation study to determine that increasing the maximum ankle net muscle moment by 100% resulted in 64% increase in the maximum recoverable lean angle from 5.3 deg to 8.7 deg. Matrangola and Madigan (2009) also performed a simulation study, and determined that increasing ankle strength 15% resulted in a one degree increase in maximum recoverable lean angle that could be achieved using the ankle strategy (albeit when modeling an obese individual) . While both of

these studies provide useful insight on the importance of ankle plantar flexor strength when using an ankle strategy, larger perturbations typically elicit hip movement, in addition to ankle movement, when recovering balance from a backward movement of a support surface without stepping (Runge et al. 1999). The addition of hip movement often occurs when pure ankle strategy is not sufficient in restoring balance, and the two joints often exhibit counter-phase motions (Horak and Nashner 1986; Runge et al. 1999). As such, a logical extension of the previous works would be to investigate non-stepping balance recovery without constraining the movement to ankle strategy.

The purpose of this study was to investigate the effects of age-related strength loss on non-stepping balance recovery capability after a perturbation while standing, without constraining the movements to ankle strategy. Two experiments were conducted. In the first experiment, human subjects testing was used to investigate age-related differences in non-stepping balance recovery capability. We hypothesized that older adults would exhibit reduced balance recovery capability compared to young adults. Given this cross-sectional design is susceptible to confounding factors that can also contribute to age-related differences, and that manipulating muscle strength (and no other factors) in human subjects is difficult and time-consuming, a second experiment was conducted to isolate the effects of age-related strength loss on balance recovery capability. In this second experiment, a simulation study was performed to investigate the effects of manipulating muscle strength on balance recovery capability. The results from this study will help provide insight into how age-related strength loss contributes to the impaired balance recovery capability among older adults.

### 5.3 Methods

#### *Human Subjects Testing*

Two groups of adults completed the study including five young adults (ages 20-30) and six community-dwelling older adults (ages 70-80) (Table 5.1). All subjects were able to walk unassisted. The study was approved by the local Institutional Review Board, and written consent was obtained from all subjects prior to participation.

Table 5.1 Subject characteristics (mean  $\pm$  SD)

	<b>Young</b>	<b>Older</b>
<b>Age (years)</b>	23.4 $\pm$ 2.4	73.2 $\pm$ 2.2
<b>Sex (M/F)</b>	3/2	3/3
<b>Height (cm)</b>	174 $\pm$ 7	165 $\pm$ 9
<b>Mass (kg)</b>	72.5 $\pm$ 8.1	68.3 $\pm$ 7.4
<b>BMI (kg/m<sup>2</sup>)</b>	23.9 $\pm$ 1.9	25.0 $\pm$ 1.9

Balance recovery capability was quantified by finding the maximum backward displacement of a pneumatic, instrumented, moving platform (PIMP) that subjects could withstand without stepping. The backward displacement of platform induced a forward loss of balance, and the forward displacement elicited a backward loss of balance. PIMP translated 0-0.25 m backward and 0-0.15 m forward in approximately 350 msec (Figure 5.1). PIMP displacement was modulated by increasing peak translation speed (e.g. peak translation speed was 1.1 m/sec for a displacement of 0.19 m; Figure 5.2).



Figure 5.1 Picture of PIMP set up

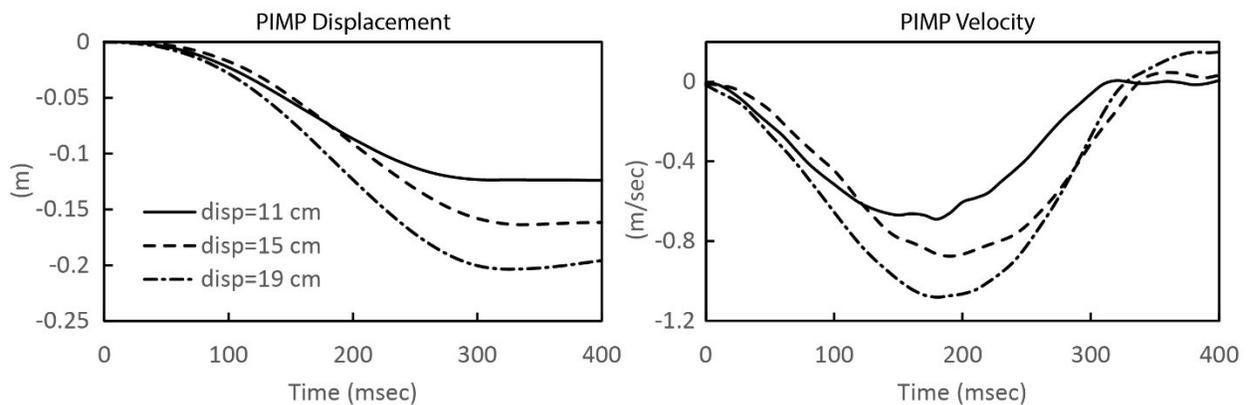


Figure 5.2 PIMP displacement and velocity for three different perturbations.

At the beginning of testing, subjects stood barefoot on the PIMP with their feet approximately shoulder width apart, eyes open, and while looking straight ahead. They were instructed to remain relaxed, try their best not to step, and remain standing still after the perturbation. An assistant stood nearby to provide assistance with recovering balance, if needed. The first trial began with the PIMP moving approximately 0.02 m backward. After a successful (i.e. non-

stepping and without assistance) trial, another trial was performed with displacement increased by 0.01 m. After a failed trial (i.e. the subject stepped or required assistance to maintain balance), another trial was repeated at the same displacement. If this repeated trial was successful, then the PIMP displacement was increased 0.01 m. This process was repeated until three failed trials occurred at the same platform displacement. Both forward and backward platform translations were presented in random order to prevent anticipation of translation direction. The maximum backward displacement that each subject could successfully maintain balance without stepping or requiring assistance was considered as the maximum recoverable platform displacement (MRPD).

During all trials, body position was sampled at 100 Hz using a six-camera motion analysis system (MX-T10, Vicon Motion Systems Inc., L.A, CA) and low-pass filtered at 40 Hz (4th order Butterworth filter). Passive reflective markers were placed bilaterally at the acromion process, anterior superior iliac spine, posterior superior iliac spine, greater trochanter, lateral femoral epicondyle, lateral malleolus, heel, and head of the 5<sup>th</sup> metatarsal. The marker data were used to measure subject kinematics for the subsequent simulation study.

MRPD was compared between young and older groups using a *t*-test. A significant level of  $p < 0.05$  was used, and the analysis was performed using JMP v10 (SAS Institute, Inc, Cary, NC)

### *Simulation Study*

A simulation study was performed to investigate the effect of manipulating muscle strength on balance recovery capability. This simulation study was performed in three steps described in

detail below. Briefly, step one determined the MRDP of a model representing a young subject. This step established a baseline measure of balance recovery capability of the model for comparison with subsequent simulations after manipulating muscle strength. In step two, muscle strength was manipulated to represent age-related strength loss, and the MRPD was again determined. This step determined the effect of age-related strength loss on MRPD, and allowed a comparison with age differences in MRPD among the human subjects. In step three, age-related loss was induced at each joint/exertion individually (hip extension, hip flexion, ankle plantar flexion, ankle dorsiflexion) to determine the relative effect of decreasing the strength of specific muscle groups on MRPD.

A two-segment, sagittal plane, torque-driven model of a young male adult (age=21 years, height=1.7 m, weight=66.5 kg) was used. One segment represented the lower limbs (shank and thigh), and the other segment representing a HAT (head, arms, and trunk) segment (Figure 5.3). Torque actuators at the ankle and hip represented muscle actuation. This simplified model did not include a knee joint due to relatively small changes in knee angle ( $\sim 5$  deg) during human subjects testing. It also did not include a foot segment due to relatively small heel rise ( $< 2.5$  cm) during human subjects testing. Therefore a pin joint, representing the ankle, connected the distal end of the lower limb to the floor. Segment mass-inertial parameters were estimated from de Leva (1996). The equations of motion were derived using Autolev software (OnLine Dynamics, Inc., Sunnyvale, CA), and integrated using a fixed-step-size (0.001 sec) 4<sup>th</sup>-order Runge-Kutta method. All the coding was performed in MATLAB (The MathWorks, Natick, MA).

