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Abstract approved:

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A backward fall is a major cause of hip fractures. Hip fractures have the potential to increase the mortality rate of older adults. Thus, the prevention of backward falls by older adults is a critical subject to be studied. Specifically, it remains unknown what effect aging-related losses in muscle strength have on the ability to recover balance by stepping following a backward loss of balance. There were two specific aims in this study. The first aim was to develop two-dimensional musculoskeletal models of young and older adults. The second aim was to quantify the effects of the normal aging-related loss in lower extremity strength on the ability to recover balance following a backward step from a backward balance loss by describing the feasible regions of balance recovery. Two-dimensional musculotendon models of young and older adults were developed and used in a six-link model to simulate backward balance recovery of young and older adults. The feasible regions for young and older adults were described for two combinations of initial backward and downward velocities: 15% body height/s (*Bh/s*) (condition 1) and 30% *Bh/s* (condition 2). For condition 1, older adults showed a 25% foot length (*fl*) more posterior forward boundary at a hip height of

50% *Bh* and a 12.2% *fl* more anterior rear boundary at 47.5% *Bh* than young adults. For condition 2, older adults showed a 6.4% *fl* more posterior forward boundary at 48.5% *Bh* and a 12.8% *fl* more anterior rear boundary at 47.5% *Bh* than young adults. The feasible regions for young and older adults showed similar ranges of hip height in conditions 1 and 2. The results suggest that older adults are less able than young adults to regain balance after a very short or very long backward step from a backward loss of balance. Thus, the aging-related loss in muscle strength could impair the ability to recover balance during the stance phase after a backward recovery step. However, if a medium step is taken, older adults can show the same ability to recover balance as young adults.

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Effects of the Aging-Related Loss in Lower Extremity Strength on the Feasible Region for Balance Recovery

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I understand that my dissertation will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my dissertation to any reader upon request.

Norio Kadono, Author

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CONTRIBUTION OF AUTHORS

Dr. Pavol was involved in and assisted with the design, data collection, interpretation, and writing for Chapters 2 and 3.

TABLE OF CONTENTS

TABLE OF CONTENTS (Continued)

LIST OF FIGURES

LIST OF FIGURES (Continued)

LIST OF TABLES

Effects of the Aging-Related Loss in Lower Extremity Strength on the Feasible Region for Balance Recovery

1. Introduction

Falls can have severe impacts on an older adult's life. In 2000, 10,000 older adults died from falls and 2.6 million older adults were treated at hospitals for fall-related injuries (Stevens et al., 2006). The medical costs of fatal and non-fatal fall-related injuries in 2000 were $19.2 billion (Stevens et al., 2006). Thus, the prevention of falls and fall-related injuries is an important societal issue.

Among the most serious concerns related to falls is the potential to induce hip fractures (Nevitt et al., 1993). Hip fractures were seen among 0.2% of older women aged 65 or older who experienced a fall (Nevitt et al., 1993). Leibson et al. (2002) reported that hip fractures increased the mortality rate in older adults by 10%. Older adults who suffer from hip fractures also develop limitations in mobility and lose independence (Wolinsky et al., 1997). Fall-induced hip fractures can thus decrease older adults' quality of life. Backward falls are associated with the largest risk of hip fractures among different types of falls (Smeesters et al., 2001). Therefore, there is a need to understand the factors that influence the ability of an older adult to prevent a backward fall.

Backward falls could result from aging-related changes in muscle. A decline in muscle strength results from changes in the structure and properties of muscle with advancing age (Larsson et al., 1979; Sato et al., 1984; Frontera et al., 1991, 2000; Thompson, 2002). Among the changes observed, older adults lose 10% of the peak muscle mass that they acquired at young ages by age 65 years

(Lindle et al., 1997) and older adult muscles exhibit a greater proportion of slow twitch fibers (Larsson et al., 1978). As a result of these changes, older adults lose 20 to 60% of peak muscle torque, power, and shortening velocity (Thom et al., 2007).

It has been reported that lower extremity muscle strength plays an important role in maintaining balance (Duncan et al., 1993; Wolfson et al., 1995; Daubney & Culham, 1999). Older adults who had difficulties in controlling balance showed lower ankle dorsiflexion and plantarflexion strength than older adults who had no difficulties in controlling balance (Duncan et al., 1993). Carter et al. (2002) and Karinkanta et al. (2005) found that knee extensor strength is significantly associated with static and dynamic balance. Consistent with these studies, Whipple et al. (1987) found that fallers showed lower peak torques and powers at the ankle and knee joints than non-fallers. In addition, Daubney and Culham (1999) reported that older adults who experienced no falls showed greater strength in ankle dorsiflexion and hip extension than older adults who experienced falls. Thus, fallers may have weaker muscle strength at the ankle, knee, and hip joints than non-fallers. The physical weakness of fallers could be associated with falls.

