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THE INFLUENCE OF PASSIVE ANKLE JOINT POWER ON BALANCE RECOVERY

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THE INFLUENCE OF PASSIVE ANKLE JOINT POWER ON BALANCE RECOVERY

By

Stephanie E. Hamilton

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Preface

Chapters 2-4 of this dissertation contain writing and data that has not been published but is planned for future submission in three successive papers. Dr. Karen Roemer, Dr. Rupak Rajachar, and Dr. Sean Kirkpatrick provided content and guidance for the introduction and discussion sections of each of these chapters. Dr. Karen Roemer assisted in the design of the methodology for these papers. Dr. Martyn Smith assisted with statistical design for the results section of each of these chapters. Stephanie E. Hamilton recruited subjects and collected data for each of these studies. Additionally, Stephanie compiled the ideas provided by her advising committee and wrote each of the sections included in each of these chapters incorporating the ideas presented by each committee member, as well as performing the data analysis as outlined by the plan created with Dr. Smith's assistance.

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Abstract

Over one–third of Americans over the age of 65 fall each year, costing more than $19 billion in health care costs in 2000. Many adults 65+ who have not experienced a fall still fear falling, and fear can decrease quality of life and increase the likelihood of falls. Several factors such as muscle strength, power, stiffness and tendon properties change in the human body with age affecting balance, which has been tagged as a fall risk predictor. Additionally, balance recovery strategies also differ between young and older adults, with young adults primarily utilizing their ankle joint and older adults utilizing their hip. The role of passive ankle joint power in balance recovery is unknown. Therefore, we conducted three studies. In Study 1, we investigated the role of passive ankle joint power in balance recovery of young subjects and tested if the contribution of passive power to net ankle joint power changed with perturbation speed. In Study 2, we explored the factor of age in the contribution of passive ankle joint power to net ankle joint power. In Study 3, we searched for a link between the contribution of passive ankle joint power to net ankle joint power and balance recovery strategy. Passive joint torque through the full range of motion was collected for each subject. Each subject performed 5 stepping tasks at two speeds, fast and slow. Joint kinematics and kinetics were collected for each trial. Inverse dynamics were performed and net ankle joint torque and net ankle joint work were computed. Passive ankle joint torque models were optimized for each subject, and passive ankle joint powers were determined. In Study 1, there appeared to be no difference in net or passive joint powers with respect to perturbation speed. In Study 2, age affected net ankle joint powers and passive uniarticular plantar- and dorsiflexor powers. In Study 3, we noted a change in balance recovery strategy between young and older adults. We were unable to predict balance recovery strategy index based off of the percent contribution of passive ankle joint work to net ankle joint work. These studies bring greater clarity to the role of passive ankle joint power with respect to balance recovery.

Chapter 1
Literature Review

1.1 Fall Risk and Balance

According to the Administration on Aging (AOA), 40.4 million people in the United States are age 65 or older, a population which has increased 15.3% since 2000. The AOA also reports that one-third of Americans over the age of 65 experience falls each year, many of which result in injury and sometimes loss of independence. Health care costs for fall-related injuries are soaring with more than $19 billion spent in 2000 and a projected $54.9 billion in 2020. Additionally, the fear of falling can limit mobility and decrease the quality of life, as well as potentially increasing the likelihood of falling due to decreases in physical ability (Murphy & Isaacs, 1982).

Because of the widespread nature of this problem, several researchers have searched for predictors of fall risk. Graafmans et al (1996) performed a prospective study on fall risk factors in aging adults with the goal of creating a fall risk profile to categorize the strongest fall predictors. This research group investigated several mobility measures including balance measured during a tandem-stance, leg-extension strength during chair-stands, and walking ability. Participants also underwent cognitive function testing and blood pressure measurements. Of the mobility impairments discovered, 48% of fall prevalence was due to an inability to perform the balance task. Similarly, Lord et al (1994) found that postural stability measures were age-independent predictors of falls in community-dwelling adults over the age of 60. Stenhagen et al (2013) conducted a longitudinal study over 6 years to pinpoint fall risk factors in aging adults. They investigated several risk factor categories such as medical and psychological health, medication use, sensory and neuromuscular issues, and balance and mobility capabilities. Subjects that underwent falls in the course of the study indicated that reduced mobility, heart dysfunction and functional impairment were three main fall predictors. Several other studies underline postural stability measures such as the Berg Balance Test, Tinetti Scale, and postural sway measurements as indicators of fall risk (Goncalves, Ricci, & Coimbra, 2009; Kulmala et al., 2007; Robbins et al., 1989; Thapa, Gideon, Brockman, Fought, & Ray, 1996; Tinetti, Speechley, & Ginter, 1988; Toraman & Yildirim, 2010).

Some researchers have tested the connection of postural stability to fall risk through intervention programs. While some studies seemed to demonstrate immediate benefit from exercise interventions, most early intervention programs struggled to reduce number of resultant falls, possibly due to improper choice of exercise protocol or the level of frailty of the study participants (Hornbrook et al., 1994; Hu & Woollacott, 1994; Mulrow et al., 1994; Reinsch, Macrae, Lachenbruch, & Tobis, 1992). However, some more recent work has demonstrated success with exercise interventions and fall reduction. Barnett *et al* (2003) saw a reduction of rate of falls (40%) in the following 12 months in the exercise group versus the control. Lord *et al* (2003) also observed a decreased rate of falls (22%) for the weight-bearing exercise group compared with the control. Due to these findings, the U.S. Preventive Services Task Force issued a recommendation for community-dwelling adults ages 65 and over to participate in exercise or a physical therapy program to reduce or prevent fall risk (Moyer & Force, 2012).

1.2 Age-Related Musculotendon Changes

As the human body ages, it experiences changes in musculotendinous factors that affect balance and mobility. Research has identified several musculature differences in the aging population compared with young adults that influence balance. These include reduced strength, reduced cross-sectional area of Type II muscle fibers, reduced power, increased stiffness, and changes in muscle activity and tendon properties (Billot, Simoneau, Van Hoecke, & Martin, 2010; Hasson & Caldwell, 2012; Hortobagyi et al., 1995; Ikezoe, Asakawa, Fukumoto, Tsukagoshi, & Ichihashi, 2012; Izquierdo, Aguado, Gonzalez, Lopez, & Hakkinen, 1999; Larsson, Grimby, & Karlsson, 1979; Onambele, Narici, & Maganaris, 2006; Sorock & Labiner, 1992; Studenski, Duncan, & Chandler, 1991; Thelen, 2003; Whipple, Wolfson, & Amerman, 1987). Each of these changes directly affects force production in amount and velocity of contraction, altering how the body responds to, and ultimately can handle an external force. This review will address specifically changes in muscle strength, muscle power and tendon stiffness as they contribute to net joint torque production, and ultimately, human balance.

1.2.1 Muscle Strength

Various studies have addressed the relationship of reduced strength, or decreased muscle force, and decreased balance. Several things change in muscle tissue with age, resulting in decreased force production. One of the most documented changes is the change with respect to muscle fibers. First, it appears that the cross sectional area of muscle tissue decreases with age (Izquierdo, Ibanez, et al., 1999). Larsson *et al* (1979) acknowledged an atrophying of Type II muscle fibers with age which could explain a portion of the overall muscle atrophy. These changes affect the amount of force production. Carter *et al* (2002) found that knee extension strength was highly correlated with both static and dynamic balance measures in older adult females. Brech *et al* (2013) had similar findings regarding knee flexion and extension strength relating to the sit-to-stand test, another measure of postural control. Ribeiro *et al* (2009) investigated improvements on plantar and dorsiflexor muscle strength and balance using the Functional Reach Test. Their work demonstrated that improving strength in the plantarflexor muscles specifically aided balance measures. Barrett and Lichtwark (2008) found through simulation that when both maximal force output of a muscle and muscle excitation were decreased, the maximal recoverable lean angle (MRLA) (or the angle of forward tilt between the trunk and the line perpendicular to the ground surface in an ankle strategy for successful balancing following perturbation) was diminished.

Studies have probed the question of whether or not changes in muscle strength with age stem partially from changes in muscle activity. Muscle activity is the electrical activity that spreads across the muscle due to the incoming neural signal. The changes in activity result in contraction or relaxation. Klein *et al* (2001) investigated changes in muscle strength in young and old men and also measured muscle activity through electromyography (EMG) of the arm flexors and extensors. They found an increase in coactivation, or activating of opposing muscles at the same time, in the older men. Schmitz *et al* (2009) examined muscle activation during gait and also found increased coactivation in the ankle and knee muscles. Increases in coactivation can affect overall movement as it reduces net joint moments. It does appear though, through a study by Reeves *et al* (2004), that muscle coactivation can be decreased through a training regimen. In addition to coactivation changes, Kubo *et al* (2007) found a decrease in muscle activation during a maximal isometric voluntary contraction (MVC) of the plantarflexor muscles in older adults compared with young, meaning the muscles "turned on" at a lower rate. This could be indicative of a cause of muscular strength decrease in older adults.

Muscle force is the key component in active joint torque. Muscle force can be modulated through contraction through information gained from stretch receptors as well as other visual and sensory receptors. Active joint torque is necessary for humans to complete their desired movements.

1.2.2 Muscle Power

Muscle contraction velocity decreases with age, resulting in decreased muscle power (Hasson & Caldwell, 2012; Larsson et al., 1979). This problem is compounded by decreased muscle strength. Muscle power is a measure of muscle force multiplied by contraction velocity. DeVito *et al* (1998) performed a study on aging females and measured variables such as force, power and velocity during a series of jump movements and found that peak power output was a significant indicator of functional decline in aging females due to the decrease in maximal contraction velocity. Yu *et al* (2007) investigated changes with age in both genders in single muscle fibers and found a significant change in maximum contraction velocity. Interestingly, Yu also performed knee extension exercises in this same study and found that there was a marked torque decrease at faster speeds in the aging population. Peak power measures have been proposed as a more critical variable in determining functional limitations compared with muscle strength, and this muscular change exhibits a decline earlier in aging adults (Metter, Conwit, Tobin, & Fozard, 1997; Suzuki, Bean, & Fielding, 2001). Whipple *et al* (1987) identified a decreased power output of the ankle musculature at higher contraction velocities as an indicator of fall risk. Other studies have attributed change in recovery strategy from a balance perturbation to inadequate power production at lower extremity joints (Carty, Cronin, Lichtwark, Mills, & Barrett, 2012). Improvements in muscle power compared with muscle strength have been shown to generate more significant changes in performance-based measures (Bean et al., 2010). Orr *et al* (2006) identified specific applications of power training to balance with an intervention program targeting adults over 60 years old with several power training protocols, finding that power training improved balance measures. Piirainen *et al* (2014) confirmed that various types of power training resulted in improvement in dynamic balance measures. While it is apparent that aging adults experience changes in muscle strength, it seems that muscle power measures provide even greater insight into mobility changes, particularly regarding balance, as it accounts for both changes in muscle strength and contraction velocity.

1.2.3 Tendon Stiffness

The tendon itself plays a very important role in causing movement. The muscle fibers contract, producing active force; those fibers connect to the tendon which transmits the force to bone.

A tendon has inherent mechanical properties which causes an augmentation of the force that muscle fibers produce through contraction. Tendons behave similarly to a rubber band, such that when no load is placed upon it causing it to lengthen or when the load causes it to lengthen to just below its slack length, it produces no force. The force the tendon produces is directly related to its stiffness, or its resistance to length change. Once the tendon lengths above the slack length, it first exhibits low stiffness due to its crimp structure flattening, producing lower forces. After the crimp flattens, the tendon produces linear force due to greater stiffness.

Tendon stiffness changes throughout one's lifetime. Stiffness can also be described using its inverse property, compliance. Waugh et al. (2012) observed that tendon stiffness increases from childhood to adulthood. As adults age, however, their tendons become less stiff, or inversely, more compliant. Stenroth et al. (2012) found that adults ages 70-80 years old had 17% greater Achilles tendon compliance compared with adults ages 18-30 years old.

Because of the way that muscle connects to tendon and tendon to bone to transmit force, tendon force can be represented as the product of muscle force and pennation angle, or the angle at which the fibers are aligned to the tendon. If the tendon becomes more compliant, this will affect the amount of muscle force that is transmitted to the bone through the tendon.

The force that the tendon produces through its mechanical makeup contributes to the passive force. The muscle fibers also produce some passive fibers due to their own mechanical makeup as well as the connective tissue that binds them together. These passive forces are the force component responsible for passive joint torque, a torque that is strictly based on properties of the muscle and tendon and cannot be altered through voluntary contraction.

1.3 Balance Recovery

During balance recovery from a perturbation due to a trip or a step, an individual's movement involves flexion and extension of joints in order to move their center of mass (COM) to within their base of support. This is due to the direction of the perturbation as well as the inherent instability in the skeletal structure anterior-posteriorly. As balance recovery is investigated, focus begins at the lower extremity joints and moves upward as the lower joints affect and sometimes enhance the movement of joints above. The most common joints investigated are the ankle and hip joints.

Several researchers have classified balance recovery dependent on joint "strategy" employed to regain stability. Some of these methods include the ankle strategy and the hip strategy, shown in Figure 1.1 (Nashner & McCollum, 1985). Investigation into these methods provides another layer of understanding regarding human balance. Additionally, while most of the following studies present data on normal stance, Clifford and Holder-Powell (2010) demonstrated that these same classifications can also be applied to single leg stance, as they discerned different recovery strategies between dominant and non-dominant legs.

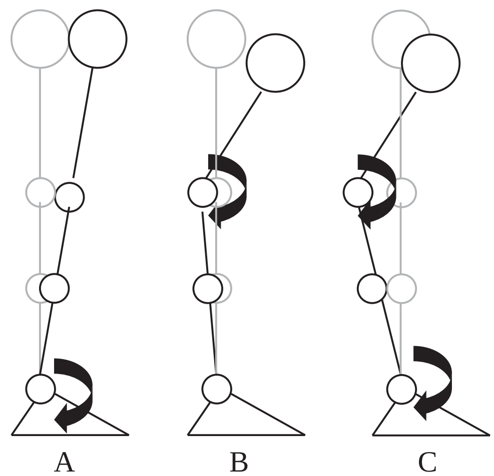

Figure 1.1: Demonstration of the various types of fixed-support balance recovery strategies; A: Ankle Strategy; B: Hip Strategy; C: Mixed Strategy

Many studies have examined these strategies to determine why a particular strategy is chosen. It appears that circumstances of the environment as well as the individual, such as footing type or size of surface, different diseases or health conditions, lower extremity muscle weakness, the size of the perturbation, the goal of postural recovery (minimization of muscle activation, etc.), and the velocity of perturbation, affect the choice of recovery strategy (Carty, Barrett, Cronin, Lichtwark, & Mills, 2012; F. B. Horak & Nashner, 1986; F. B. Horak, Nashner, & Diener, 1990; Kuo, 1995; Kuo & Zajac, 1993b; Runge, Shupert, Horak, & Zajac, 1999; Zajac, 1993)

1.3.1 Ankle Strategy

When an individual experiences a perturbation and ankle joint torque is sufficient to regain stability, it is referred to as the “ankle strategy” (F. B. Horak & Nashner, 1986). Typically, the ankle strategy is utilized by individuals undergoing smaller postural perturbations at lower velocities, performing less demanding tasks, or balancing on a larger support surface

(Fujisawa et al., 2005; Fay B. Horak, 2006; Runge et al., 1999). The ankle strategy also appears to be related to lower frequency sway movements (McCollum & Leen, 1989).