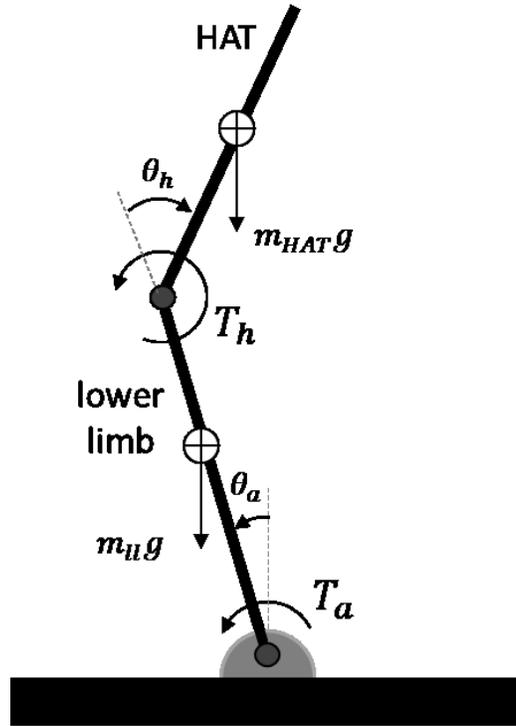


Figure 5.3 Schematic of the two-segment, rigid-body model,  $\theta_h = \text{hip angle}$ ,  $\theta_a = \text{ankle angle}$ ,  $T_h = \text{hip torque}$ ,  $T_a = \text{ankle torque}$ ,  $HAT = \text{heads, arms, and trunk segment}$ ,  $ll = \text{lower limb}$

Torque produced by actuators at the ankle and hip was determined by a model incorporating multiple aspects of muscle function. This torque was the sum of a passive elastic torque that was a function of angle and angular velocity (Riener and Edrich 1999), and an active torque that was a function of maximum isometric torque, joint angle, joint angular velocity, and activation level (Ashby and Delp 2006; Cheng et al. 2008; Selbie and Caldwell 1996):

$$T_{passive} = f(\theta, \dot{\theta})$$

$$T_{active} = T_{max} \times f(\theta) \times h(\omega) \times a(t)$$

$$T = T_{active} + T_{passive}$$

where  $T_{max}$  was the maximum isometric torque (Nm),  $f(\theta)$  and  $h(\omega)$  were dimensionless scaling factors for angle and angular velocity which ranged from 0 to 1, and  $a(t)$  was the joint activation which varied from -1 to 1 to represent maximum activation of the flexors and extensors, respectively. The  $T_{max}$  values for the young model were chosen from previously reported data with a representative young subject's height and weight (Anderson et al. 2007). The torque-angle relation,  $f(\theta)$ , was found using a polynomial fit of existing experimental data (Hoy et al. 1990). The torque-angular velocity relation,  $h(\omega)$ , was defined as:

$$\begin{cases} h(\omega) = \frac{(\omega_0 - \omega)}{(\omega_0 - \Gamma\omega)} & , \omega/\omega_0 < 1 \\ h(\omega) = 0 & , \omega/\omega_0 \geq 1 \end{cases}$$

where  $\omega$  is the instantaneous angular velocity,  $\omega_0$  is the maximum angular velocity, and  $\Gamma$  is the shape factor of the torque-angular velocity curve (Alexander 1989; Selbie and Caldwell 1996). The joint activation,  $a(t)$ , is used as an analog of muscle activation, but applies to the flexor or extensor muscle group at a joint rather than an individual muscle. The values for  $a(t)$ , was determined from "activation nodes" spaced at 50 msec increments for the duration of the simulation. The value of joint activation between consecutive nodes was determined using linear interpolation (Figure 5.4).

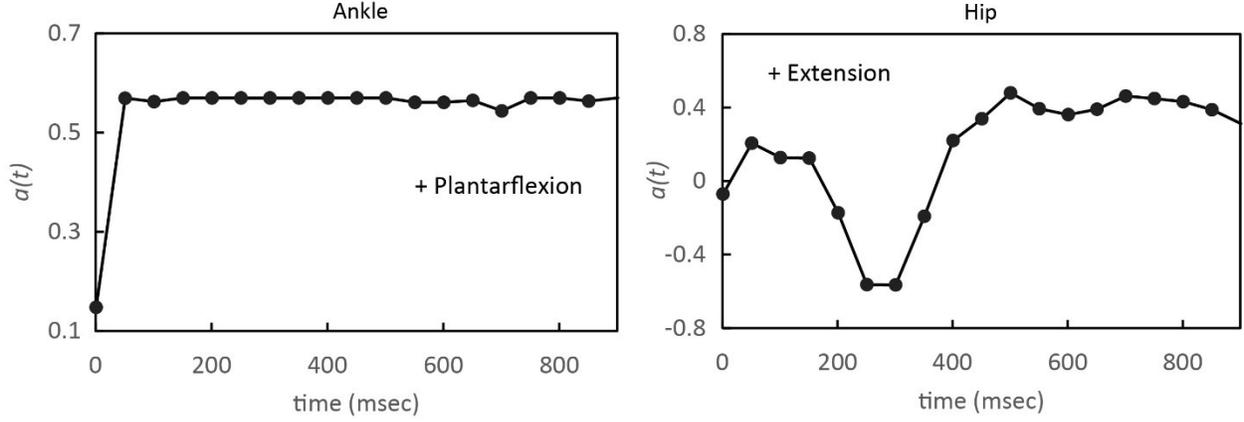


Figure 5.4 Sample joint activation at the ankle and hip joints for a representative balance recovery simulation (MRPD=19 cm). Time zero indicates the start of PIMP displacement.

The values of the joint activation nodes were determined using simulated annealing and a cost function based upon model kinematics. The cost function was developed based upon the work of Yang et al. (2008) who used forward dynamic simulations and dynamic optimization to identify states leading to a backward balance loss following a slip during gait. The cost function was defined as:

$$f = w_1 \int_{t_s}^{t_f} |X_{COM} - X_{ankle}| dt + w_2 \int_{t_s}^{t_f} |\dot{X}_{COM} - \dot{X}_{ankle}| dt + w_3 \int_{t_i}^{t_f} e(\theta(t)) dt$$

$$+ w_4 \int_{t_i}^{t_f} e(\dot{\theta}(t)) dt + w_5 \sqrt{\sum_{i=1}^2 \dot{q}_i^2}$$

The first and second terms in the cost function were used to minimize the horizontal COM displacement ( $X_{com}$ ) relative to the ankle ( $X_{ankle}$ ) and the horizontal COM velocity ( $\dot{X}_{COM}$ ) relative to the ankle ( $\dot{X}_{ankle}$ ), respectively. The value of  $t_s$  which represented the reaction time,

was set to 150 msec based upon a study by Runge et al. (1999) who investigated balance recovery after backward translation of a support surface.  $t_f$  was the time at the end of simulation.

The third and fourth terms restricted the joint angles and angular velocities to remain within the physiological limit (Luttgens and Wells 1982) throughout the simulation, where  $t_i$  was initial

time of the simulation, and  $e(\theta(t)) = \sum \varphi(s_i(t))$  where  $\varphi(s_i(t)) = \begin{cases} s_i^- - s_i & s_i < s_i^- \\ 0 & s_i^- < s_i < s_i^+ \\ s_i - s_i^+ & s_i > s_i^+ \end{cases}$ ,

$s_i^-$ , and  $s_i^+$  represented the upper and lower physical bound of joint angle and angular velocity

(Yang et al. 2008). The last term minimized the square root of segments angular velocities ( $\dot{q}$ )

and was used to bring the model to the rest at the end of the simulation. The scale of  $w$  was the

weight for each term and was determined by trial and error to make the model better simulate the

human subject behavior. The simulation was terminated when changes in consecutive objective

function values did not exceed 0.1 for 20 iterations.