In keeping with the studies that found that lower extremity strength is associated with static and dynamic balance (Carter et al., 2002; Binda et al, 2003; Karinkanta et al., 2005), the effects of strength training on balance control and falls prevention have been studied (Amiridis et al., 2005; Hauer et al., 2001; Barnett et al., 2003; Orr et al., 2006). Amiridis et al. (2005) reported that older adults who had four weeks of ankle strength training showed an increase in dorsiflexion moment and an improvement in static balance. Orr et al. (2006) reported that a 10-week knee strength training program increased knee strength and improved static and dynamic balance control among older adults. The studies of Hauer et al. (2001) and Barnett et al. (2003) showed that an exercise training program could increase lower extremity strength among weak older adults and that weak older adults who underwent the training program showed lower rates of falling than those who did not undergo the training. Therefore, physically weak older adults can improve balance control and prevent themselves from falling by participating in an exercise training program. Lower extremity strength could be a biomechanical factor that contributes to controlling static and dynamic balance.

Successful recovery from a loss of balance requires a change in the base of support, typically by stepping. A balance recovery motion with a step consists of a stepping phase followed by a stance phase. It has been reported that aging-related losses in muscle strength can influence the stepping phase among healthy older adults (Thelen et al, 1997; Wojcik et al., 1999; Thelen et al., 2000; Madigan & Lloyd, 2005b). An older adult generates ankle joint moments at a slower rate than a young adult, and stepping velocity by an older adult is slower than that by a young adult (Thelen et al., 1996; Wojcik et al., 1999). Consistent with this, Madigan and Lloyd (2005b) reported that older adults showed lower peak ankle, knee, and hip joint velocities in the stepping leg than young adults. Thus, a decrease in stepping velocity of an older adult could result from decreases in the rates of joint moment development that lead to reduced peak joint velocities in the lower extremity. The lower stepping velocity could affect the ability of an older adult to recover balance with a step.

In addition to the stepping phase, the stance phase following a balance recovery step by older adults has been studied (Madigan & Lloyd, 2005a). Since the stance phase requires a person to support the weight of the body and stop its downward motion, the stance phase may require greater lower extremity strength than the stepping phase. Madigan and Lloyd (2005a) found that older adults generated smaller knee extension moments than young adults in recovering from a forward loss of balance and relied on ankle plantarflexion and hip extension moments to compensate for a smaller knee extension moment. On the other hand, Pavol et al. (2002) studied the characteristics of muscle strength in fallers and non-fallers among older adults and found that fallers showed greater lower extremity strength than non-fallers. Thus, it still remains unknown whether the ability to recover from a backward loss of balance is deteriorated by the normal aging-related loss in muscle strength.

The feasible region for balance recovery can be used to determine the effect of the normal aging-related loss in muscle strength on the ability to regain stability after a loss of balance in the backward direction. The feasible region for balance recovery has been described in terms of the range of initial anteroposterior positions and velocities of the center of mass (COM) for which the COM can be brought to rest above the base of support (BOS) (Pai & Patton, 1997; Iqbal & Pai, 2000). The feasible region for dynamic balance recovery can be computed using a musculoskeletal model. Since a musculoskeletal model can simulate the behaviors of young and older adult muscles, a dynamic model of balance recovery can determine the effect of the normal aging-related loss in muscle strength on the ability to perform a backward balance recovery motion.

The feasible region has been used to study the ability to prevent a loss of balance (Pai & Patton, 1997; Pai & Iqbal, 1999; Iqbal & Pai, 2000; Yang et al., 2007). Pai and Patton (1997) used the feasible region to investigate the effect of the normal aging-related loss in muscle strength on the ability to maintain balance in a standing position, without stepping. They found that the model of an older adult had a smaller feasible region than that of a young adult. Therefore, the normal aging-related loss in muscle strength can affect the ability to prevent a backward balance loss. However, if a balance loss occurs, stepping is generally required to regain balance. This stepping can result in considerable hip and knee flexion at step touchdown (Pavol & Pai, 2007). The model of Pai and Patton (1997) could not take this hip and knee motion into account. Thus, it remains unknown whether the normal aging-related loss in muscle strength can deteriorate the ability to control balance after the completion of a backward step following a backward balance loss. The present study will address this issue and the limitation of the model of Pai and Patton. In this study, a modified version of the feasible region will be employed. The modified version of the feasible region will consist of the ranges of initial COM horizontal positions and hip vertical positions from which a person can bring the COM to rest and regain balance for a given initial backward and downward velocity.