In order to model the ankle strategy, researchers have adopted the single inverted pendulum to represent the movement (Gage, Winter, Frank, & Adkin, 2004; Karlsson & Frykberg, 2000; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). The shank, thigh, and trunk move as one body, pivoting around the ankle joint. The single inverted pendulum states that the horizontal acceleration of the COM of an individual is proportional to the difference between the COM and center of pressure measures (Murray, Seireg, & Scholz, 1967).

1.3.2 Hip Strategy

When an individual cannot produce adequate ankle joint torque to recover from perturbation, often the hip strategy is utilized to regain balance successfully. The hip strategy involves the hip musculature being activated to control COM motion instead of the ankle joint. Kuo and Zajac (1993a) observed that the hip strategy appears to be more efficient at controlling COM, due to it requiring less muscle activation. The hip strategy is often utilized during perturbations of higher velocities, those of higher sway frequency, in situations that the support surface is smaller, and when the COM's vertical component falls in front of the base of support (Fujisawa et al., 2005; Johansson, Magnusson, & Akesson, 1988; McCollum & Leen, 1989; Runge et al., 1999).

Pure hip strategies, where the hip is acting with no assistance from the ankle joint, can also be modeled as a single inverted pendulum. More typically though, the ankle acts at a smaller degree compared with the hip, and a mixed strategy is born. This strategy can be modeled as a double inverted pendulum, with pivots located at both the ankle and the hip. The ankle and hip joints move counter-phase with each other.

1.3.3 Phases of Balance Recovery

There are several measurements used in the literature to quantify balance. One measurement that is gaining popularity due to the insight it provides is the measure of time to stabilization, the time it takes a subject to recover from a perturbation and to stabilize their center of mass (Gribble & Robinson, 2009). Jonsson et al (2004) detected two different phases in balance recovery, dynamic and static. In the dynamic phase, there is a rapid change in force variability. When a subject achieves the static phase, a certain level of force variability is maintained. Dingenen et al (2013), in agreement with the previous study, found that after 4 seconds from transitioning from a double-legged stance to a single-legged stance, subjects had reached stabilization. Other recent research has been able to better describe age-related balance recovery differences by segmenting balance recovery into phases (da Silva, Bilodeau, Parreira, Teixeira, & Amorim, 2013; Erika Jonsson, Henriksson, & Hirschfeld, 2007; Parreira et al., 2013; Roemer & Raisbeck, 2015).

1.3.4 Age-Related Changes in Balance Recovery

Age appears to be a factor with respect to balance recovery strategy. When considering the lower extremity joints, several researchers have noted age-related changes in the ankle joint musculature as being linked to falls or changes in balance, specifically the dorsiflexor muscles (Daubney & Culham, 1999; Studenski et al., 1991; Whipple et al., 1987). Manchester *et al* (1989) found that in certain balance scenarios, older adults utilized a proximal-to-distal muscle sequencing strategy, which encompasses the hip strategy, significantly more frequently than younger adults.

In order to investigate use of the ankle strategy and age, several studies have measured the maximal recoverable lean angle (MRLA) in the ankle strategy to determine a threshold before other strategies are employed. Mackey and Robinovitch (2006) found significant reductions in MRLA in elderly females compared with young. Hsiao-Wecksler and Robinovitch (2007) also found that regardless of recovery strategy employed, the younger women could undergo greater MRLA’s compared with older women. Barrett and Lichtwark (2008) created a simulation of a single inverted pendulum model and used muscle excitations that allowed for an MRLA of 7.2^{o}. Several parameters representing muscle and tendon tissue responses were altered to reflect aging musculotendon tissue to determine how the factors individually and collectively affected balance recovery capabilities. While several factors affected MRLA individually, the combined affect resulted in a 23.1% change in MRLA compared with the baseline condition.

While differences in balance recovery strategy with age have been established, the cause of these changes is unknown. It appears, as stated above, that the changes in the dorsiflexor muscles affect balance capabilities. The plantar flexor muscles, prime movers at the ankle joint in stabilizing and dynamic movements including balance recovery, have been shown to have decreased force production and reduced peak torque with age (Morse, Thom, Reeves, Birch, & Narici, 2005; Thelen, Schultz, Alexander, & AshtonMiller, 1996). These muscle changes could present a mechanism for the change in balance recovery strategy. Investigation into musculotendinous tissue properties is necessary to evaluate the location of this mechanism.

To date, the interaction of muscle stiffness and power have not been considered in the plantar flexors during balance recovery. The change of these muscular parameters with age may be integral components reducing the ability to produce adequate ankle joint torque to recover from perturbation using ankle strategy. Insight into the effects these changing muscle parameters have on balance recovery could improve fall risk assessments as well as guide specific intervention programs designed to reduce the risk of falls in aging adults.

1.4 Musculotendon Unit Passive Joint Torque Model

The muscle-tendon unit (MTU) is a complex organism. There are portions of the MTU that produce force and others that dampen or augment force dependent on their tissue characteristics. Understanding these components and how they contribute to muscle function is crucial for understanding human movement. Additionally, proper identification of changes within the MTU could lead to treatment of patients with conditions such as cerebral palsy or other disorders. Though there have been advances in technology, some MTU tissue parameters cannot yet be measured directly *in vivo*. Therefore, in order to investigate the MTU's passive components in this study, a passive joint torque model will be used.

1.4.1 Hoang et al. Passive Torque Model

Mechanical properties of the MTU can be estimated non-invasively by using a passive torque model developed by Hoang et al. (2005). Hoang's model was extended from a model developed by Herzog et al. that determined the active force-length relationship of muscle (Herzog, Read, & Terkeurs, 1991; Herzog & Terkeurs, 1988).

1.4.1.1 Model Assumptions

This model is based on several assumptions that must be stated. First, this model separates biarticular muscles from uniarticular muscles. There are only 2 muscles which cross both the ankle joint and knee joint, namely, the gastrocnemius and the plantaris. The plantaris is a small muscle that many people do not have. Because of this, it is assumed that biarticular passive MTU forces are due strictly to the gastrocnemius.

The gastrocnemius and the uniarticular muscles, both plantar- and dorsiflexors, were assumed to possess an exponential torque-angle relationship. This assumption also appears to provide a good fit for passive torque-angle curves in the index finger (Esteki & Mansour, 1996).

Finally, it was assumed that passive torques arise primarily from the uniarticular plantar- and dorsiflexor muscles as well as from the gastrocnemius. Any passive tension that may be accrued from other tissues such as nerves, skin and blood vessels was considered neglible.

1.4.1.2 Model Description

The model is based on the exponential torque-angle relationship. Data collected during experimental passive trials is fit to the model, optimizing 3 parameters for each of the following groups: uniarticular plantarflexors, uniarticular dorsiflexors, and the gastrocnemius muscle. The following equation was used to model passive ankle joint torque (τ_{ankle}):

Equation 1.1

$$\tau_{ankle}\{\theta_a, l_g\} = \left(a_p * e^{k_p(\theta_a-\theta_P)} - a_p\right) \qquad \theta_a > \theta_P$$
$$+\left(a_d * e^{k_d(\theta_D-\theta_a)} - a_d\right) \qquad \theta_a < \theta_D$$
$$+m_g * \left(a_g * e^{k_g\left(l_g-l_G\right)} - a_g\right) \qquad l_g > l_G$$

where θ_a is instantaneous ankle angle; l_g is instantaneous Gastrocnemius length; $a_p, k_p, a_d, k_d, a_g, k_g$ are stiffness constants for structures crossing the plantar, dorsal aspects of the ankle, respectively, as well as the gastrocnemius muscle; θ_P, θ_D are the ankle angles at which the respective uniarticular structures go slack; m_g refers to the gastrocnemius moment arm at the ankle; and l_G is the gastrocnemius slack length.

Gastrocnemius length and moment arm were calculated using the formulation put forward by Grieve et al. (1978),

Equation 1.2

$$l_g = l_{ref} + (A_0 + A_1\theta_a + A_2\theta_a^2 + K_0 + K_1\theta_k + K_2\theta_k^2) * l_s/100$$

Equation 1.3

$$m_g = \frac{1.8}{\pi} * l_s * (A_1 + (2A_2\theta_a))$$

where l_{ref} refers to the distance between the lateral epicondyle of the femur and lateral malleolus of the fibula with the knee and ankle at 90°; $A_0, A_1, A_2, K_0, K_1, K_2$ are coefficients found in Table 1.1; θ_k is instantaneous knee angle; and l_s is shank length .

Table 1.1: Coefficients for gastrocnemius length and gastrocnemius moment arm (Grieve et al., 1978)

A_0	A_1	A_2	K_0	K_1	K_2
-22.18468	0.30141	-0.00061	6.46251	-0.07987	0.00011

1.5 Summary and Hypotheses

The purpose of this dissertation is three-fold. First, we aim to determine if muscle-tendon stiffness, calculated as passive ankle joint power contributes to net ankle joint power differently under different movement velocity conditions. Second, two different age groups will be investigated to determine if this contribution of passive joint power to net joint power is affected by age in addition to movement velocity. Third, overall balance recovery strategy will be considered in order to determine if there is a link between increasing passive mechanisms due to increasing muscle-tendon stiffness and the inability to control balance recovery with the ankle joint alone.

Hypothesis 1*:* As the movement velocity increases, the demand placed upon the muscle-tendon unit to produce force increases. Muscles must contract more rapidly to adhere to the demand placed upon them by the particular movement in order for the movement to be completed successfully. Julian and Sollins (1975) found that in isolated frog skeletal muscle fibers, as contraction velocity increased, muscle stiffness also increased. Because muscle stiffness arises out of a muscle fiber's resistance to length change, as a fiber undergoes greater length change due to a higher contraction velocity, the passive force and thus passive joint power due to muscle-tendon stiffness should increase. It is hypothesized that increased stepping velocity will result in an increased contribution of passive joint power to net joint power.

Hypothesis 2*:* Previously in this review, it has been documented that older adults experience an increase in muscle stiffness and an increase in tendon compliance. Because these quantities are often investigated separately, it is unknown how the passive joint powers in the older adults will compare with the young. Theoretically, because of this increase, an older adult should produce greater passive forces during the same movement when performed by both older adults and young. Thelen (2003) found that older adults produced greater passive forces compared with young adults in experimental data from old and young adults performing ankle joint movement with an age-adjusted musculoskeletal model. It is hypothesized that the contribution of passive joint power to net joint power with increasing velocity will be greater in the elderly compared with their young counterparts.

Hypothesis 3*:* Due to musculoskeletal changes with age, older adults have demonstrated decreased ankle joint moments and lower rate change in moment generation during recovery movements compared with young adults (Mackey & Robinovitch, 2006; Pijnappels, Bobbert, & van Dieen, 2005). Because of increased contribution of passive joint power to net power in older adults, the rate torque generation at the ankle may be insufficient to allow balance recovery to be obtained with the "ankle strategy". It is hypothesized that increased contribution of passive joint power to net joint power will reduce the amount of 'active control" of the ankle joint, resulting in a "hip strategy" or a "mixed strategy" for balance recovery.

Chapter 2

The Influence of Perturbation Speed on Passive Ankle Joint Power Contribution

2.1 Introduction

According to the Administration on Aging (AOA), 40.4 million people in the United States are age 65 or older, a population which has increased 15.3% since 2000. The AOA also reports that one-third of this population experience falls each year, resulting in injuries and loss of independence. Health costs for fall-related injuries are soaring, with $54.9 billion projected in 2020. Additionally, fear of falling can limit mobility, decrease quality of life, and increase likelihood of falling due to decreased physical activity (Murphy & Isaacs, 1982). Researchers have linked decreasing balance as a predictor of fall risk (Goncalves et al., 2009; Graafmans et al., 1996; Thapa et al., 1996; Toraman & Yildirim, 2010).

As the human body ages, it experiences changes in musculotendinous (MTU) factors that affect balance and mobility, including reduced muscle force, reduced muscle power, increased stiffness, and changes in muscle activity and tendon properties (Hasson & Caldwell, 2012; Ikezoe et al., 2012; Izquierdo, Aguado, et al., 1999; Onambele et al., 2006; Studenski et al., 1991; Thelen, 2003). Each of these changes directly affects force production and contraction velocity, altering how the body responds to an external force and contributes to net force resulting in movement.

Research has addressed the relationship of decreased muscle force to decreased balance (Barrett & Lichtwark, 2008; Brech et al., 2013; Ribeiro et al., 2009). Peak muscle power, affected by contraction velocity and muscle strength, has been proposed as a more critical variable in determining functional limitations compared with muscle force, exhibiting declines earlier in aging adults (Hasson & Caldwell, 2012; Metter et al., 1997; Suzuki et al., 2001). Studies have attributed inadequate power production at lower extremity joints to fall risk and changes in balance recovery strategy to return to balance after a perturbation (Carty, Cronin, et al., 2012; Whipple et al., 1987).

Previous literature has highlighted the necessity of breaking a balance task into phases, dynamic and static (Erika Jonsson et al., 2007; E. Jonsson et al., 2004; Parreira et al., 2013; Roemer & Raisbeck, 2015). Different balance measures such as sway area, sway velocity, directional sway and force variability undergo rapid changes during the dynamic phase of balance. Once subjects reach the static phase, stabilization takes place. The dynamic phase has been shown to last about 4 seconds (Dingenen et al., 2013). Assessing this phase is an important piece in evaluating a person's balance capabilities.

Joint power production is the product of net joint torque and joint angular velocity. Joint torque is composed of active torques, those gained through muscle contraction, and passive torques, those acquired through tissue properties. Lately, passive torque has been highlighted as a contributor to movement, at times resulting in impairments (Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998; Silder, Whittington, Heiderscheit, & Thelen, 2007; Whittington, Silder, Heiderscheit, & Thelen, 2008). Changing passive forces in the musculature surrounding the ankle has been shown to affect gait and balance capabilities (Ho & Bendrups, 2002; Muraoka, Muramatsu, Takeshita, Kanehisa, & Fukunaga, 2005; Whittington et al., 2008).

Humans perform movements of varying velocities throughout their daily activities. Higher velocities predispose individuals to falls. It is unknown how passive torque contributes to net joint torque as an individual recovers from a perturbation affecting balance. It is hypothesized that increasing stepping velocity will increase the contribution of passive joint torque to net joint torque in the ankle during balance recovery.

2.2 Methods

2.2.1 Subjects

The experiment was performed on 21 subjects (11 males and 10 females, ages 20-30). Subject characteristics are detailed in Table 2.1. Subjects were free from diabetes, neuromuscular disorders, or lower extremity surgeries. Subjects completed a questionnaire regarding their physical fitness habits to ensure a recreationally active sample. Each subject provided informed consent using procedures approved by the Michigan Technological University Institutional Review Board.

Table 2.1: Subject characteristics; group means are shown with standard deviation.