The total simulation duration was 900 msec which was determined from human subjects' data to

be sufficient to determine the outcome (recovery or fall) of each trial. The platform motion

profile was adapted from the PIMP displacement during human subjects testing (Figure 5.5).

Starting with a platform displacement of 2 cm, the optimization was performed to determine if

the model could recover balance. The trial was deemed a recovery if the whole-body COM did

not move anteriorly past the limit of the base of support (head of the 5th metatarsal), and the

horizontal velocity of the whole-body COM became negative (posterior movement), and

remained negative, for at least 500 msec. If these two criteria were not satisfied, then the trial

was deemed a fall. These criteria were selected based upon an investigation of the human

subject's successful and failed recoveries (Figure 5.6). After a successful recovery, the

magnitude of platform displacement was increased by 1 cm, and the optimization was repeated. This process was continued until the MRPD was determined.

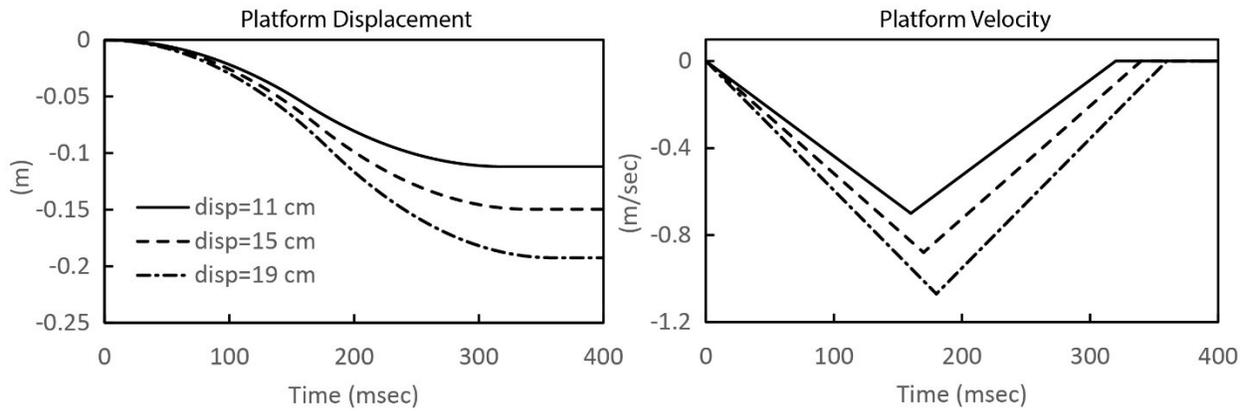


Figure 5.5 Adapted platform displacement and velocity for three different perturbation magnitudes.

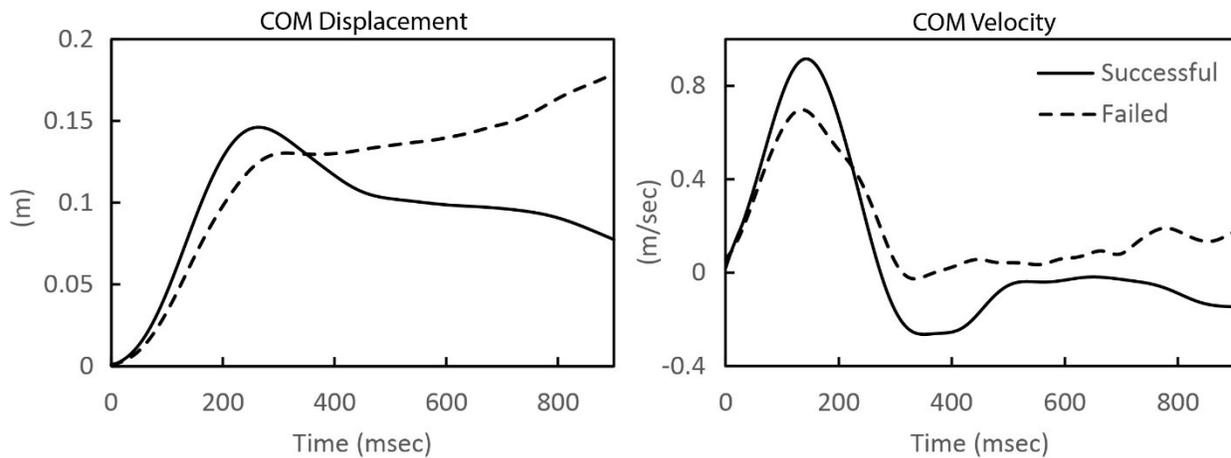


Figure 5.6 The whole-body COM displacement and COM velocity for a successful (solid line) and failed (dash line) balance recovery (young human subject data)

The first step of the simulation study was to determine the maximum limit of joint activation (absolute value) that, when applied to both extension and flexion at the hip and ankle, matched

the model MRPD with the MRPD of a representative young male human subject that was modeled. This aligned the model performance with the human subject performance. After identifying this maximum joint activation, it was used as the maximum possible joint activation (+1 or -1) in subsequent simulations.

In step two, the maximum isometric strength ( $T_{max}$ ) in the torque model was reduced to mimic age-related loss of muscle strength. This included reducing individual  $T_{max}$  values for hip extensor (HE), hip flexor (HF), ankle plantar flexor (PF), and ankle dorsiflexor (DF). The amount of reduction was based upon the data of Harbo et al. (2012), who reported age differences in lower limb strength between young adult males (15-30 years) and older adult males (70-80 years). Based on their data,  $T_{max}$  was reduced by 35% at HE, 25% at HF, 35% at PF, and 16% at DF to mimic age-related strength loss experienced by an older male (Table 5.2). MRPD was then found using the same procedure described above.

In step three, the effect of reducing  $T_{max}$  of HE, HF, DF, and PF individually on MRPD was investigated. To do this, we investigated eight different cases of age-related strength loss (Table 5.2). In the first four cases ( $HE_{35\%}$ ,  $HF_{25\%}$ ,  $PF_{35\%}$ ,  $DF_{16\%}$ )  $T_{max}$  at the HE, HF, PF, and DF was individually reduced based on the values reported by Harbo et al. (2012). In the second four cases ( $HE_{20\%}$ ,  $HF_{20\%}$ ,  $PF_{20\%}$ ,  $DF_{20\%}$ )  $T_{max}$  at the HE, HF, PF, and DF was individually reduced by 20%. In each case, MRPD was found using the same procedure described above. These two methods of strength reduction enabled us to evaluate the relative effect of loss of muscle strength at different joint on MRDP when considering typical reductions in  $T_{max}$  associated with aging, and when considering equal reductions in  $T_{max}$ .

Table 5.2 Different cases of reduction in  $T_{max}$  compared to the young model to perform the simulation (- means no change compared to the young model)

Young		Old	$HE_{35\%}$	$HF_{25\%}$	$PF_{35\%}$	$DF_{16\%}$	$HE_{20\%}$	$HF_{20\%}$	$PF_{20\%}$	$DF_{20\%}$
$T_{max}$ (Nm)		Reduction in $T_{max}$ (%)								
$HE$	178	35%	35%	-	-	-	20%	-	-	-
$HF$	125	25%	-	25%	-	-	-	20%	-	-
$PF$	105	35%	-	-	35%	-	-	-	20%	-
$DF$	36	16%	-	-	-	16%	-	-	-	20%

## 5.4 Results

### Human Subjects Testing

The MRPD was 27% smaller ( $p=0.014$ ) among the older group ( $11.8 \pm 2.1$  cm) compared to the young group ( $16.2 \pm 2.6$  cm; Figure 5.7).