In the present study, a dynamic musculoskeletal model was developed and used to investigate what effect the aging-related loss in muscle strength has on the feasible region of balance recovery, and hence the ability to prevent a backward fall. In general, the components of a musculoskeletal model are rigid links representing the segments of the body, joints between these links, musculotendon actuators that simulate muscles acting across the joints, and passive joint moments that simulate the passive restraints to joint motion. In the model that was developed, Hill-type musculotendon actuators, governed by simulated control signals, were used to simulate the properties of young and older adult muscles.

The components of Hill-type muscle models are a contractile element (*CE*) that represents the active state dynamics of muscles and a passive elastic (*PE*) element and a series elastic (*SE*) element that represent the passive state dynamics of muscle and tendon, respectively (Audu & Davy, 1985; Zajac, 1989). The active muscle force generated by the *CE* is determined by its activation dynamics, length-tension, and force-velocity relationships. The *PE* and *SE* elements are modeled as nonlinear-elastic springs and generate passive muscle and tendon forces, respectively. The behaviors of the muscle activation dynamics, length-tension and force-velocity relationships, and nonlinear-elastic springs are determined by adjusting parameter values in a muscle model.

Hill-type muscle models can be used as equivalent or individual muscle actuators (Audu & Davy, 1985). Equivalent muscle actuators model multiple muscles that have the same function as one muscle group; individual muscle actuators represent a single muscle. Equivalent muscle actuators have less complex structures and lower computational expenses than if multiple individual muscle actuators were used (Audu & Davy, 1985; Blemker & Delp, 2006). Equivalent and individual muscle actuators can be used to examine the contributions of muscle forces produced by muscle groups and single muscles, respectively, to a desired motor task. In addition, passive joint moment models can simulate passive joint moments developed by the elasticity of the soft tissues to enforce joint ranges of motions within anatomical ranges (Riener & Edrich, 1999). Hence, in the first part of the present study, Hill-type muscle models and passive joint moment models were designed to create a musculoskeletal model that could be used to determine the effect of normal aging-related losses in muscle strength on the ability to prevent a backward fall.

Musculoskeletal models have been used in forward dynamic simulations to investigate how single muscles or muscle groups contribute to a motion of the body (Audu & Davy, 1985; Davy & Audu, 1987; Thelen & Anderson, 2006). However, forward dynamic simulation demands high computational expenses to determine an optimal control signal. In order to decrease computational expenses, the numbers of muscle models and control signals need to be minimized to the extent possible. Therefore, our goal in the first part of the present study was to develop a musculoskeletal model with the least number of muscle models for use in forward dynamic simulations.

Through adjusting the values of muscle actuator parameters to simulate the normal aging-related declines in muscle strength, forward dynamic simulations can be used to examine the effect of these declines in strength on accomplishing a desired motor task. The parameter values determine the characteristics of muscle activation dynamics, length-tension and force-velocity relationships, and nonlinear-elastic behavior of the *PE* and *SE* elements. In the first part of the present study, parameter values of young adult muscles were found to fit the simulated joint moments to data reported in the literature. The behavior of older adult muscles could then be simulated by adjusting the muscle model parameter values determined for young adults. The adjusted parameter values for older adults were identified by reviewing the literature to determine the effects of aging on the force-generating properties of muscle and tendon. These two sets of parameter values were then used in determining the feasible regions of balance recovery for young and older adults in the second part of this study.

1.1 Rationale & Purpose

Falls and fall-related injuries are major problems for older adults. In particular, backward falls are a main cause of hip fracture. Thus, there is a need to focus on the prevention of backward falls to protect the quality of life of older adults. The effects of normal aging-related declines in strength on the ability to regain balance have been studied. However, there are still conflicts in the results of previous studies. Madigan and Lloyd (2005a) found that older adults generated significantly lower knee extensor moments than young adults in stabilizing the body after completing a step for balance recovery. On the other hand, Pavol et al. (2002) found that older adult fallers after a trip showed significantly stronger lower extremity muscles than older adult non-fallers. Therefore, it still remains unclear whether normal aging-related declines in lower extremity strength influence the ability of a healthy older adult to recover from a backward balance loss. A mathematical modeling approach was thus employed to investigate the effects of aging-related declines in lower extremity strength on the ability to regain balance following a backward step after a backward loss of balance. A musculoskeletal model was developed to simulate the behaviors of young and older adult muscles. This model was then used to study the effect of aging-related declines in lower extremity strength on the ability to restore stable posture following a backward step after a backward balance loss.