	Females	**Males**	**Total**
Age:	23.8 (3.5)	22.9 (3.5)	23.3 (3.5)
Height (cm):	165.3 (7.7)	180.4 (6.8)	173.2 (10.5)
Weight (kg):	67.34 (9.33)	78.19 (9.34)	73.02 (10.66)
Bmi (kg/m^2):	24.65 (3.09)	24.04 (2.58)	24.33 (2.78)
Exercise session/wk	3	3	3
Exercise session length (min)	45-60	30-45	30-45
Session intensity (1=low,10=high)	4-6	4-6	4-6

2.2.2 Experimental Protocol

2.2.2.1 Electromyography Protocol

All testing was performed on the left leg. Surface electromyography (EMG) was collected from four muscles during each task using the Delsys Bagnoli System (Boston, MA); these muscles included Tibialis Anterior (TA), Soleus (SOL), Medial Gastrocnemius (MG), and Lateral Gastrocnemius (LG). Skin surface was shaved and cleaned with alcohol and surface electrodes were attached to the skin with a double sided interface. EMG was collected at 1000Hz. Signals were bandpass filtered (10-500Hz).

2.2.2.2 Passive Testing Protocol

Each subject participated in the passive testing protocol. Subjects were seated on the Biodex System 3 Pro Dynamometer (Biodex, Shirley, NY). They sat upright with their hip flexed 110°, their knee fully extended, and their testing leg supported by the ankle attachment attached to the dynamometer. The ankle was positioned initially at 90° flexion. Straps crossing the trunk, hips, thigh and foot held the subject in place and restricted movement to the ankle joint. The investigator passively moved the subject's ankle through the full range of motion (ROM), setting stopping points when the subject began to feel a slight stretch in the joint.

Each subject performed three resting trials to measure resting EMG, limb and attachment weight. The subject performed three passive trials during which the Biodex moved the subject's foot through the full ROM at 30°/s. Subjects were instructed to relax during this protocol, and EMG was obtained to ensure the musculature remained at rest. Trials were corrected for torque due to limb and attachment weight.

2.2.2.3 Balance Task

Three dimensional coordinates of subject motion was captured using a motion analysis system (VICON, Oxford Metrics, Oxford, England). Reflective markers were placed on the skin overlaying the lower extremity joints and on the segments between the joints on the left leg as well as on the trunk. Kinematics were captured at 200Hz.

This balance task was comprised of a single step moving into a single-leg balance task. The step was chosen as the means of producing a self-driven perturbation. Subjects moved from a double-leg stance into a single-leg stance following the step, simulating gait initiation. Additionally, the center of mass is relocated through the step, moving it outside of the base of support and thus initiating balance recovery. The single-leg stance provides greater challenge compared to double-leg stance, is more sensitive to differences in healthy adults, and is commonly used to assess balance (Erika Jonsson et al., 2007; E. Jonsson et al., 2004; Roemer & Raisbeck, 2015; Zumbrunn, MacWilliams, & Johnson, 2011). Subjects stood approximately 45cm from the center of the force plate (AMTI, Watertown, MA), allowing them to take one step forward and balance near the center of the force plate. Subjects were given a cadence and asked to step in place with the rhythm. Two cadences were used to change the velocity of the self-driven perturbation created during step initiation into the balancing task. Subjects were given several practice trials before data collection to familiarize themselves with the cadence and movement. When subjects felt comfortable with the cadence, they stepped onto the center of the force place and balanced on the stance leg for 10s or until failure. Successful balance task criteria included subjects stepping in pace with provided rhythm, ability to maintain upright stance, and maintaining stance for a minimum of 7.5s. Subjects completed 5 successful stepping tasks at each cadence for a total of 10 trials. Force plate data was captured at 1000Hz.

2.2.3 Passive Joint Torque Model

2.2.3.1 Model Explanation

In order to estimate passive torque arising from the biarticular and uniarticular musculature of the ankle, a model developed by Hoang et al. was used (2005). All MTU's were assumed to have exponential torque-angle relationships, including the Gastrocnemius. The following equation was used to model passive ankle joint torque (τ_{ankle}),

Equation 2.1

$$
\begin{aligned}
\tau_{ankle}\{\theta_a, l_g\} = \; & \left(a_p * e^{k_p(\theta_a - \theta_P)} - a_p\right) && \theta_a > \theta_P \\
& + \left(a_d * e^{k_d(\theta_D - \theta_a)} - a_d\right) && \theta_a < \theta_D \\
& + m_g * \left(a_g * e^{k_g\left(l_g - l_G\right)} - a_g\right) && l_g > l_G
\end{aligned}
$$

where θ_a is instantaneous ankle angle; l_g is instantaneous Gastrocnemius length; $a_p, k_p, a_d, k_d, a_g, k_g$ are stiffness constants for structures crossing the plantar, dorsal aspects of the ankle, respectively, as well as the gastrocnemius muscle; θ_P, θ_D are the ankle angles at which the respective uniarticular structures go slack; m_g refers to the gastrocnemius moment arm at the ankle; and l_G is the gastrocnemius slack length.

Gastrocnemius length and moment arm were calculated using the formulation put forward by Grieve et al. (1978),

Equation 2.2

$$l_g = l_{ref} + (A_0 + A_1\theta_a + A_2\theta_a^2 + K_0 + K_1\theta_k + K_2\theta_k^2) * l_s/100$$

Equation 2.3

$$m_g = \frac{1.8}{\pi} * l_s * (A_1 + (2A_2\theta_a))$$

where l_{ref} refers to the distance between the lateral epicondyle of the femur and lateral malleolus of the fibula with the knee and ankle at 90°; $A_0, A_1, A_2, K_0, K_1, K_2$ are coefficients found in Table 2.2; θ_k is instantaneous knee angle; and l_s is shank length.

Table 2.2: Coefficients for gastrocnemius length and gastrocnemius moment arm from Grieve et al. (1978)

A_0	A_1	A_2	K_0	K_1	K_2
-22.18468	0.30141	-0.00061	6.46251	-0.07987	0.00011

One trial was chosen for each subject as the "Model Identification" trial. Joint angles from this trial were used as inputs into the model. The model coefficients were optimized using Matlab's *lsqcurvefit* function (The Mathworks Inc., Natick, MA).

2.2.3.2 Model Validation

Once the model was optimized for each subject, the model coefficients were used to predict joint torques for subject Model Validation trials. Root mean square error measured the difference between the measured and predicted passive joint torques.

Additionally, ranges used for model coefficient optimization were validated with the literature. Ranges for gastrocnemius slack length were compared with Delp et al (1990), Hasson & Caldwell (2012), and Hoang et al (2007). Some of these previous works grouped the medial and lateral gastrocnemius muscles together while others differentiated between them. Ranges for plantarflexor and dorsiflexor slack angle (θ_P, θ_D, respectively) were also validated with the literature (Hirata, Kanehisa, Miyamoto-Mikami, & Miyamoto, 2015; Koo, Guo, Cohen, & Parker, 2014).

2.2.4 Biomechanical Modeling and Data Analysis

Marker and force plate data were digitized and trimmed to start the trial when the ground reaction force vector equaled 90% of body mass multiplied by gravitational vector, indicating initiation of weight acceptance phase. Marker and force plate data were filtered using a zero-phase low-pass 4th order Butterworth filter with a cut off frequency of 10Hz. Two-dimensional ankle, knee and hip joint angles were first determined using a Matlab script adapted from Reinschmidt and van den Bogert (Reinschmidt, 1997). Full extension was represented at 0° and full flexion, 180°. After joint angles were computed, joint angular velocity was calculated. Two-dimensional inverse dynamics were calculated using an adapted Matlab script from van den Bogert and Koning (Bogert, 1996). Hip, knee and ankle joint moments were determined. Joint moments were multiplied by joint angular velocity to produce joint powers.

Subject trials were divided into “dynamic” and “static” phases, with the dynamic phase consisting of the first 4 seconds of the trial. The static phase was comprised of the remaining portion of the balance task. Because the subject gains stability during the dynamic phase, the dynamic phase will be the analyzed phase for this task.

Trial kinematics and kinetics for each subject within each speed condition were averaged together to create one representative trial for each speed, resulting in two trials per person. According to the passive model, moment arms were estimated for the plantarflexor muscles (including the gastrocnemius) and the dorsiflexor muscles using Grieve et al's method (1978). The optimized and validated parameters for the model were applied to the representative trial joint angles and the passive joint moment was determined for the balance task. Additionally, the model estimated passive moments produced by the gastrocnemius MTU, uniarticular plantarflexor MTU and uniarticular dorsiflexor MTU. The net and individual muscle group passive moments were multiplied by the ankle joint velocity, and the passive joint powers were calculated. The passive joint power was divided by the net ankle joint power to provide a percentage value representing passive joint power contribution to balance recovery. Finally, passive and net joint powers, passive joint power contribution percentage, and ankle angular velocity were averaged for the dynamic phase.

All data was tested to ensure normal distribution prior to running statistical tests. A one-way ANOVA was used to test for interactions between stepping speed with regards to average passive joint power contribution, average net ankle joint power, average passive ankle joint power, average individual MTU power and average ankle joint angular velocity. Statistical analyses were performed using JMP Pro 11 (JMP®, Version 11. SAS Institute Inc., Cary, NC, 1989-2007).

2.3 Results

The passive exponential model optimized and validated for each subject reproduced passive joint moments from the input knee and ankle angles. Basic statistics for the model coefficients are summarized in Table 2.3.

Table 2.3: Mean, Median and Quartile values of coefficients estimated during optimization

	a_p	k_p	θ_p	a_d	k_d	θ_d	a_g	k_g	l_g
Mean	7.51	0.04	79.35	1.88	-0.04	76.62	3.17	94.22	0.38
Median	9.47	0.03	77.81	1.08	0.02	73.04	10.00	96.86	0.38
1st Quartile	6.55	0.02	75.00	-1.36	-0.13	69.87	-9.99	83.70	0.34
3rd Quartile	10.00	0.09	84.50	6.24	0.06	77.65	11.00	107.3	0.41

The average root mean square error for Model Identification was 1.961 Nm (1.434) for ankle flexion-extension. The average root mean square error for Model Validation was 1.031 Nm (0.523).

There was no significant difference present between net average ankle joint powers for the fast and slow balance task during the dynamic phase (p=0.9037). Additionally, no significant difference was seen for average passive ankle joint power (p=0.9778). The resulting contribution percentage subsequently is not significantly different (p=0.8531). Results for these variables are shown in Figure 2.1.

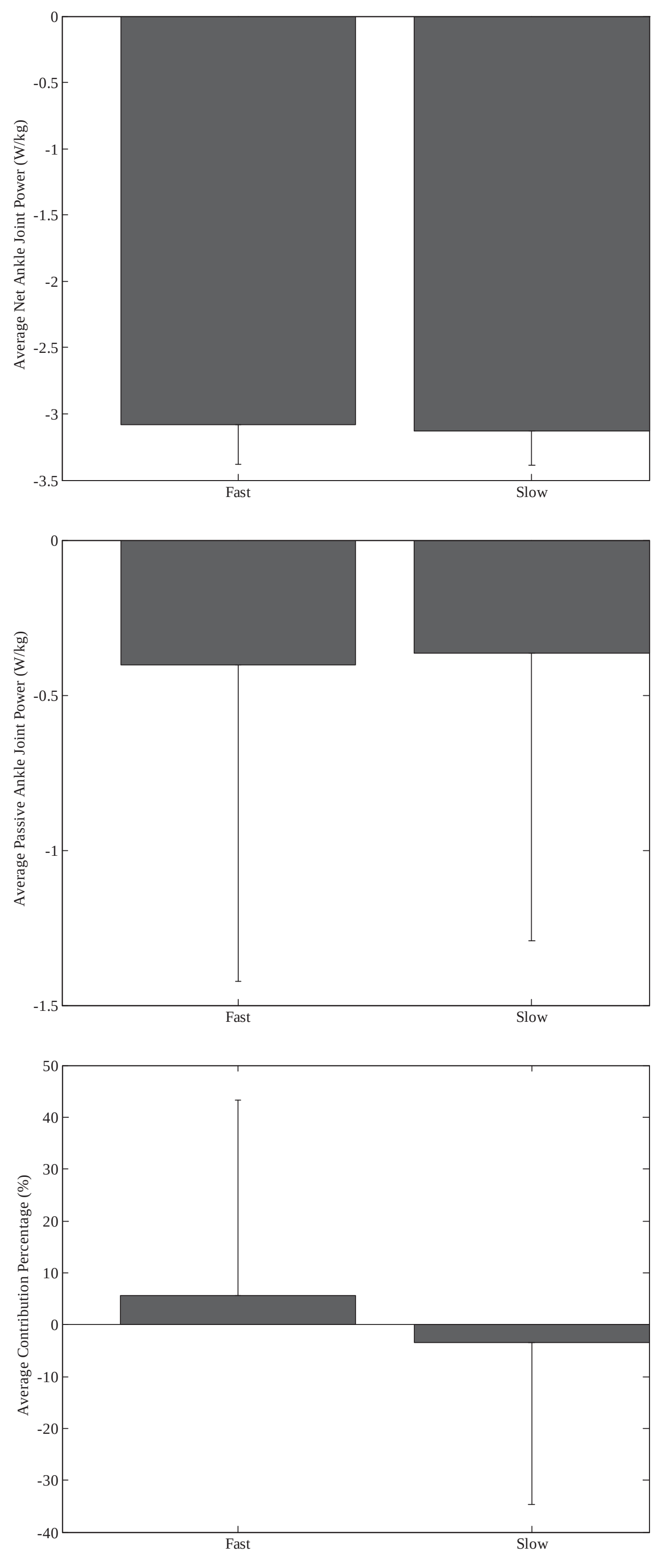
0
-0.5
-1
-1.5
-2
-2.5
-3
-3.5
Average Net Ankle Joint Power (W/kg)
Fast
Slow
0
-0.5
-1
-1.5
Average Passive Ankle Joint Power (W/kg)
Fast
Slow
50
40
30
20
10
0
-10
-20
-30
-40
Average Contribution Percentage (%)
Fast
Slow

Figure 2.1. Mean values and standard errors shown for fast and slow speed conditions. A) Mean net joint; B) Mean passive ankle joint power; C) Mean contribution percentage for passive to net ankle joint powers

The passive model produced estimations for biarticular and uniarticular muscles passive torque production at the ankle joint. There was no significant difference for passive power for the biarticular Gastrocnemius (p=0.9816), the uniarticular plantarflexors (p=0.9666) or the uniarticular dorsiflexors (p=0.9545) between the fast and slow conditions during the dynamic phase of the balance task. These results are shown in Figure 2.2.