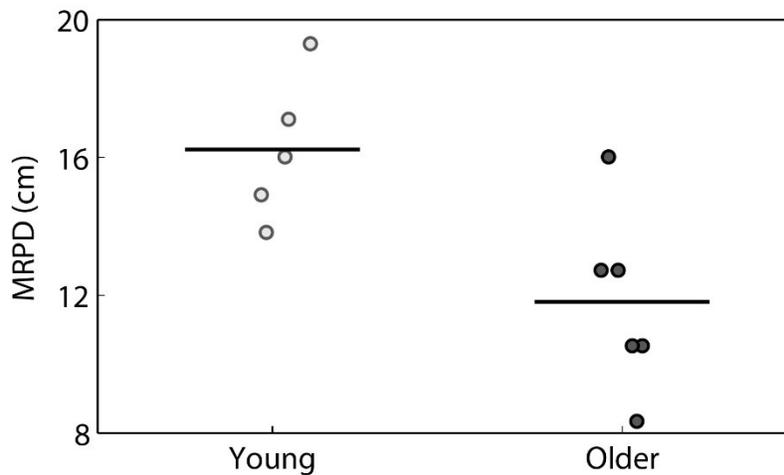


Figure 5.7 MRPD of each subject along with group means indicated by horizontal lines.

### Simulation Study

A representative young male subject was selected to model for the simulation study who exhibited the closest ankle and hip kinematics to the average kinematics observed among young subjects. The MRPD for this representative subject was 19 cm. In step one of the simulation study, it was determined that the maximum joint activation needed to be scaled by a factor of 0.57 in order for the model MRPD to match the MRPD of the representative human subject. The simulation joint angles were similar to the representative human subject's joint angles during MRPD trials (Figure 5.8), with a root-mean-squared error between the simulation and the human subject of 1.0 degrees at the ankle and 2.4 degrees at the hip. The simulation ankle was initially dorsiflexed (peak dorsiflexion at ~300 msec) and the hip initially extended (peak hip extension at ~250 msec) due to the platform displacement. Recovery from the platform displacement then resulted in ankle plantar flexion and hip flexion. Ankle torque in the simulation exhibited a plantar flexor torque immediately following perturbation onset, and which peaked at ~200 msec. Hip torque in the simulation initially exhibited a small hip extensor torque at perturbation onset, followed by a hip flexor torque that peaked at ~250 msec, and then a hip extensor torque that peaked at ~500 msec to decelerate the forward movement of the HAT and bring the model to the upright position (Figure 5.9).

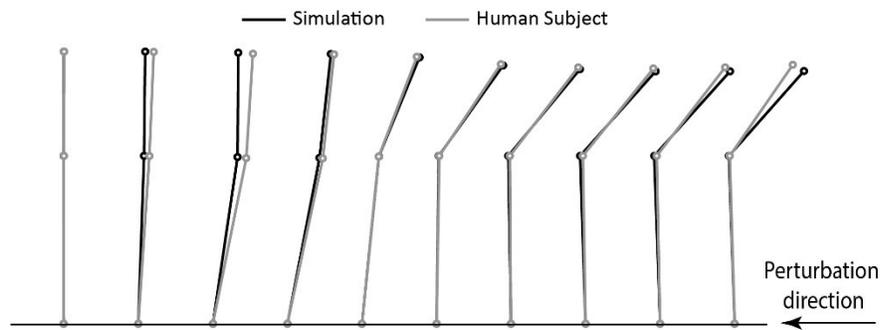


Figure 5.8 Time history body position of the human subject and the simulation during the MRPD trial. Frames are separated by 100 msec.

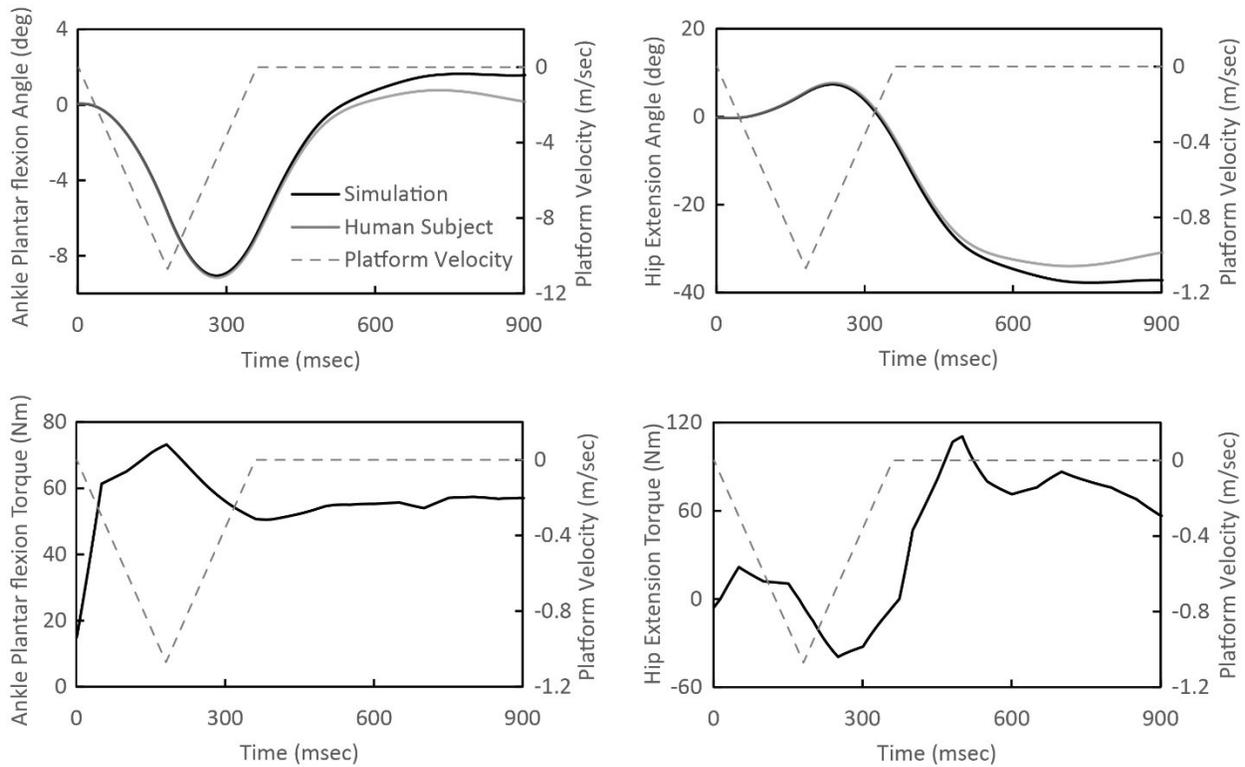


Figure 5.9 Joint angles, torques, and platform velocity at MRPD for the young model

In step two of the simulation study, the MRPD decreased from 19 cm to 12 cm (37%) after applying age-related strength loss to the model. The MRPD of 12 cm after strength loss was similar to the mean MRPD among older human subjects of 11.8 cm. The general pattern of joint angles and torques over time were similar to those prior to strength loss, with some differences in peak magnitudes (Figure 5.10). In particular, peak ankle dorsiflexion angle, peak ankle plantar flexor torque, and peak hip extensor torque were all smaller after age-related strength loss.

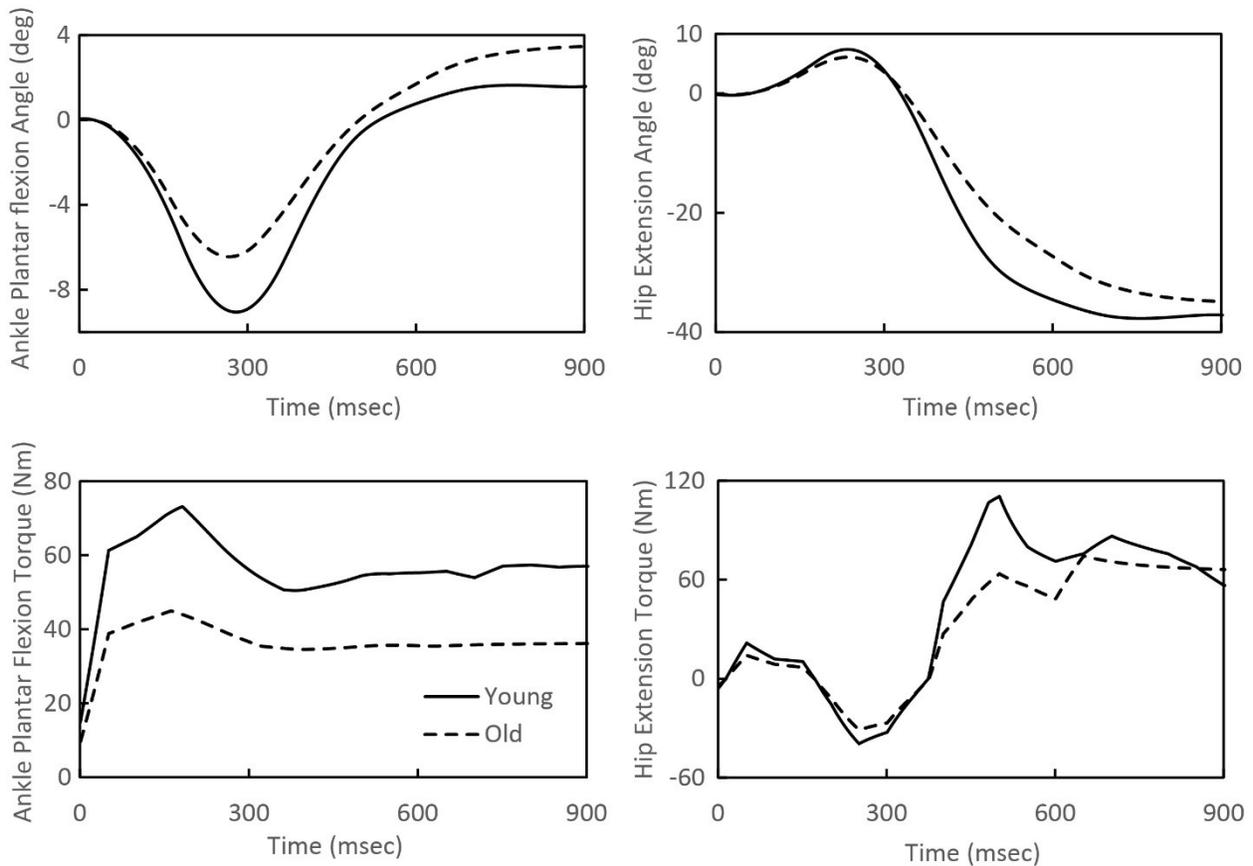


Figure 5.10 Joint torque and angles at MRPD for the young and older models.

In step three of the simulation study, the MRPD was lower after some, but not all applied individual strength losses. Compared to the MRPD value of 19 cm for the young model, MRPD was reduced 37% to 12 cm for  $PF_{35\%}$ , 5% to 18 cm for  $HF_{25\%}$ , and 21% to 15 cm for  $PF_{20\%}$ . MRPD was not affected for  $HE_{35\%}$ ,  $DF_{16\%}$ ,  $HE_{20\%}$ ,  $HF_{20\%}$ ,  $PF_{20\%}$ ,  $DF_{20\%}$  (Figure 5.11).

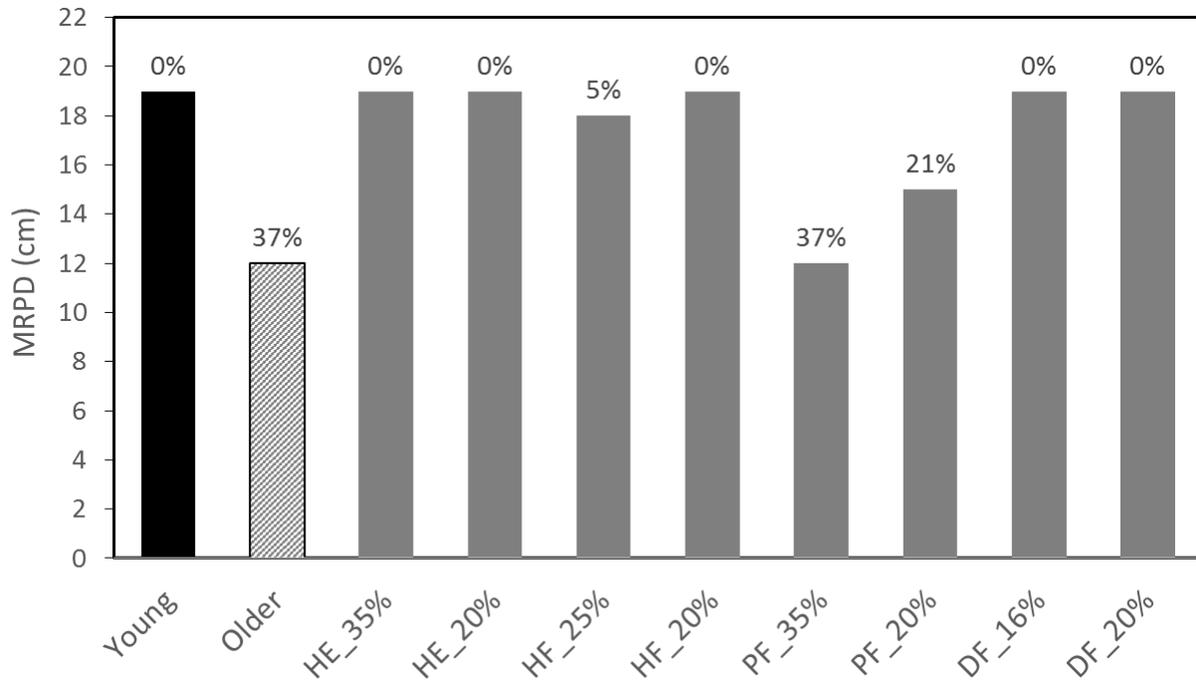


Figure 5.11 Effect of loss of strength on MRPD. Data labels represent the percent reduction in MRPD compared to the young model (19 cm).

Although MRPD was only affected by three cases of strength reduction, joint angles and peak joint torques during recovery differed between many of these cases of strength loss including those not exhibiting a decrease in MRPD (Figure 5.12, Figure 5.13, and Figure 5.14). In cases of  $HE_{35\%}$  and  $HE_{20\%}$ , which exhibited the same MRPD as the young model, the joint excursion was lower at the ankle and the hip compared to the young model. The peak torques did not differ in ankle plantar flexion but was lower in hip flexion and hip extension compared to the young model. In cases of  $HF_{25\%}$  and  $HF_{20\%}$ , which exhibited 5% reduction and no reduction in MRPD, respectively, compared to the young model, the joint excursion was higher at the ankle and did not differ at the hip compared to the young model. The peak torque did not differ in ankle plantar flexion and hip extension, but was lower in hip flexion compared to the young

model. In the cases of  $PF_{35\%}$  and  $PF_{20\%}$  which exhibited 37% and 21% reduction in MRPD, respectively, compared to the young model, the joint excursions was higher at ankle and the hip compared to the young model. The peak torque was lower in ankle plantar flexion, slightly higher in hip flexion and extension in case of  $PF_{35\%}$  and did not differ in hip flexion and hip extension in case of  $PF_{20\%}$  compared to the young model. In the cases of  $DF_{16\%}$  and  $DF_{20\%}$  which exhibited the same MRPD as the young model, the joint excursion and peak joint torques did not differ compared to the young model.