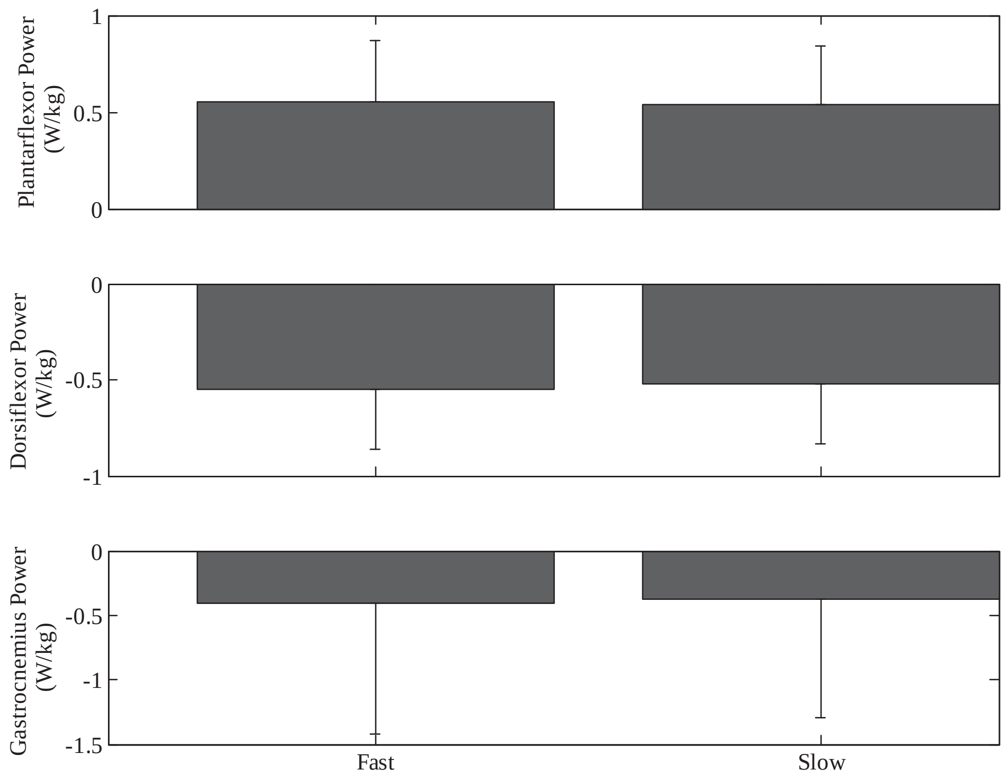

Figure 2.2: Mean passive powers for the uniarticular plantarflexors, uniarticular dorsiflexors and Gastrocnemius MTU's. Standard error is shown.

Average ankle joint velocity was not significantly different between the fast and slow conditions during the dynamic phase of balance recovery. Results are shown in Figure 2.3.

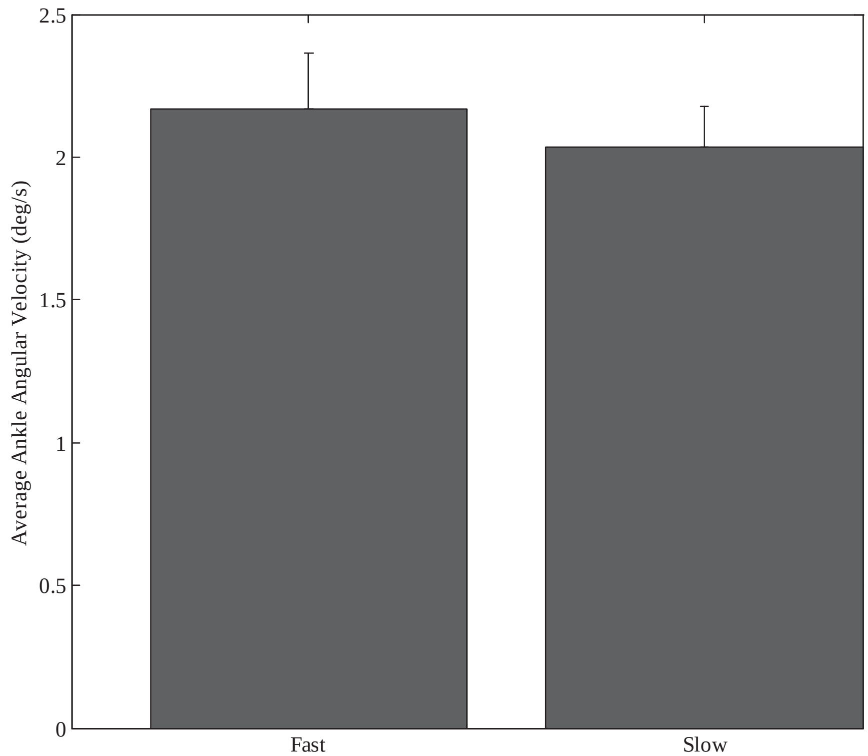

Figure 2.3: Mean and standard error for ankle angular velocity for the fast and slow speed conditions during the dynamic phase of balance recovery.

Average time series data for the fast and slow conditions is shown in Figure 2.4. Additionally, Figure 2.5 shows the average joint power curves for the net and passive ankle joint powers.

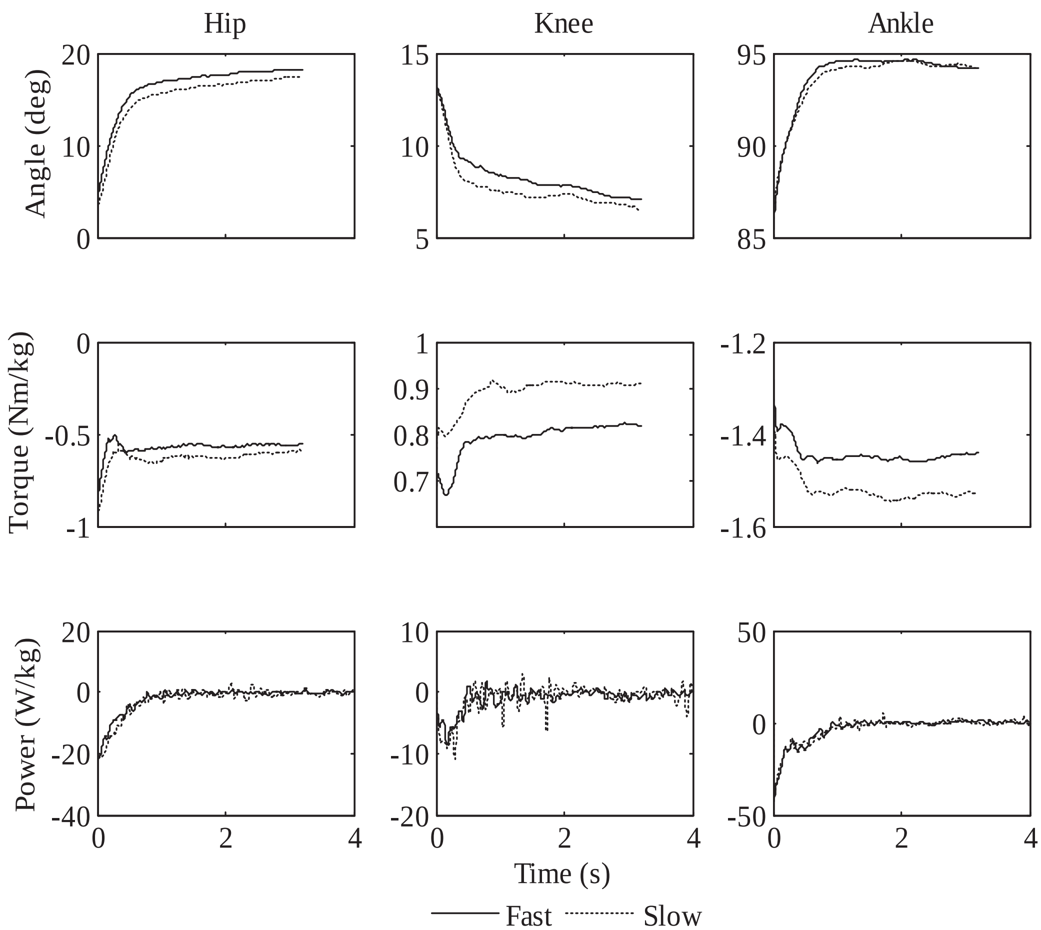

Figure 2.4: Time series data for ankle, knee and hip joints for fast and slow conditions over the length of the dynamic phase of balance recovery.

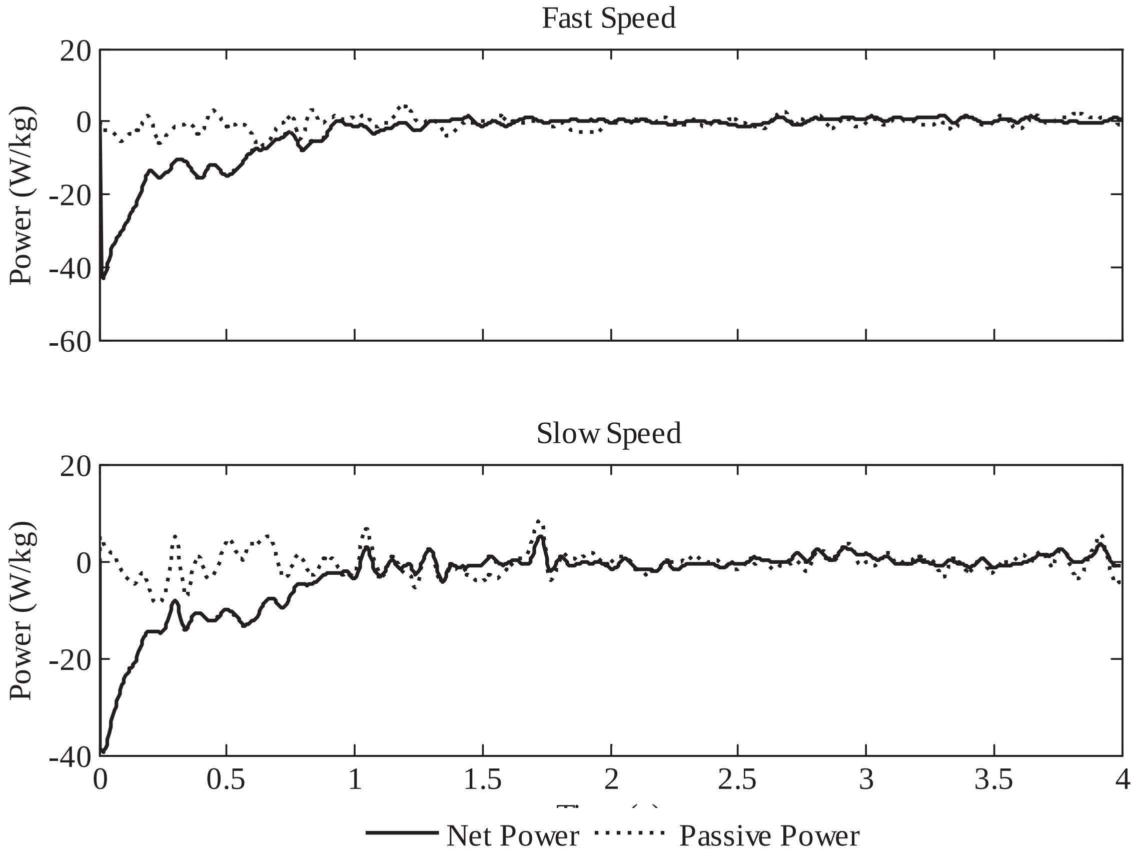

Figure 2.5: Time series data for net and passive ankle joint powers for the fast and slow speeds over the duration of the dynamic phase of balance recovery.

2.4 Discussion

In this study, we investigated the role of passive joint power during balance recovery in young adults. Specifically, we desired to explore how passive joint power contributed to net ankle joint torque following different speeds of initial self-induced perturbation.

Average net ankle joint power and average ankle joint velocities were not significantly different between the fast and slow stepping conditions. The net ankle joint powers and net ankle joint torque which in part comprises the net ankle joint power is similar to that of previous studies (DeVita & Hortobagyi, 2000; Hall, Woollacott, & Jensen, 1999). No change in net ankle joint power or average ankle angular velocity between perturbation speed conditions is surprising as the body must undergo greater acceleration initially to move the center of mass successfully to the center of the force plate. The lower extremity joints are largely responsible for producing adequate torque at the appropriate velocity to propel the body forward at the desired rate. These joints must also act to counteract a perturbation with lower extremity joint torques at the proper velocity to return the center of mass to within the base of support to prevent a fall. This result seems to indicate that the ankle joint is not the primary torque producer during the balance recovery in one or both of the speed conditions. These results do show, however, a dramatic reduction in net ankle joint power over the course of the dynamic phase of the balance recovery task, mirroring the rapid decrease in force variability seen in the data published by Jonsson et al (2004).

Passive ankle joint power, individual MTU passive powers, and percent contribution of passive joint power also were not significantly different between the fast and slow stepping conditions. Passive ankle joint torque has been shown in the literature to contribute to human balance (Morasso & Sanguineti, 2002; Muraoka et al., 2005; Winter et al., 1998). Additionally, Whittington et al (2008) found that passive ankle joint work, or passive ankle joint power integrated, increased with speed when analyzed in human gait. Passive joint powers arise from passive tissues surrounding and within the MTU. Increases in passive joint powers should be traced back to rises in passive powers from the MTU's. It was hypothesized that the passive component would be larger during the fast step compared with the slow step. Due to the high variability in the data, this hypothesis was not upheld.

For the present study, passive ankle joint torque, and thus passive ankle joint power, was estimated using a passive torque model (Hoang et al., 2005). One limitation associated with this model is the number of optimized model coefficients which makes it difficult to verify the uniqueness of the numerical result. While steps were taken to validate certain parameters from the literature, the range for the remaining parameters could cause the variability present in the data. This study could also be limited by the speeds chosen for the self-driven perturbation. The slow speed moved the subject at approximately 0.48m/s and the fast speed, 0.63m/s. The change in speed may not be great enough to elicit differences in net or passive ankle joint power or ankle angular velocity.

In conclusion, while it appears that passive ankle joint power does indeed contribute to net ankle joint power, it is unclear how perturbation speed affects this contribution. Future studies should consider greater changes in perturbation speed to better understand the role of speed with respect to passive ankle joint power.

Chapter 3

The Influence of Age and Perturbation Speed on Passive Ankle Joint Power Contribution

3.1 Introduction

As a person ages, their musculoskeletal system undergoes many changes. These muscular changes affect one's quality of life as daily movements become more difficult or unsafe. Muscle strength becomes diminished with age which is one of the most noticeable changes. Muscle fibers atrophy, reducing numbers of slow and fast twitch fibers (Doherty, Vandervoort, & Brown, 1993). These contractile properties changes cause reductions in muscle strength capacity and contraction velocity which in turn affect muscle power, ultimately affecting joint powers (Ochala, Lambertz, Pousson, Goubel, & Van Hoecke, 2004).

Muscle and tendon tissue also have passive elastic properties which change with age. Ochala et al (2004) observed that older men possessed increased musculotendinous stiffness, or the resistance of the muscle-tendon unit (MTU) to length change, in the plantarflexor muscles compared with a young male sample. Muscle stiffness plays a role in the passive moment produced in a joint which contributes significantly to total joint moments (Whittington et al., 2008). The passive moment is augmented by adjacent joint position due to biarticular muscles (Muraoka et al., 2005; Riener & Edrich, 1999; Salsich & Mueller, 2000; Silder et al., 2007). Muscle stiffness measurements also increase with increasing force production and contraction velocity (Julian & Sollins, 1975; McNair, Hewson, Dombroski, & Stanley, 2002; Weiss, Hunter, & Kearney, 1988). Gajdosik et al (2005) highlighted that the increase of muscle stiffness with stretching velocity exhibits itself greater in older women compared with young women.

As MTU's contract at increasing contraction velocities, joint powers will increase. Additionally, passive power will be more evident under greater contraction velocities due to passive power originating in part from passive MTU properties, such as stiffness. While studies have identified that MTU properties change with age, to date no research has identified the interaction of passive MTU power with net joint power and age in regards to balance. It is hypothesized that the contribution of passive joint power to net joint power with increasing velocity will be greater in the elderly compared with their young counterparts during a normal movement, such as balance recovery.