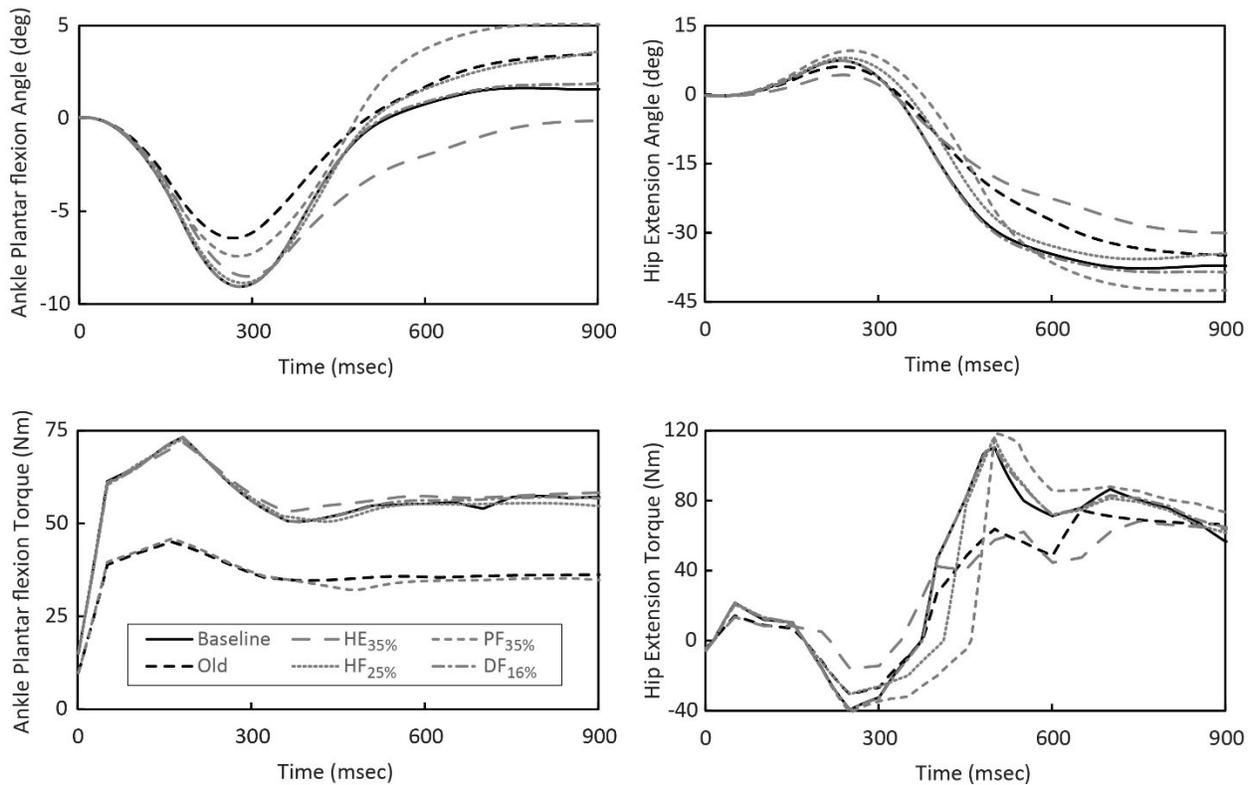


Figure 5.12 Joint torques and angles at MRPD for young and older models, and cases of  $HE_{35\%}$ ,  $HF_{25\%}$ ,  $PF_{35\%}$ ,  $DF_{35\%}$ .

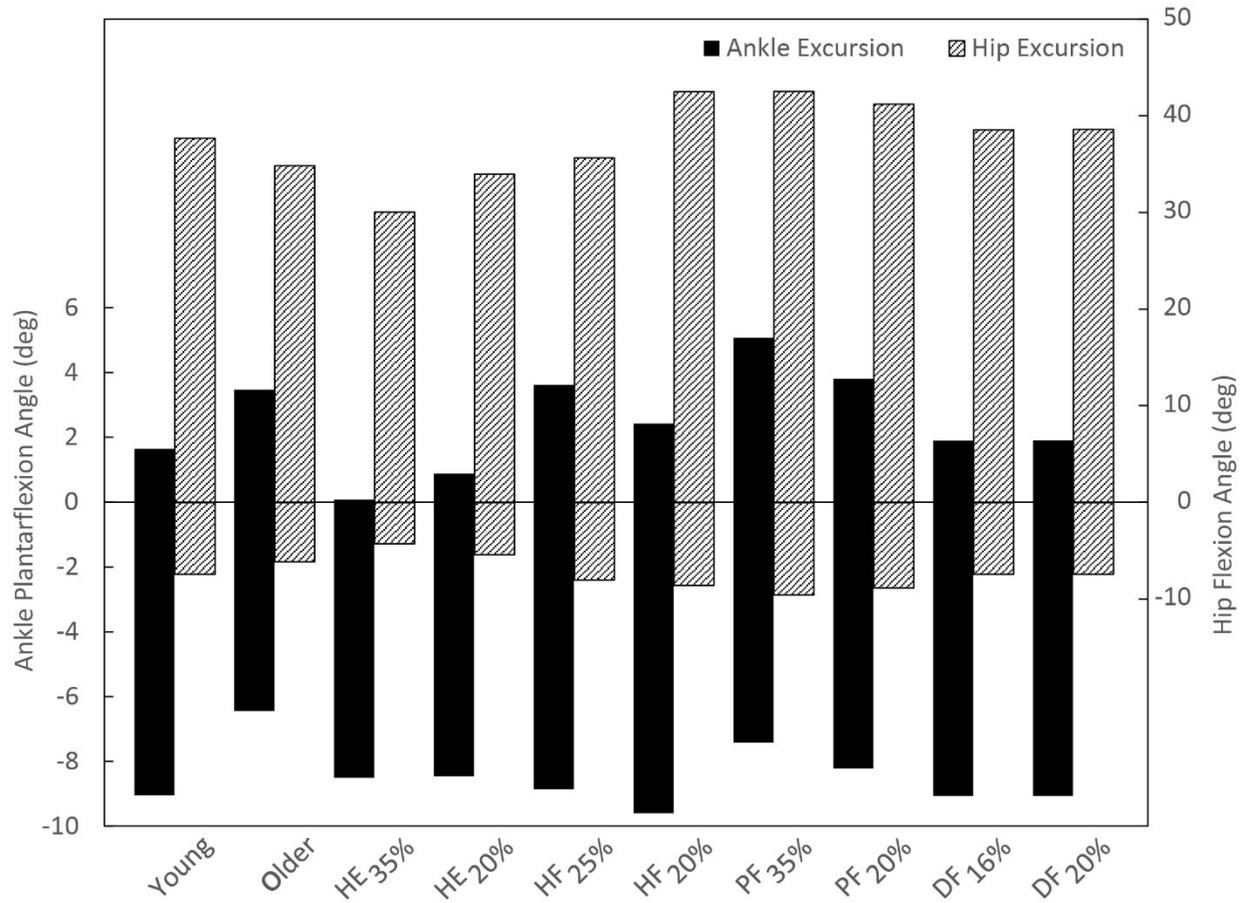


Figure 5.13 Peak joint angles at MRPD for young and older models, and all cases of strength loss. The top of the bars indicate peak ankle plantar flexion/hip flexion angle and the bottom of the bars indicate peak dorsiflexion/hip extension angle. The total height of the bars indicate ankle and hip joint angle excursions.