3.2 Methods

3.2.1 Subjects

The experiment was performed on 41 subjects, males and females divided into two age groups (11 males and 10 females, ages 20-30 and 10 males and 10 females, ages 60-70). Subject characteristics are contained in Table 1. Subjects were free from diabetes as well as a history of neuromuscular disorders or lower extremity surgeries. Additionally, subjects completed a questionnaire regarding their physical fitness habits. This data was used to ensure a recreationally active sample in order to better compare the older adults to the young. Informed consent was gained from each subject using procedures approved by the Michigan Technological University Institutional Review Board.

Table 3.1: Average subject characteristics with standard deviation shown.

	Young Adults		Older Adults	
	Females	**Males**	**Females**	**Males**
Age:	23.8±3.5	22.9±3.5	63.0±3.2	64.7±2.8
Height (cm)	165.3±7.7	180.4±6.8	162.3±6.1	176.7±7.3
Weight (kg)	67.34±9.33	78.19±9.34	68.25±8.57	82.08±12.2
BMI (kg/m^2)	24.65±3.09	24.04±2.58	25.93±3.13	26.18±2.86
Sessions/ Wk	3	3	3	3
Session Length (min)	45-60	30-45	30-45	30-45
Intensity (1=low,10=high)	4-6	4-6	4-6	4-6

3.2.2 Experimental Protocol

All subjects began the testing session with height and weight measurements. All testing for this study was performed on the left leg.

3.2.2.1 Electromyography Protocol

Surface electromyography (EMG) was collected from four muscles during each task using the Delsys Bagnoli System (Boston, MA); the four muscles included the Tibialis Anterior (TA), Soleus (SOL), Medial Gastrocnemius (MG), and Lateral Gastrocnemius (LG). Skin surface was shaved and cleaned with alcohol and surface electrodes were attached to the skin with a double sided interface. EMG was collected at 1000Hz. Signals were bandpass filtered (10-500Hz).

3.2.2.2 Passive Testing Protocol

Each subject participated in the passive testing protocol. The subject was seated on the Biodex System 3 Pro Dynamometer (Biodex, Shirley, NY). They sat upright with their hip flexed 110°, their knee fully extended, and their left leg supported by the ankle attachment attached to the dynamometer. The ankle was positioned initially at 90° flexion. Straps crossing the trunk, hips, thigh and foot held the subject in place and restricted movement to the ankle joint. The investigator passively moved the subject's ankle through the full range of motion (ROM), setting stopping points when the subject began to feel a slight stretch in the joint.

Each subject performed three resting trials to measure resting EMG, limb weight, and attachment weight. The subject performed three passive trials during which the Biodex moved the subject's foot through the full ROM at 30°/s. Subjects were instructed to relax during this protocol, and EMG was obtained to ensure the musculature remained at rest.

Maximal voluntary isometric contractions (MVC) were measured for the plantar- and dorsiflexor muscles as well. The ankle angle was repositioned to 90^0 and locked at that angle. Subjects performed three MVC's for each muscle group, alternating groups. They received 1 minute of rest between trials. EMG was collected during these measurements to provide a standard to mark the resting trials as truly at rest. All trials were corrected for torque due to limb and attachment weight.

3.2.2.3 Balance Task

Three dimensional coordinates of subject body motion was captured using a motion analysis system (VICON, Oxford Metrics, Oxford, England). Reflective markers were placed on the skin overlaying the lower extremity joints as well as on the segments between and the trunk. Body kinematics were captured at 200Hz. Marker trajectories were filtered with a zero-phase 4^{th} order Butterworth filter with a cutoff frequency of 10Hz.

Subjects stood approximately 45cm back from the center of the force plate (AMTI, Watertown, MA). Subjects were given a cadence and asked to step in place with the beat of the rhythm. Two cadences were used to change the velocity of the self-driven perturbation during step initiation. When subjects felt comfortable with the cadence, they stepped onto the center of the force place and held their balance on the left leg for 10 seconds or until failure. Subjects completed 5 successful stepping tasks at each cadence for a total of 10 trials. Successful trials were defined as trials that the stepping cadence was maintained, lasting a length of 7.5s or greater, and were the subject remained upright. Force plate data was captured at 1000Hz and was filtered with a zero-phase 4th order Butterworth filter with a cutoff frequency of 10Hz.

3.2.3 Passive Model Explanation

In order to estimate passive torque arising from the biarticular and uniarticular musculature of the ankle, a model developed by Hoang et al. was used (2005). All MTU's were assumed to have exponential torque-angle relationships, including the gastrocnemius. The following equation was used to model passive ankle joint torque (τ_{ankle}),

Equation 3.1

$$
\begin{aligned}
\tau_{ankle}\{\theta_a, l_g\} = \; & \left(a_p * e^{k_p(\theta_a - \theta_P)} - a_p\right) && \theta_a > \theta_P \\
& +\left(a_d * e^{k_d(\theta_D - \theta_a)} - a_d\right) && \theta_a < \theta_D \\
& +m_g * \left(a_g * e^{k_g\left(l_g - l_G\right)} - a_g\right) && l_g > l_G
\end{aligned}
$$

where θ_a is instantaneous ankle angle; l_g is instantaneous gastrocnemius length; $a_p, k_p, a_d, k_d, a_g, k_g$ are stiffness constants for structures crossing the plantar, dorsal aspects of the ankle, respectively, as well as the gastrocnemius muscle; θ_P, θ_D are the ankle angles at which the respective uniarticular structures go slack; m_g refers to the gastrocnemius moment arm at the ankle; and l_G is the gastrocnemius slack length.

Gastrocnemius length and moment arm were calculated using the formulation put forward by Grieve et al. (1978),

Equation 3.2

$$l_g = l_{ref} + (A_0 + A_1\theta_a + A_2\theta_a^2 + K_0 + K_1\theta_k + K_2\theta_k^2) * l_s/100$$

Equation 3.3

$$m_g = \frac{1.8}{\pi} * l_s * (A_1 + (2A_2\theta_a))$$

where l_{ref} refers to the distance between the lateral epicondyle of the femur and lateral malleolus of the fibula with the knee and ankle at 90°; $A_0, A_1, A_2, K_0, K_1, K_2$ are coefficients found in Table 2; θ_k is instantaneous knee angle; and l_s is shank length.

Table 3.2: Coefficients for gastrocnemius length and gastrocnemius moment arm from Grieve et al. (1978)

A_0	A_1	A_2	K_0	K_1	K_2
-22.18468	0.30141	-0.00061	6.46251	-0.07987	0.00011

One trial from the subjects' passive Biodex trials was used to identify model coefficients through optimization. The model coefficients were optimized using Matlab's *lsqcurvefit* function (The Mathworks Inc., Natick, MA). The model was validated using other passive Biodex trials for the subject and minimizing the root mean square error between the measured and predicted ankle joint torque.

3.2.4 Data Analysis

Marker and force plate data were digitized and exported to Matlab (The Mathworks Inc., Natick, MA). Trials were trimmed to start the trial when the ground reaction force vector equaled 90% of body mass multiplied by gravitational vector in order to capture the ending portion of the weight acceptance phase. Marker and force plate data were filtered using a zero-phase low-pass 4th order Butterworth filter with a cut off frequency of 10Hz. Two-dimensional ankle, knee and hip joint angles were first determined using a Matlab script adapted from Reinschmidt and van den Bogert (Reinschmidt, 1997). Full extension was represented at 0° and full flexion, 180° for all joints. After joint angles were computed, joint angular velocity was calculated. Two-dimensional inverse dynamics were calculated using an adapted Matlab script from van den Bogert and Koning (Bogert, 1996). Hip, knee and ankle joint moments were determined. Joint moments were multiplied by joint angular velocity to produce joint powers.

Subject trials were divided into “dynamic” and “static” phases, with the dynamic phase consisting of the first 4 seconds of the trial (Dingenen et al., 2013). The static phase was comprised of the remaining portion of the balance task. Because the subject gains stability during the dynamic phase, the dynamic phase will be the analyzed phase for this task.

Trial kinematics and kinetics for each subject within each speed condition were averaged together to create one representative trial for each speed, resulting in two trials per person. According to the passive model, moment arms were estimated for the plantarflexor muscles (including the gastrocnemius) and the dorsiflexor muscles using Grieve et al’s method (1978). The optimized and validated parameters for the model were applied to the representative trial joint angles and the passive joint moment was determined for the balance task. Additionally, the model estimated passive moments produced by the gastrocnemius MTU, uniarticular plantarflexor MTU and uniarticular dorsiflexor MTU. The net and individual muscle group passive moments were multiplied by the ankle joint velocity, and the passive joint powers were calculated. The passive joint power was divided by the net ankle joint power to provide a percentage value representing passive joint power contribution to balance recovery. Finally, passive and net joint powers, passive joint power contribution percentage, and ankle angular velocity were averaged for the dynamic phase.

All variables were tested to ensure normal distribution. A two-way ANOVA was used to determine interactions present between the age groups and speed conditions with regards to percent contribution, net ankle joint work, passive ankle joint work, individual MTU work, and ankle angular velocity. Statistical analyses were performed using JMP Pro 11 (JMP®, Version 11. SAS Institute Inc., Cary, NC, 1989-2007).

3.3 Results

The passive exponential model was optimized for each subject and reproduced the passive joint torques measured on the Biodex using knee and ankle angles as inputs. The average root mean square error was 1.883 Nm (±1.937) for ankle flexion-extension. Once the model was optimized, it was validated for each subject. The average root mean square error for model validation was 0.979 Nm (±0.549). Basic statistics for the optimized model coefficients are shown in Table 3.3.

Table 3.3: Mean, Median, 1st and 3rd Quartiles for optimized Passive Model parameters

	a_p	k_p	θ_p	a_d	k_d	θ_d	a_g	k_g	l_g
Mean	6.33	0.04	81.70	2.88	-0.05	77.95	2.85	90.16	0.38
Median	8.64	0.03	81.23	2.48	-0.01	74.80	10.00	94.59	0.37
1st Quartile	1.43	0.02	75.00	-0.68	-0.10	68.89	-10.0	75.13	0.34
3rd Quartile	10.00	0.08	85.66	7.40	0.06	86.03	15.00	109.5	0.41

Net ankle joint power was not significantly affected by the speed condition (p=0.4573). However, net ankle joint power was significantly greater in older adults compared with the young (p=0.0369). Passive ankle joint power had no interactions for speed or age (p=0.4158, p=0.2219, respectively). Passive ankle joint power percent contribution was also not significantly affected by speed or age (p=0.4434, p=0.1387, respectively). These results are shown in Figure 3.1.

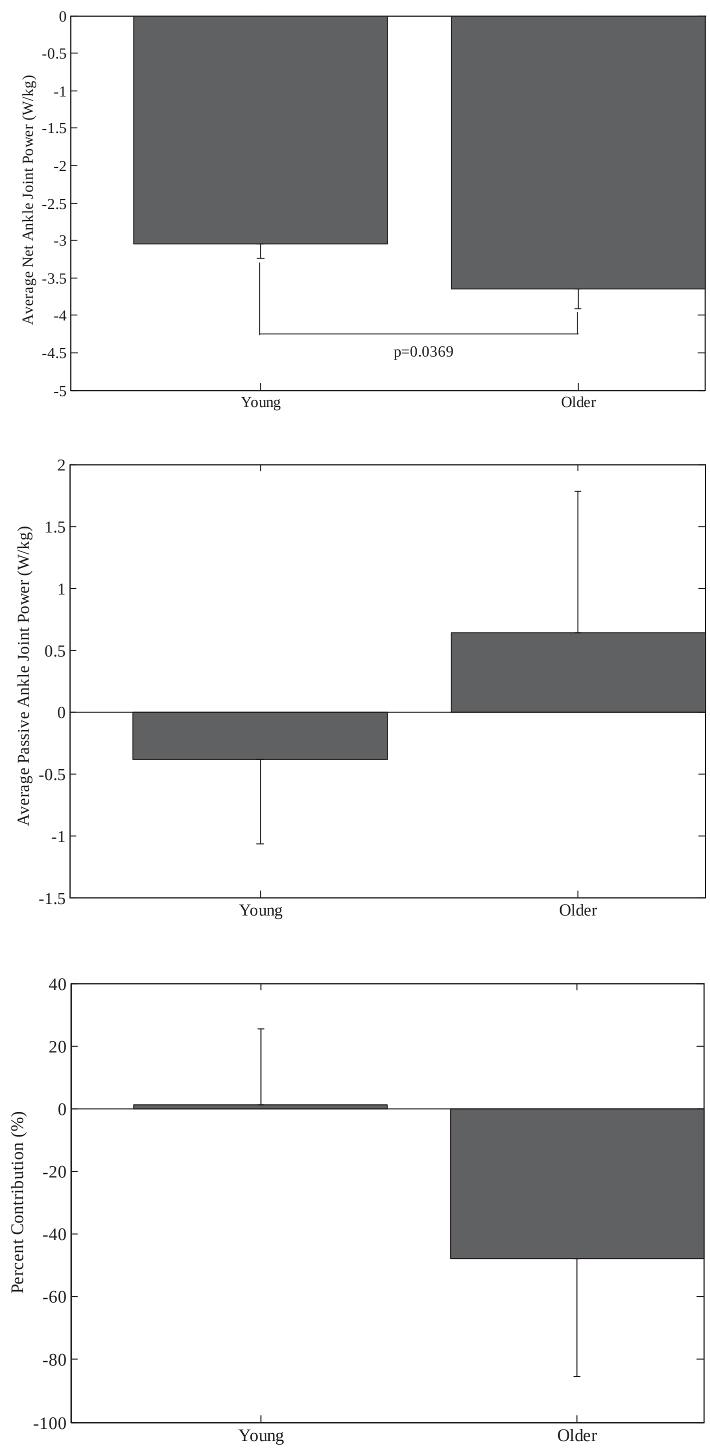

Figure 3.1: Mean values and standard error for young and older adults. A) Net ankle joint power; B) Passive ankle joint power; C) Passive ankle joint power to net ankle joint power percent contribution

The biarticular gastrocnemius muscle had no interactions present for speed or age ($p=0.4042$, $p=0.3550$, respectively). The uniarticular plantarflexors produced significantly more passive power in young adults compared with older adults ($p=0.0360$). The uniarticular dorsiflexors also produced greater passive power in young adults compared with older adults ($p=0.0236$). There were no interactions with respect to speed for the uniarticular muscles. These results are shown in Figure 3.2.