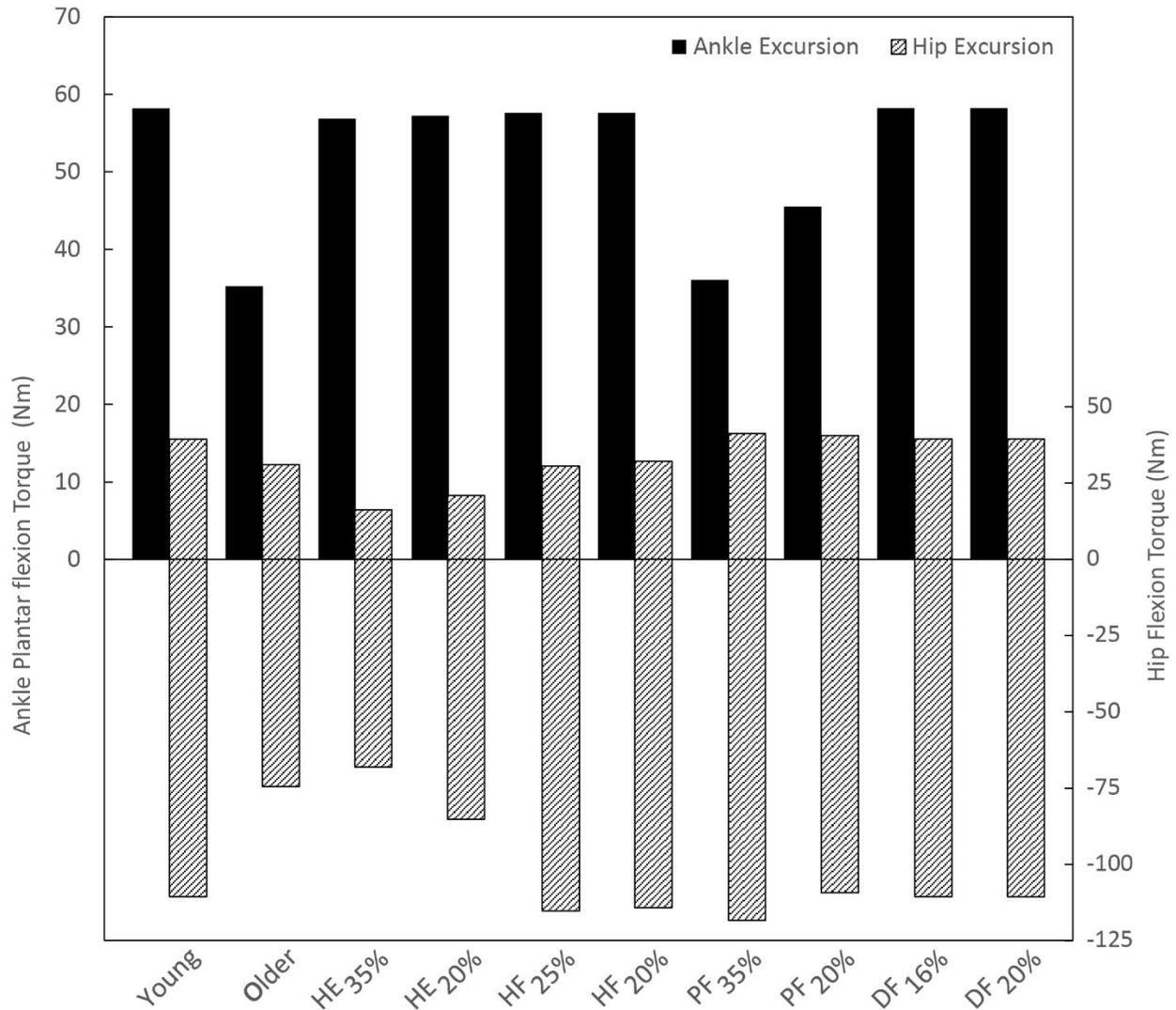


Figure 5.14 Peak joint torques at MRPD for young and older model, and all cases of strength loss. The top of the bars indicate peak ankle plantar flexion/hip flexion torque and the bottom of the bars indicate peak dorsiflexion/hip extension torque. The total height of the bars indicate ankle and hip joint torque excursions.

## 5.5 Discussion

The purpose of this study was to investigate the effects of age-related strength loss on non-stepping balance recovery capability after perturbation while standing, without constraining the

movements to ankle strategy. Two experiments were conducted. In the first experiment, human subjects testing was used to investigate age-related differences in non-stepping balance recovery capability. Our hypothesis was that older adults would exhibit reduced balance recovery capability compared to young adults. This hypothesis was supported because MRPD was 27% lower among older adults. This age-related reduction in MRPD was slightly lower than 36% reduction in maximum lean angle from which subjects recovered using ankle strategy (Mackey and Robinovitch 2006) and lower than 31-47% reduction in maximum lean angle from which subjects recovered by taking a single step (Madigan and Lloyd 2005; Wojcik et al. 1999). Although these studies support an age-related reduction in balance recovery capability, important differences between studies should be noted. The first difference is that the movement strategies allowed during testing (non-stepping vs. single step) differed. Balance recovery using a single step is a more dynamic task that requires knee movement, and greater involvement and coordination of the lower extremity musculature compared to balance recovery without stepping. As such, it may be more sensitive to the adverse effects of age on the neuromuscular system. For example, It has been reported that during balance recovery by stepping, older adults exhibited delays in activating the hip flexors and knee extensors of step leg before and during the swing phase (Thelen et al. 2000). This delay can exacerbate the reduction in balance recovery capability due to the age-related loss of muscle strength and may explain the larger age-related difference in stepping balance recovery capability compared to non-stepping. The slightly lower age-related difference in non-stepping balance recovery observed here (27%) compared to the study by Mackey and Robinovitch (2006) (36%) may be explained by allowing for the use of hip motion in present study. In fact, it has been reported that older individuals relying more on hip strategy to recover balance compared to their young counterparts (Manchester et al. 1989;

Woollacott 1993) therefore, constraining the movement to ankle strategy could have a larger impact on older adults. The second difference between studies is that the perturbation in present study involved translation of a support surface while others involved release from lean angle (Mackey and Robinovitch 2006; Madigan and Lloyd 2005; Wojcik et al. 1999). These different methods would result in differing mechanical and sensory stimuli. For example, while translation of a support surface induces shear force at the foot-sole, such force does not exist in the forward lean method, therefore, there are differences in cutaneous/sensory stimuli at the onset of these perturbations (Mansfield and Maki 2009). Moreover, the difference in point of application of the perturbation could affect the induced pattern of motion (Liu et al. 2003), which may be differentially affected by age.

In the second experiment, simulation study was performed to investigate the effects of age-related strength loss on balance recovery capability. When the muscle strength at all joints/exertions was reduced concurrently as is the case in an older adult (older model), the reduction in MRPD of the model (37%) was in relatively good agreement with the human subject study (27%). During the MRPD trial of the older model, the peak torques at the ankle and hip were reduced, however, the larger reduction in peak ankle plantar flexor torque (38%) compared to the reduction in peak hip flexor torque (21%) was likely due to the fact that the ankle plantar flexor strength was reduced by a larger percentage (35%) than HF strength (25%). These results are consistent with findings that older adults rely more on hip to recover their balance (Manchester et al. 1989; Woollacott 1993).

When reducing strength at each joint/direction individually, the relatively large reduction in MRPD in the cases of  $PF_{35\%}$  (37%) and  $PF_{20\%}$  (21%) emphasize the importance of ankle strength in non-stepping balance recovery (Barrett and Lichtwark 2008; Mackey and Robinovitch 2006; Matrangola and Madigan 2011), even when movement is not constrained to the ankle strategy. In fact, the 21% reduction in MRPD in case of  $PF_{20\%}$  was in good agreement with reported 19% reduction in maximum recoverable lean angle in case of 20% decrease of isometric ankle plantar flexor force when the single segment model recovered using ankle strategy (Barrett and Lichtwark 2008). Consistent with the findings here, middle-aged patients with distal muscle weakness (limb girdle muscular dystrophy) exhibited greater difficulty recovering balance after a backward tilt of a support surface, compared to the patients with proximal muscle weakness (distal spinal muscular atrophy) (Horlings et al. 2009). The small reduction in MRPD in the cases of HF strength loss (5% for  $HF_{25\%}$  and 0% for  $HF_{20\%}$ ) was mainly due to 21% lower peak hip flexor torque since no differences in peak ankle plantar flexion and hip extension torque was observed compared to the young model. This lower peak hip flexor torque may suggest the higher usage of ankle vs. hip compared to the young model.

In case of  $PF_{35\%}$ , the model exhibited 24% higher peak hip flexor and 57% higher peak hip extensor torques compared to the old model, although they exhibited the same MRPD. Moreover, in case of  $PF_{35\%}$ , the joint excursion was 20% higher at the ankle, and 27% higher at the hip. The higher peak torques at the hip and higher hip excursion in case of  $PF_{35\%}$  suggests the greater reliance on the hip during balance recovery compared to the older model, but this greater reliance was insufficient to maintain the same MRPD as in the young model. The higher ankle excursion observed can be due to the higher hip excursion while recovering balance which

moved the lower limb backward and consequently increased the ankle excursion (Runge et al. 1999). The reason for the higher peak hip extensor torque could be to compensate for higher peak hip flexion.

The lack of difference in MRPD between the cases of  $HE_{35\%}$ ,  $HE_{20\%}$ ,  $DF_{16\%}$ ,  $DF_{20\%}$  and the young model suggests lower influence of these muscle groups on non-stepping balance recovery capability. Although the hip extensors were responsible for decelerating the forward movement of HAT and bringing the model to the upright position once the perturbation stopped, balance recovery capability was mostly influenced by the magnitude of ankle plantar flexor torque in early recovery phase. Therefore, the reduction in hip extensor strength did not have a strong influence on MRPD. When comparing the cases of  $HE_{35\%}$ ,  $HE_{20\%}$  to the young model, lower peak hip extension and hip flexion torque was observed in cases of HE strength reduction which resulted in lower hip excursion and consequently lower ankle excursion (a more erect posture) during balance recovery. A possible reason could be that the model did not have the required hip extensor strength to compensate the higher magnitude of hip excursions thus decided recover with smaller hip excursion. To further support this hypothesis, the magnitude of hip excursion was less reduced in case of  $HE_{20\%}$  (20%) compared to the  $HE_{35\%}$  (24%). The lack of effect of dorsiflexor strength reduction in MRPD was mainly due to their small involvement in non-stepping balance recovery. When comparing the cases of  $DF_{16\%}$ ,  $DF_{20\%}$  to the young model, the peak joint torques and joint excursions did not differ.

There are several limitations to this study. First, simulations in this study was performed using a torque-driven model which did not allow the manipulation of specific muscles. Muscle actuators

allow the model to consider bi-articular muscles and co-contraction of muscles. However, previous studies have reported good agreement between human subject and model kinematics using torque-driven model (Ashby and Delp 2006; Yeadon and King 2002) as observed in present study. Second, the model was only two-dimensional, and therefore neglects movements any potentially relevant muscle forces that act out of the sagittal plane. However, no visual asymmetry was observed in the human subject study and the balance recovery stayed in the sagittal plane, therefore the use of two-dimensional model doesn't seem to impact the results. Third, this simplified model did not include a knee joint. The knee flexion during balance recovery when the hip flexion is involved, controls the backward movement of lower limb (Runge et al. 1999) which can reduce the ankle excursion of the model. Therefore, future model can include knee joint to better assess balance recovery strategies. Fourth, we only focused on age-related loss of muscle strength, but aging can impact other neuro-musculoskeletal parameters that were either not explicitly modeled or not our focus.

## **5.6 Conclusions**

Non-stepping balance recovery capability after a perturbation while standing, was reduced among older adults when the movement was not constraint to the ankle strategy. The results of the simulation study suggested that reduced plantar flexor strength plays a major role in reducing the capability to recover balance, even though the movement was not constrained to the ankle. The hip flexor muscle strength also influenced the balance recovery capability when weakened but the magnitude of the impact was significantly smaller than the effect of ankle plantar flexors strength. The findings help improve our understanding of the contribution of age-related strength

loss on non-stepping balance recovery capability, and provide insight on the relative importance of the individual lower extremity muscle groups.

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## **Chapter 6 - Conclusions**

### **6.1 Summary**

Three studies were completed to evaluate how strength is affected by obesity and age, and how these effects influence gait and balance recovery. The first study (Chapter 3) investigated the effects of obesity and age on extension and flexion strength at the hip, knee, and ankle through measuring the isokinetic maximum voluntary contractions. Obese females exhibited lower relative strength (to body mass) at the ankle and hip, similar to the lower relative strength exhibited at the knee. Obese females also exhibited higher absolute strength, but only for 3 of 6 lower limb exertions investigated. This lack of uniformity across the 6 exertions is likely due to the still unclear underlying biomechanical mechanism responsible for these strength differences, which may also be influenced by aging.

The second study (Chapter 4) investigated the effects of obesity, age and their interaction on relative effort at the ankle, knee, and hip during gait through evaluating the relative effort which was expressed by peak NMM during gait as a percentage of the maximum available NMM. Obese females exhibited higher relative effort at the hip, knee, and ankle during gait. The higher relative effort, especially in ankle plantar flexion where little/no reserve in strength was available, may be a major contributing factor to compromised walking ability among obese individuals.

The third study (Chapter 5) investigated the effects of age-related strength loss on non-stepping balance recovery capability after a perturbation while standing, without constraining the movements to ankle strategy. Two experiments were conducted. In the first experiment, human subjects testing was used and the findings suggested that non-stepping balance recovery capability was reduced among older adults. In the second experiment, a simulation study was performed to individually manipulate muscle strength and the results suggested that reduced plantar flexor strength plays a major role in reducing the capability to recover balance, even though the movement was not constrained to the ankle. The hip flexor muscle strength also influenced the balance recovery capability when weakened but the magnitude of the impact was significantly smaller than the effect of ankle plantar flexors strength.

The goal of this dissertation was to understand the mechanisms contributing to mobility limitation and higher risk of falls among obese and older adults. There are three main contributions of these studies. First, this research comprehensively investigated the effect of obesity and age on lower extremity strength, which help in designing strength training interventions for maintaining mobility among obese and older individuals. Second, this research provided valuable information on the effect of muscle strength on gait alterations associated with obesity, and can contribute to strategies to improve and maintain mobility among individuals who are obese. Third, this research investigated the relative effect of age-related loss of muscle strength at different joint/exertions on non-stepping balance recovery while the movement was not constrained to ankle. The effect of aging on balance recovery using ankle strategy has been extensively studied but such effects have not been investigated when the movement is not

constraint to ankle. As such, the results can help guide the development of strength training interventions to reduce the risk of falls among older individuals.

## 6.2 Expected Publications

Table 6.1 Expected publications from studies. Note: \* Accepted

Chapter	Title	Journal
3	Relative strength at the hip, knee, and ankle is lower among younger and older females who are Obese	Journal of GERIATRIC Physical Therapy *
4	Relative effort during gait is higher among obese	Journal of Biomechanics
5	Effect of age-related strength loss on non-stepping balance recovery	Gait and Posture

## 6.3 Future Works

Future research is needed to better understand the findings reported in present study. The first two studies have included only female participants, therefore future studies are required to generalize the findings to both genders. In the second study, the maximum isometric available net muscle moment was utilized to estimate the relative effort during gait. A future study which account for the effect of angle and angular velocity on maximum available net muscle moment can more accurately evaluate the obesity-related difference in relative effort. Moreover, the study only looked at the sagittal plane kinematics and kinetics. It would be beneficial to perform similar analysis on the relative effort in other planes in future studies. Finally, the study only investigated joint net muscle moments and the related relative effort. Further investigation of the joint power among obese and applying the concept of relative effort to those measurements can

give more insight into effect of obesity on gait/mobility. In the third study, the model developed to evaluate the effects of age-related loss of muscle strength on balance recovery capability, can be utilized to investigate the effect of other age-related changes of neuromuscular parameters on non-stepping balance recovery. Furthermore, the future model can benefit from knee joint to more accurately evaluate model response to perturbation.