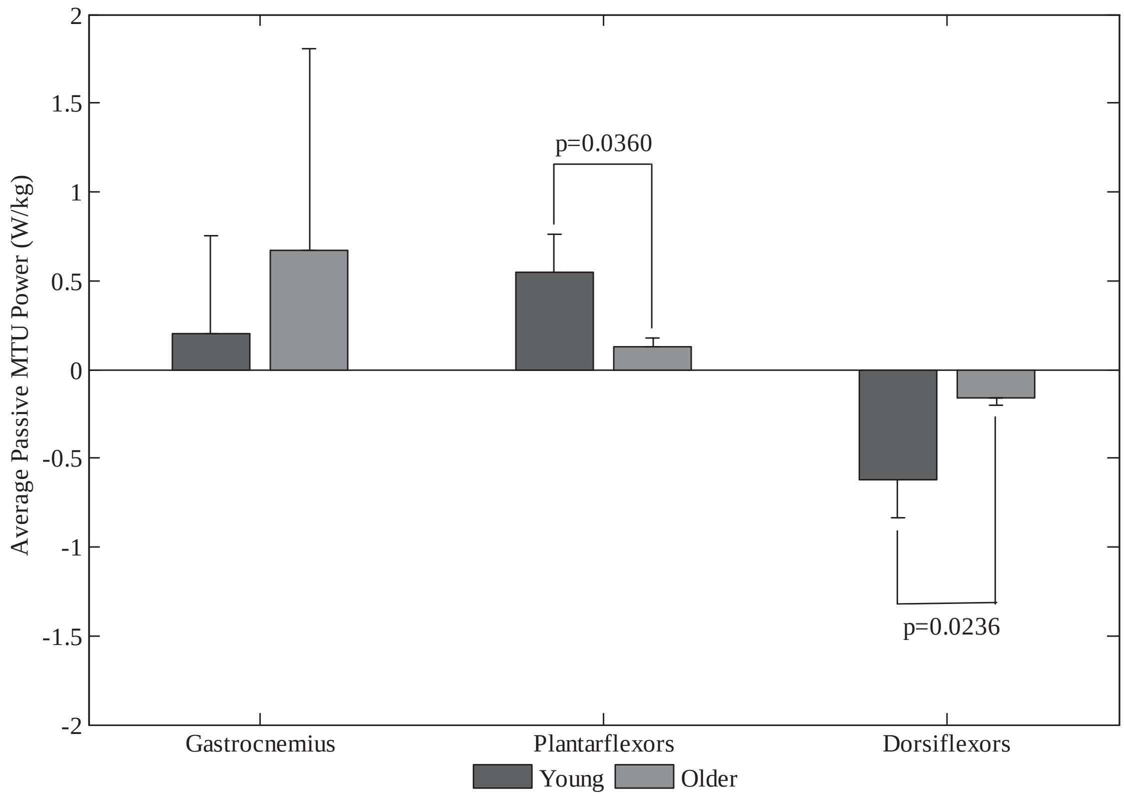

Figure 3.2: Mean passive MTU powers for young and older adults. Standard error bars are shown.

Ankle angular velocity had no interactions for speed or age ($p=0.1907$, $p=0.4755$, respectively). Ankle angular velocity is shown for both age and speed conditions in Figure 3.3.

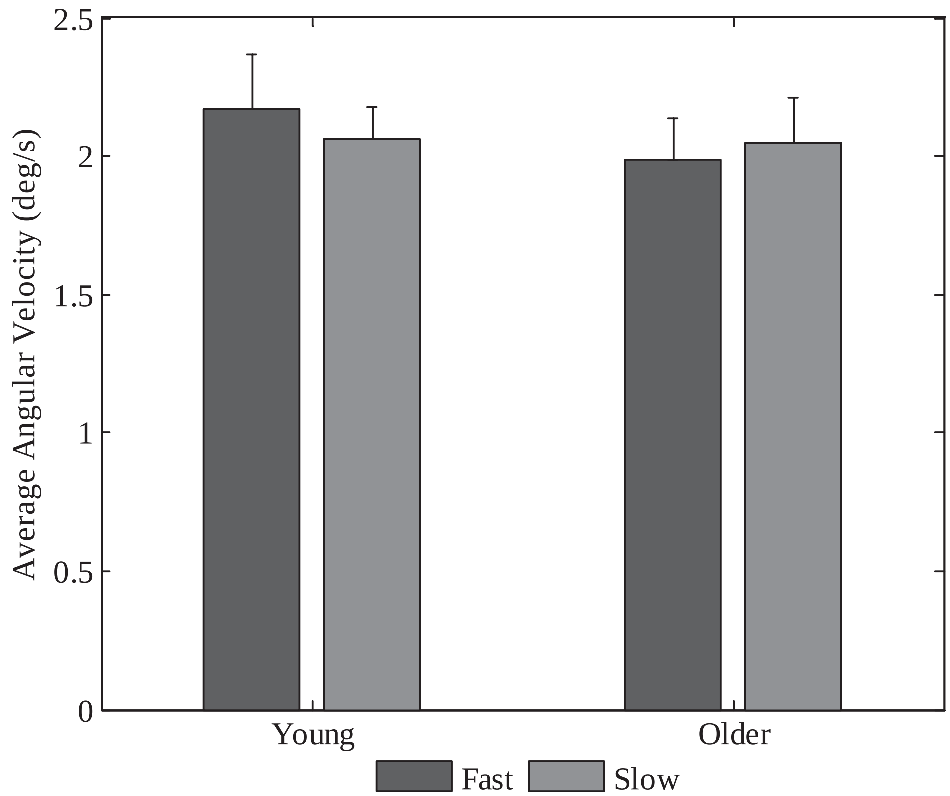

Figure 3.3: Mean ankle angular velocity with standard error shown for both age and speed conditions during the dynamic phase of balance recovery.

Time series plots are included for the hip, knee and ankle joint to depict angles, torques and joint powers for both age and speed conditions. These are shown in Figure 3.4. Additionally, passive and net ankle joint power curves for the speed and age conditions are shown in Figure 3.5.

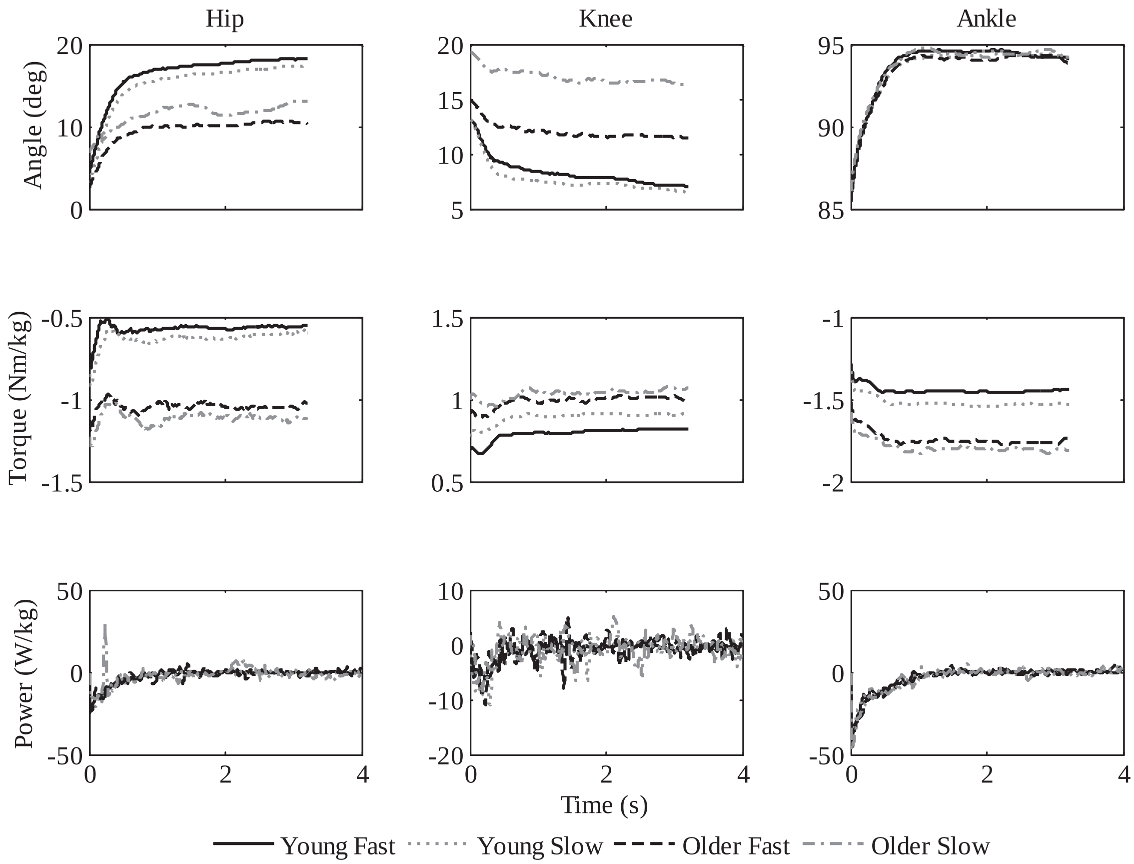

Figure 3.4: Averaged time series plots for the hip, knee and ankle joints for both age and speed conditions.

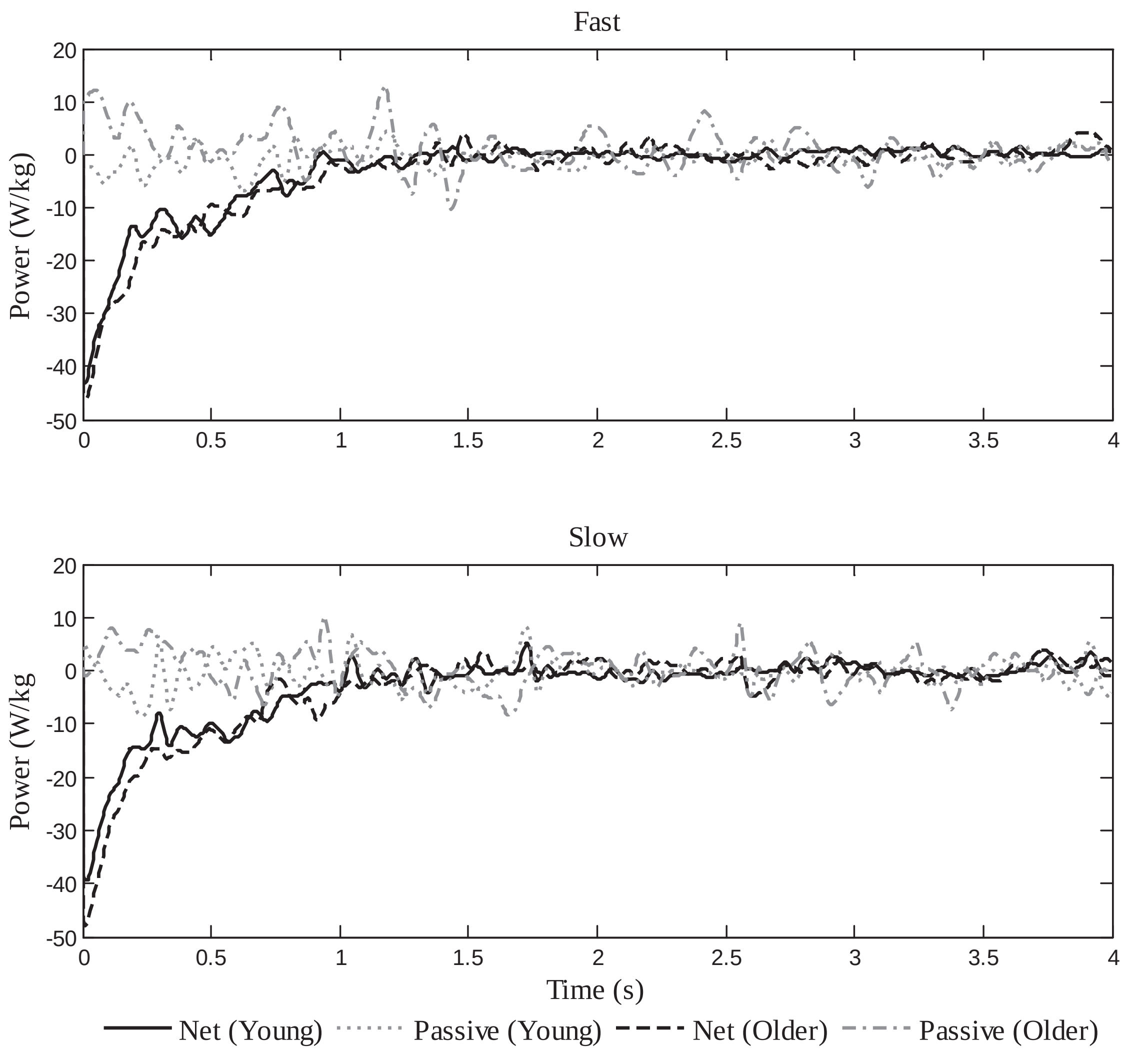

Figure 3.5: Average time series plots of young and older adult net and passive ankle joint powers for fast and slow speed conditions.

3.4 Discussion

In this study, we investigated the role of passive ankle joint power and its contribution to net ankle joint power during a balance task. Specifically, we manipulated perturbation speed of perturbation to determine if contribution of passive ankle joint power to net ankle joint power was different with young adults compared with older adults.

While average net ankle joint power had no interactions present for perturbation speed, it was significantly increased in older adults compared with young adults. Joint powers are comprised of joint torques and angular velocity. In the current study, there was no difference in ankle angular velocity for the age or speed conditions. The changes in the speed of perturbation may not have been high enough to elicit changes in ankle angular velocity. Additionally, the older adults chosen for this study were recreationally active, and this may diminish any effects of age on ankle angular velocity. Therefore, tt appears that this change in net ankle joint power stems from a change in net ankle joint torque. Consider this, the finding seems to conflict with Hall et al (1999) who investigated balance recovery and found no difference in maximal ankle joint torque between age groups regardless of perturbation speed or direction. Madigan et al (2006) found that older males produced greater ankle joint torque during a phase in a forward leaning balance task compared with younger men. They also found lower knee joint torques and higher hip joint torques in older men compared with younger men. Future research should investigate the relationship of ankle, knee and hip joint powers in relation to passive joint powers during balance recovery.

While "net" passive ankle joint power was unaffected by speed or age, passive power from the uniarticular dorsiflexor and plantarflexor muscles was affected by age. Uniarticular plantarflexor muscles, such as the Soleus, increased passive power in young adults compared with older adults. Uniarticular dorsiflexor muscles, such as the Tibialis Anterior, also had increased passive power in young adults compared with older adults. Simoneau et al (2005; 2009) found decreases in maximal plantarflexor torque acting as agonist and antagonist in older adults compared with young. They also found age-related decreases in dorsiflexor antagonist maximal torque (2009). While the current study did not assess maximal plantar- and dorsiflexor torque, the current results could be indicative of muscular changes that exhibit themselves as muscular weakness.

Percent contribution of passive ankle joint power to net ankle joint power was not significantly different for the speed or age condition, despite the differences in net ankle joint power with respect to age. Passive ankle joint power had high variability in the data, masking any potential differences. Some of this variability in could be due to the current study making no distinction for gender. Foure et al (2012) identified gender differences in ankle joint stiffness which would affect the passive joint torque and thus passive joint power.

Some limitations involved in this study included the inability to control stepping velocity. While subjects did indeed produce fast and slow steps into their balancing task, we could not control the exact rate that the steps were produced. This limitation could explain some of the variability in the data due to their being a range of fast and slow stepping velocities. The difference between the stepping velocities also may not have been great enough to elicit changes. Secondly, the passive torque model utilized in this study contained nine coefficients that needed to be tuned through optimization. While three of the nine coefficients (plantarflexor slack angle (Hirata et al., 2015), dorsiflexor slack angle (Koo et al., 2014), and gastrocnemius slack length (Delp et al., 1990; Hasson & Caldwell, 2012)) have typical ranges described in the literature, the six remaining coefficients were less concrete. This problem was attempted to be overcome through applying a model validation using other trials, but there is potential that the solution arrived at was not unique, adding unnecessary variability to the data.

In conclusion, it is apparent age affects net ankle joint power and individual MTU powers during the dynamic phase of balance recovery. Future investigations should explore the role of gender with respect to passive ankle joint power. Additionally, passive ankle joint powers should be investigated in relation to balance recovery strategies. Passive ankle joint work may explain why certain recovery strategies are utilized.

Chapter 4

The Influence of Passive Joint Power on Balance Recovery Strategy

4.1 Introduction

Human balance is essential to daily living. Balance can be defined as maintaining the center of mass within its base of support. When the center of mass moves outside of the base of support, it must be returned to within those limits, otherwise failure occurs. This can result in injury and loss of independence.

Human balance has been investigated and explained in a variety of ways. Horak and Nashner (1986) investigated the idea of balance recovery strategy. This is the movement pattern one utilizes to recover from a perturbation. They defined the ankle strategy as the more simplistic strategy which utilizes only the ankle joint to regain balance. This strategy is primarily seen in recovery situations that take place on a flat or wider surface and at a slower translation. The hip strategy uses the hip joint primarily to regain balance and is typically employed in perturbations of faster translation or on narrower support surfaces (Kuo, 1995; Nashner & McCollum, 1985).

While these recovery strategies have been demonstrated in the lab, often the testing situation involves specific constraints such as limiting knee movement and conditions using a tilting platform to observe these strategies. These testing environments can help us glean insight to these recovery strategies, but they fail to mimic real life situations. More commonly, a mixed strategy is utilized for balance recovery, triggering both the ankle and hip joints to respond to the perturbation. The mixed strategy represents a continuum between the ankle and hip strategies.

Clifford and Holder-Powell (2010) investigated balance recovery strategies in single leg balance and observed ankle and hip strategies being used for recovery. Single leg balance reduces the support surface width and increases the challenge of the task. Single leg stances may also more closely replicate falling or tripping scenarios.

Another factor affecting movement is passive joint power production (Silder et al., 2007; Whittington et al., 2008). Because of the composition of muscles and tendons, they produce passive forces as length changes. Increased passive joint powers in the ankle joint may be linked to a change balance recovery strategy, causing the hip joint to compensate for reduced active control of the ankle joint. It is hypothesized that greater contribution of passive ankle joint power to net ankle joint power will be correlated to a mixed strategy that uses more of the hip strategy than the ankle strategy.

4.2 Methods

4.2.1 Subjects

41 subjects were recruited from the community and campus of Michigan Technological University. Subjects provided informed, signed consent prior to participation according to the protocol approved by Michigan Technological University's Institutional Review Board.

Subjects were divided into two age groups: young adults (ages 20-30) and older adults (ages 60-70). All subjects were free from neuromuscular and musculoskeletal disorders, lower extremity surgeries and diabetes. Additionally, subjects completed a physical activity questionnaire to ensure that both groups were equally active. Subject characteristics are provided in Table 4.1.

Table 4.1: Average subject characteristics with standard deviation shown.

	Young Adults		Older Adults	
	Females	**Males**	**Females**	**Males**
Age:	23.8±3.5	22.9±3.5	63.0±3.2	64.7±2.8
Height (cm)	165.3±7.7	180.4±6.8	162.3±6.1	176.7±7.3
Weight (kg)	67.34±9.33	78.19±9.34	68.25±8.57	82.08±12.2
BMI (kg/m^2)	24.65±3.09	24.04±2.58	25.93±3.13	26.18±2.86
Sessions/ Wk	3	3	3	3
Session Length (min)	45-60	30-45	30-45	30-45
Intensity (1=low,10=high)	4-6	4-6	4-6	4-6

4.2.2 Experimental Protocol

Upon coming to the lab, subjects' height and body mass measurements were taken. All following testing outlined in this protocol was performed on the left leg.

4.2.2.1 Electromyography Protocol

Surface electromyography (EMG) was collected from the Tibialis Anterior (TA), Soleus (SOL), Medial Gastrocnemius (MG), and Lateral Gastrocnemius (LG) during the passive testing protocol using the Delsys Bagnoli System (Boston, MA). Muscles were first located and marked, and skin surface was shaved and cleaned with alcohol according to SENIAM guidelines. Surface electrodes were attached to the skin with a double sided interface. EMG was collected at 1000Hz. Signals were bandpass filtered (10-500Hz).

4.2.2.2 Passive Testing Protocol

For the passive testing protocol, the subject was seated upright on the Biodex System 3 Pro Dynamometer (Biodex, Shirley, NY). They sat with their hip flexed 110°, their knee fully extended, and their left leg supported by the ankle attachment attached to the dynamometer. The ankle was positioned initially at 90° flexion. Straps crossing the trunk, hips, thigh and foot held the subject in place and restricted movement to the ankle joint. The investigator passively moved the subject's ankle through the full range of motion (ROM), setting stopping points when the subject began to feel a slight stretch in the joint.

Three resting trials were obtained to measure resting EMG, limb and attachment weight. The subject performed three passive trials during which the Biodex moved the subject's foot through the full ROM at 30°/s. Subjects were instructed to relax during this protocol, and EMG was obtained to ensure the musculature remained at rest.

Maximal voluntary isometric contractions (MVC) were measured for the plantar- and dorsiflexor muscles as well. The ankle angle was repositioned and fixed at 90°. Subjects performed three MVC's for each muscle group, alternating groups. They received 1 minute of rest between trials. EMG was collected during these measurements to provide a standard to mark for the resting trials. All trials were corrected for torque due to limb and attachment weight.

4.2.2.3 Balance Task

Three dimensional coordinates of subject body motion was captured using a motion analysis system (VICON, Oxford Metrics, Oxford, England). Reflective markers were placed on the skin overlaying the lower extremity joints, the segments between, and the trunk. Body kinematics were captured at 200Hz. Marker trajectories were filtered with a zero-phase 4th order Butterworth filter with a cutoff frequency of 10Hz.

Subjects stood approximately 45cm back from the center of the force plate (AMTI, Watertown, MA). Subjects were given a cadence and asked to step in place with the rhythm. Two cadences were used to change the velocity of the self-driven perturbation during step initiation. When subjects felt comfortable with the cadence, they stepped onto the center of the force place and held their balance on the left leg for 10 seconds or until failure. Time was provided for subjects to familiarize themselves with this movement. Subjects completed 5 successful stepping tasks at each cadence for a total of 10 trials. Force plate data was captured at 1000Hz and was filtered with a zero-phase 4th order Butterworth filter with a cutoff frequency of 10Hz.

4.2.3 Passive Model Explanation

In order to estimate passive torque arising from the biarticular and uniarticular musculature of the ankle, a model developed by Hoang et al. was used (2005). All MTU's were assumed to have exponential torque-angle relationships, including the gastrocnemius. The following equation was used to model passive ankle joint torque ($\boldsymbol{\tau_{ankle}}$),

Equation 4.1

$$\begin{aligned}\tau_{ankle}\{\theta_a, l_g\} = \; & \left(a_p * e^{k_p(\theta_a - \theta_P)} - a_p\right) && \theta_a > \theta_P \\ & +\left(a_d * e^{k_d(\theta_D - \theta_a)} - a_d\right) && \theta_a < \theta_D \\ & +m_g * \left(a_g * e^{k_g(l_g - l_G)} - a_g\right) && l_g > l_G\end{aligned}$$

where θ_a is instantaneous ankle angle; l_g is instantaneous gastrocnemius length; $a_p, k_p, a_d, k_d, a_g, k_g$ are stiffness constants for structures crossing the plantar, dorsal aspects of the ankle, respectively, as well as the gastrocnemius muscle; θ_P, θ_D are the ankle angles at which the respective uniarticular structures go slack; m_g refers to the gastrocnemius moment arm at the ankle; and l_G is the gastrocnemius slack length.

Gastrocnemius length and moment arm were calculated using the formulation put forward by Grieve et al. (1978),

Equation 4.2

$$l_g = l_{ref} + (A_0 + A_1\theta_a + A_2\theta_a^2 + K_0 + K_1\theta_k + K_2\theta_k^2) * l_s/100$$

Equation 4.3

$$m_g = \frac{1.8}{\pi} * l_s * (A_1 + (2A_2\theta_a))$$

where l_{ref} refers to the distance between the lateral epicondyle of the femur and lateral malleolus of the fibula with the knee and ankle at 90°; $A_0, A_1, A_2, K_0, K_1, K_2$ are coefficients found in Table 4.2; θ_k is instantaneous knee angle; and l_s is shank length.

Table 4.2: Coefficients for gastrocnemius length and gastrocnemius moment arm from Grieve et al. (1978)

A_0	A_1	A_2	K_0	K_1	K_2
-22.18468	0.30141	-0.00061	6.46251	-0.07987	0.00011

One trial from each subject's passive Biodex trials was used for model coefficient optimization, minimizing the root mean square error between the trial's measured torque and the model's predicted torque, using Matlab's *lsqcurvefit* function (The Mathworks Inc., Natick, MA). The ranges set for θ_P, θ_D, and l_G were determined using values found in the literature (Delp et al., 1990; Hasson & Caldwell, 2012; Hirata et al., 2015; Koo et al., 2014). The model was validated by applying the optimized coefficients to a new passive trial data set, and root mean square was minimized..

4.2.4 Data Analysis

Marker and force plate data were digitized, exported, and filtered using a zero-phase low-pass 4th order Butterworth filter with a cut off frequency of 10Hz. Trials were cut at the start of the trial to the point when the ground reaction force vector equaled 90% of the subject's body mass multiplied by the gravitational vector. A Matlab script was adapted to determine two-dimensional ankle, knee and hip joint angles (Reinschmidt, 1997). Full extension was represented at 0° and full flexion, 180° for the hip, knee and ankle. Joint angular velocity was calculated from the first derivative of joint angle. Two-dimensional inverse dynamics were calculated using an adapted Matlab script from van den Bogert and Koning (Bogert, 1996). Hip, knee, and ankle joint moments were determined. Joint torques were normalized to the subject's body mass and then multiplied by joint angular velocity to produce joint powers.

Passive joint torques were estimated for each trial using joint angles as the inputs along with the optimized parameters per subject. A representative trial was created for each speed condition for each subject by averaging the trial data together within each speed. This resulted in two representative trials for all subjects, one for the fast condition and one for the slow. Net and passive joint torques were first normalized by subject body mass and then net and passive joint powers were calculated for each subject trial by multiplying the net and passive joint moments by the ankle joint velocities.

The balance trial was segmented into dynamic and static phases. The dynamic phase was comprised of the first four seconds of the trial (Dingenen et al., 2013). The static phase was comprised of the remaining portion of the balance trial. Because the dynamic phase contains the portion of balance related to gaining stabilization, this phase was analyzed for the current study (Dingenen et al., 2013; E. Jonsson et al., 2004; Parreira et al., 2013; Roemer & Raisbeck, 2015)**.**

The net and passive joint powers were averaged over the length of the dynamic phase. Passive joint power was divided by net joint power to produce the contribution percentage, relating the two variables.

Ankle and hip joint torques were assessed to determine if a pure ankle or hip strategy was used during the dynamic phase of balance recovery. If both joints were found to contribute to recovery, a mixed strategy was found, and a *strategy index* (SI), relating hip joint torque to ankle joint torque, was used to determine the point of the continuum between ankle and hip strategy (see Figure 4.1).

Figure 4.1: Schematic of the pure ankle and hip strategies and the strategy index.

All data was tested for normality. A two-way ANOVA was performed to assess interactions present between age and speed with respect to the strategy index. A linear regression was performed to relate passive joint power percent contribution to the strategy index in order to determine if percentage can be used to predict balance recovery strategy.

4.3 Results

The passive torque model was optimized and validated for each subject to reproduce measured joint torques. The average root mean square error for Model Identification was 1.883NM (±1.937). The average root mean square error for Model Validation was 0.979NM (±0.549). Mean, median and quartile data for the optimized coefficients is shown in Table 4.3.

Table 4.3: Mean, Median, 1st and 3rd Quartiles for optimized Passive Model parameters

	a_p	k_p	θ_p	a_d	k_d	θ_d	a_g	k_g	l_g
Mean	1.899	-2.95	37.95	-2.54	-0.45	102.7	4.753	94.00	0.369
Median	2.794	-0.15	44.94	-0.88	-0.11	101.5	2.422	86.95	0.369
1st Quartile	-4.61	-1.89	20.77	-5.57	-0.26	87.30	0.085	72.51	0.355
3rd Quartile	8.388	0.004	53.78	3.669	-0.01	113.8	7.287	113.8	0.387

Hip, knee and ankle angle, torque and power data for both speed and age conditions for the length of the dynamic phase of the balance task is shown in Figure 4.2.

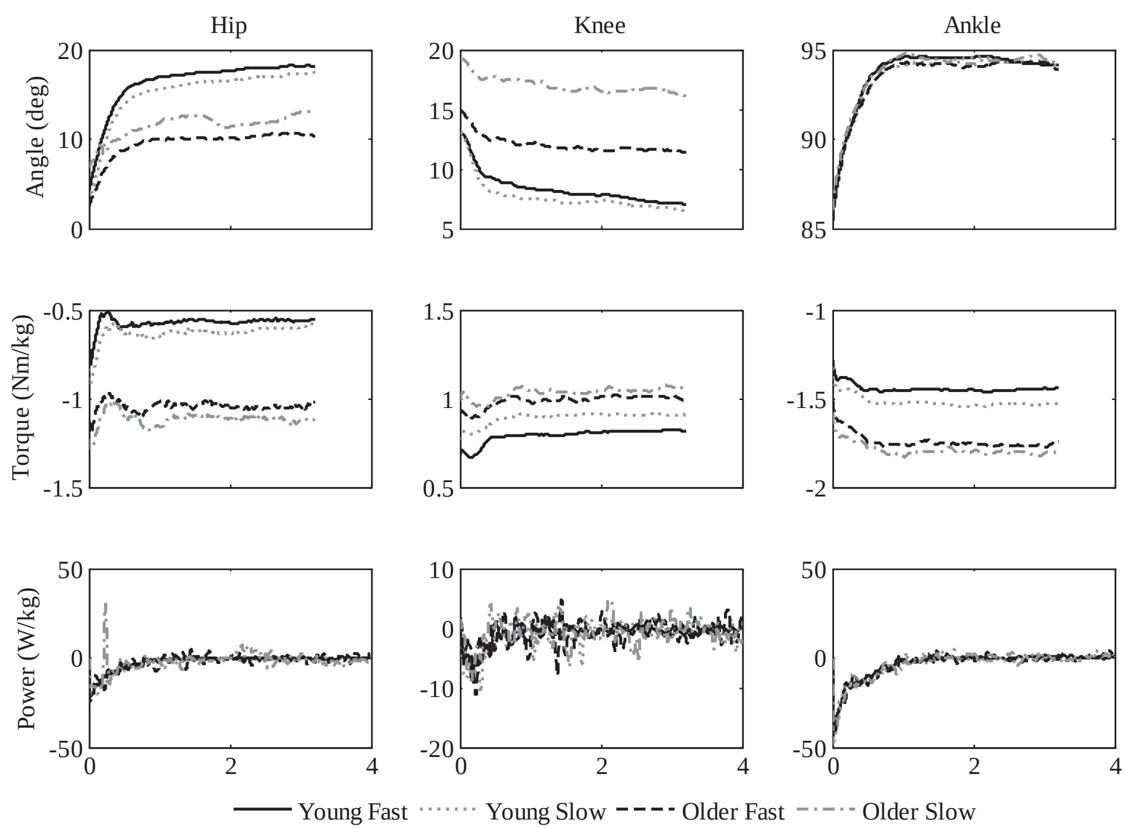

Figure 4.2: Ankle, knee and hip joint angles, torques and powers are shown for the duration of the dynamic phase of balance task; Plots depict age and speed

All subjects utilized a mixed strategy for balance recovery during the dynamic phase. There was an interaction present for strategy index with respect to age. While both young and older adults utilized more ankle joint torque in their strategy index, older adults had a significantly higher strategy index compared with young adults ($p<0.0001$). Results are shown in Figure 4.3.

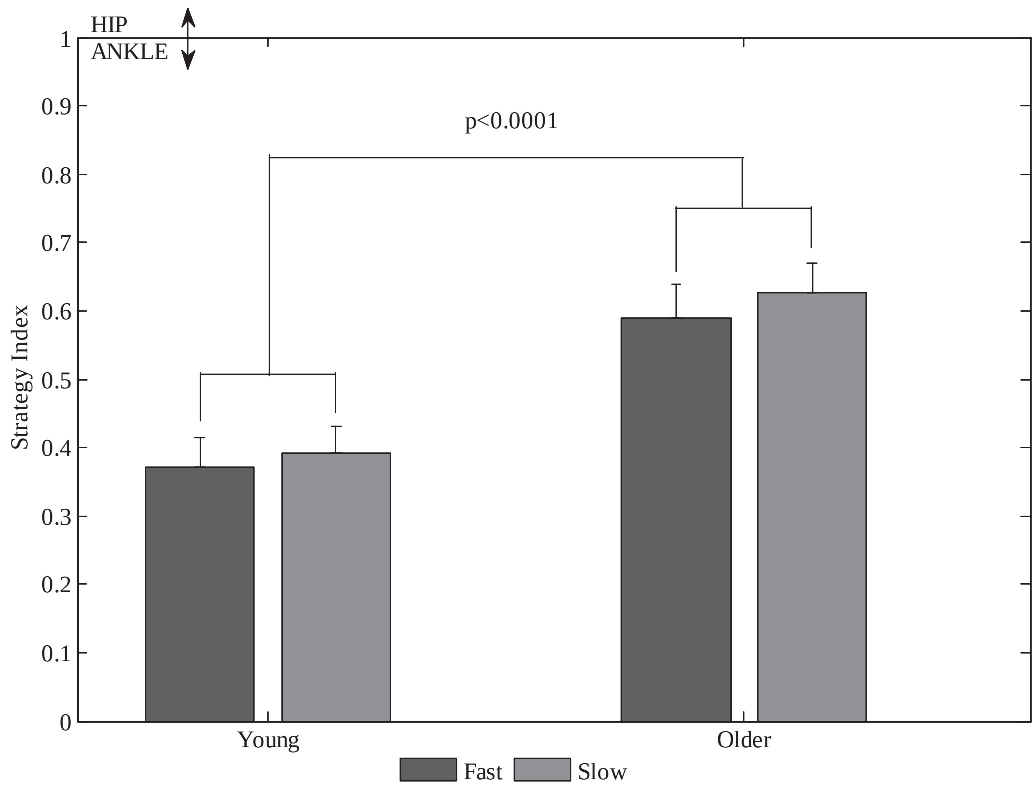

Figure 4.3: Mean strategy index with standard error plotted for both age and speed conditions. Values >1 signify more hip joint torque and values <1 signify more ankle joint torque in the mixed strategy.

A linear regression was performed relating percent contribution of passive ankle joint power to net ankle joint power and strategy index. Figure 4.4 shows the relationship of these two variables.

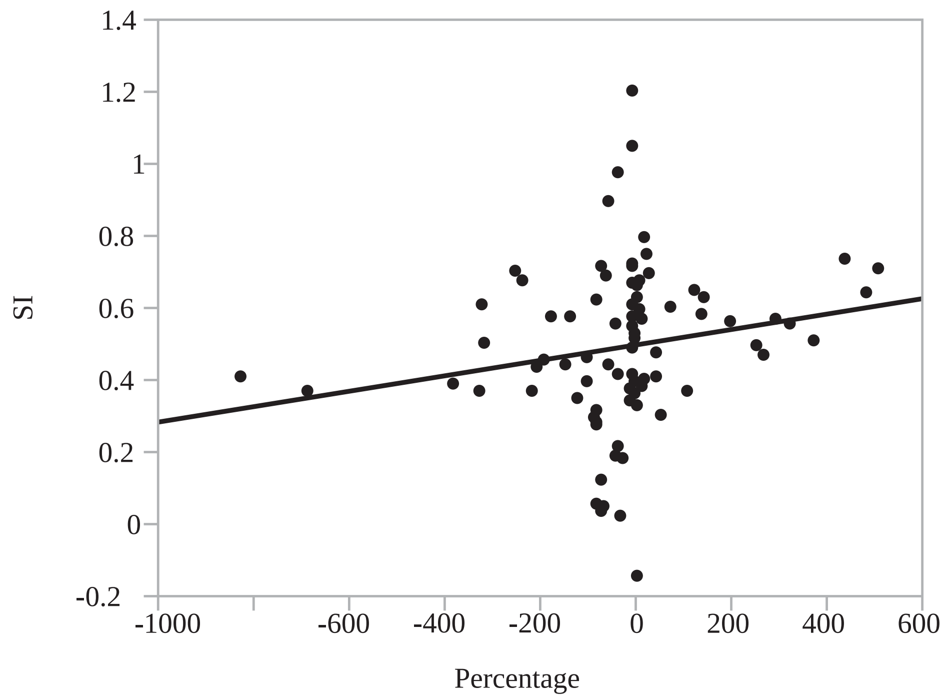

Figure 4.4: A linear regression of contribution percentage and strategy index (SI). The equation for the line of best fit was y=0.49725+0.0002141x. R^2=0.035845.

4.4 Discussion

In this study, we explored the role of passive joint power with regards to balance recovery. Specifically, we investigated the effect of age and speed on balance recovery strategy index during the dynamic phase and related the strategy index to the percent contribution of passive ankle joint power to net ankle joint power

In previous studies, joint kinematics were used as the main determinant of balance recovery strategy (F. B. Horak & Nashner, 1986; Kuo, 1995; Nashner & McCollum, 1985). These studies assess body movement to see which joint is predominantly moving. However, including kinetics in assessment of recovery strategies is important because they help indicate causation of the kinematics (Runge et al., 1999). In the current study, hip and ankle joint torques were assessed to identify if one or both joints were involved in balance recovery. If both joints were involved, a strategy index was calculated to determine the location on the continuum between pure ankle and pure hip joint strategies.

Both young and older adult subjects utilized a mixed balance recovery strategy for both fast and slow speed conditions which was to be expected. Subjects had no instructions regarding their knee position, but were free to move it as they felt necessary. Constraining the knee joint movement is a factor in seeing pure ankle and hip strategies.

While both older and young adults utilized a mixed strategy and in doing so utilized more ankle joint torque compared with hip joint torque during the dynamic phase of balance recovery, young adults had a greater component of ankle joint torque present in their balance recovery. These results indicate the beginning of a shift from ankle strategy to hip strategy in older adults in agreement with Manchester et al (1989).

It appears that perturbation speed has no effect on the strategy index, contradicting the hypothesis. This could be due to several factors. First, subjects were provided a tone that paced their movement onto the force plate. While the rate of the tone was clearly different for each speed, we could not control the exact rate at which the subject stepped onto the force plate, initiating their own perturbation. The difference in speeds may not have been sufficient to elicit a response. Additionally, our subjects were screened to be sure we were testing a recreationally active sample, both young and older adults. This screening did not account for specific activities typically performed. Preferred walking speed also was not tested, which may have skewed results as fast and slow speeds are relative to preferred walking speed.

Finally, passive ankle joint torque contribution percentage is not an adequate method of predicting strategy index. Passive ankle joint torque contribution percentage may be more closely related to other balance measures, such as sway area. Future research should investigate this relationship as it may lead to a better explanation of falls.

Chapter 5
Summary, Future Directions and Conclusion

5.1 Summary

Balance is an integral and intricate part of human movement. Balance has been linked to fall risk, and because of this risk indicator, has received much attention in the literature over the last 30 years (Goncalves et al., 2009; Graafmans et al., 1996; Thapa et al., 1996; Toraman & Yildirim, 2010). As Science & Technology continues to improve as well as mathematical models and other tools, different aspects of human balance are highlighted in hopes of pinpointing one or more factors that change as humans age, affecting balance.

Several factors have been identified that contributed to one's ability to balance that are also affected by age including muscle force, muscle power, musculotendinous stiffness, muscle activity and tendon compliance (Hasson & Caldwell, 2012; Ikezoe et al., 2012; Izquierdo, Aguado, et al., 1999; Onambele et al., 2006; Studenski et al., 1991; Thelen, 2003). These factors directly relate to the muscle-tendon unit's (MTU) to respond and produce forces to initiate movement. While these factors have been identified, they have failed to solve the problem of falls on their own.

Passive joint power is a newer concept related to human balance. Passive joint power incorporates MTU properties and contraction velocity to contribute some or all of the joint power necessary for a movement. This concept has been applied previous to gait (Silder et al., 2007; Whittington et al., 2008). We conducted a series of studies to understand if passive joint power plays a role in balance recovery, and we identified age and perturbation speed as two factors that may affect the role of passive joint power in balance recovery.

In Study 1, we investigated if passive joint power contributed to net joint power during a balance task. We also investigated if this contribution was affected by the speed of the perturbation initiating balance recovery. In Study 2, we added age to the equation to determine if passive joint power contributed to a greater extent to net joint power in older adults compared with young adults. In Study 3, we explored the relationship between passive joint power contribution and the balance recovery strategy produced by the subject.

In Study 1, we saw no difference in net ankle joint power, passive ankle joint power, individual MTU power or the contribution percentage between speeds. The speeds chosen for the particular study may not have been large enough in difference to elicit and effect. In Study 2, we found an effect of age on net ankle joint power, and on passive uniarticular plantarflexor and uniarticular dorsiflexor power. These changes maybe begin to indicate differences in balance recovery strategy which should be investigated in future studies. In Study 3, we were unable to establish a predictable relationship between percent contribution of passive ankle joint power to net ankle joint power and balance recovery strategy index. Strategy index was significantly lower in young adults compared with older adults, indicating the beginning of a shift from ankle recovery strategy towards hip strategy in older adults.

These studies are the first to investigate the role of passive ankle joint power and its contribution to net ankle joint power during a balance task. We have provided some findings that help shape the picture of the role passive ankle joint power plays in balance recovery. These findings lead us to some questions that could be answered in future investigations.

5.2 Future Directions

There are several future directions that can be pursued from this study. First, this study was performed with no differentiation for gender. Previous studies have indicated differences in muscle stiffness between genders which would affect the passive joint torque (Foure et al., 2012). Other studies have established changes in balance recovery and ankle joint torque production with gender, particularly in older adults (Carty, Cronin, et al., 2012; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001). Secondly, in this study we investigated the left leg. This was chosen arbitrarily due to the lack of clarity surrounding the issue leg dominance and leg preference. Several recent studies have begun to address the issue of leg dominance with respect to single leg stance and balance recovery (Kiyota & Fujiwara, 2014; Vieira, Coelho, & Teixeira, 2014; Young, Whitall, Bair, & Rogers, 2013). Future work should consider replicating our previous studies on the right leg for comparison. This could provide greater clarity on the role of passive ankle joint power with regards to balance recovery.

Another future direction would be to include other lower extremity joints, such as the knee and hip, to determine the role these passive contributors play in balance recovery. Whittington et al (2008) demonstrated large contributions of passive hip joint work during gait, producing an average value of 35% of the net hip joint moment. The passive joint work produced by the hip would be of particular interest in the current group of studies due to the large role the hip joint contributes to balance recovery.

Finally, it would be of importance to investigate the relationship between passive joint powers and balance measures such as sway area, sway velocity, and directional sway. Increasing the contribution of passive ankle joint power to net ankle joint power involves decreasing the amount of active control is available to produce a certain movement. This should present itself in a larger sway area or sway velocity. These measures may be more sensitive to passive joint power changes compared with balance recovery strategy.

5.3 Conclusion

This dissertation reports novel and preliminary information regarding the role that passive ankle joint power contributes towards balance recovery in young and older adults. Furthermore, we have investigated the potential link between the contribution of passive ankle joint work to net ankle joint work and balance recovery strategy to better understand human balance responses. As knowledge about human balance increases, passive joint work will be an important topic to consider as it is generated at the foundational level of muscles and tendons which are affected by age. We report preliminary results regarding passive ankle joint work, but our findings suggest that this area is important for future and further considerations.

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Appendix A

Table A.1: Raw data, subject characteristics for Studies 1-3

Subject	Age (yr)	Height (cm)	Weight (kg)	BMI (kg/m^2)	Gender
1	22	169	68.2	23.9	Female
2	24	166	79.9	29	Female
3	30	177	78.2	25	Female
4	21	164	69.5	25.8	Female
5	21	166	72.1	26.2	Female
6	26	158	57.4	23	Female
7	20	173	60	20	Female
8	24	153	49.8	21.3	Female
9	21	156	71.7	29.5	Female
10	29	171	66.6	22.8	Female
11	21	185	74.3	21.7	Male
12	20	174	67.2	22.2	Male
13	23	183	71	21.2	Male
14	29	186	93.5	27	Male
15	24	178	85.8	27.1	Male
16	30	183	78.4	23.4	Male
17	21	179	88.5	27.6	Male
18	20	172	70.3	23.8	Male
19	21	190	72.7	20.14	Male
20	23	168	69.4	24.6	Male
21	20	186	89	25.7	Male
22	65	157	76.3	31	Female
23	67	153	59.5	25.4	Female
24	60	166	85.5	31	Female
25	64	163	62.3	23.4	Female
26	69	154	63.7	26.9	Female

27	62	161	57.5	22.2	Female
28	60	165	65.9	24.2	Female
29	60	171	70.7	24.2	Female
30	63	164	74.3	27.6	Female
31	60	169	66.8	23.4	Female
32	68	183	75.5	22.5	Male
33	62	183	95.9	28.6	Male
34	64	174	83.9	27.7	Male
35	61	185	96.9	28.3	Male
36	68	171	73.8	25.2	Male
37	69	163	59.8	22.5	Male
38	63	183	88.6	26.5	Male
39	63	176	95.8	30.9	Male
40	65	169	76	26.6	Male
41	64	180	74.6	23	Male

Table A.2: Raw data, subject physical activity questionnaire responses

Subject	Sessions/Wk	Session Length (min)	Intensity
1	2	4	1
2	3	4	2
3	4	4	4
4	4	5	3
5	4	4	3
6	4	4	3
7	2	1	1
8	3	4	2
9	3	4	2
10	3	4	3
11	2	2	1
12	3	4	2
13	3	5	2
14	4	4	2
15	4	4	3
16	1	1	1
17	2	2	2
18	4	5	3
19	2	2	2
20	4	2	2
21	4	4	3
22	3	3	3
23	3	3	2
24	3	3	2
25	3	2	3
26	3	2	2
27	3	4	2
28	3	4	1

29	4	4	3
30	2	1	1
31	4	5	2
32	3	4	2
33	2	2	2
34	3	4	3
35	2	1	2
36	4	3	2
37	3	4	2
38	2	3	2
39	4	4	3
40	3	2	2
41	3	3	3