1.2 Specific Aims and Hypothesis

The aims of this study were as follows:

1. To develop a musculoskeletal model of lower extremity muscle strength and determine parameter values of the musculoskeletal model to simulate the behavior of young and older adult muscles.
2. To quantify the effect of the age-related decline in lower extremity muscle strength on the ability to stabilize the body after completing a stepping motion following a backward balance loss. The feasible region for balance recovery, as computed using the musculoskeletal model developed, was compared between young and older adults.

I hypothesized that the model of older adults would show a smaller feasible region for balance recovery than that of young adults.

A two-dimensional musculoskeletal model of young and older adults

Norio Kadono and Michael J. Pavol

2.1 Introduction

A musculoskeletal model consists of a system of linked segments driven by actuators that simulate the forces produced by the active and passive elements of muscles and the series elastic elements of tendons. A musculoskeletal model is used in forward dynamic simulation to better understand human motion. An advantage of a musculoskeletal model is its ability to simulate different musculoskeletal properties through adjustments in its parameter values. The structure of muscles undergoes alterations with advancing age (Larsson et al, 1978; Frontera et al., 1991; Frontera et al., 2000). Aging deteriorates musculoskeletal contractile properties (Lindle et al., 1997; Thompson, 2002). Thus, older adult muscles behave differently from young adult muscles. This could cause young and older adults to show different movement patterns in performing the same motor task (Madigan & Lloyd, 2005). Specifically, the aging-related loss in dynamic lower extremity strength may have an effect on falls by older adults (Whipple et al., 1987; Pavol et al., 2002). However, it still remains unknown whether the aging-related loss in dynamic lower extremity strength impairs the ability of older adults to prevent themselves from falling. By modeling the behaviors of young and older adult muscles, forward dynamic simulation can contribute to investigating the theoretical effects of the aging-related loss in dynamic lower extremity strength on the ability of older adults to recover balance. Therefore, there is a need for musculoskeletal models that can simulate the lower extremity strength of young and older adults and are suitable for forward dynamic simulation.

Aging-related changes in muscle structure and strength have been studied (Larsson et al., 1978; Sato et al., 1984; Thom et al., 2005; Ochala et al., 2007). As their age advances to 65 years, older adults lose 10 and 20% of the peak muscle mass and volume, respectively, that they had as young adults (Lindle et al., 1997; Thom et al., 2005). The aging-related losses of muscle mass and volume result from reductions in the sizes and numbers of single muscle fibers (Sato et al., 1984; Aniansson et al., 1986; Häkkinen & Häkkinen, 1991). Older adults have significantly smaller cross-sectional areas of muscles than young adults (Häkkinen & Häkkinen, 1991). Specifically, the cross-sectional areas in type IIA and IIB fibers are reduced by 15 – 25% (Aniansson et al., 1986). In addition, the number of type IIA and IIB fibers preferentially decreases with advancing age (Larsson et al., 1978; Ishihara & Araki, 1988). Older adults aged in their 70's have lost 35% and 40% of type I and II fibers, respectively, that they had in their 30's (Sato et al., 1984). Those structural changes in muscles result in 20 – 60% reductions in peak torque, power, and shortening velocity of muscles (Thom et al., 2007). Those aging-related changes in muscle strength can be simulated by adjusting parameter values in a musculoskeletal model (Thelen, 2003).

The determination of an optimal control signal has been an issue in forward dynamic simulation (Audu & Davy, 1985). Computational expenses to determine an optimal control solution are proportional to the number of muscles and complexity of design of a musculoskeletal model. Although a three-dimensional musculoskeletal model can model muscle paths and moment arms accurately, it requires a large number of muscles and complex design. A three-dimensional musculoskeletal model thus involves high computational expense to find an optimal control signal (Blemker & Delp, 2005). Since a two-dimensional musculoskeletal model requires fewer muscles and a less complex design, a two-dimensional musculoskeletal model is more suitable to forward dynamic simulation than a three-dimensional model.

Aging-related losses in muscle strength can potentially cause different movement patterns between young and older adults. To better understand the different movement patterns, there is a need to develop a musculoskeletal model that can model the behaviors of young and older adult muscles and that is suitable to forward dynamic simulation. Specifically, we sought to design a musculoskeletal model that can be used to investigate the theoretical effects of the aging-related loss in dynamic lower extremity strength on the ability of older adults to recover balance. Thus, the purpose of this study was to develop a musculoskeletal model of the lower extremity strength of young and older adults. Moreover, the model was limited to a two-dimensional musculoskeletal model with a small number of musculotendon actuators to save computational expenses in finding optimal control solutions when applied to forward dynamic simulation.

2.2 Methods

2.2.1 Musculoskeletal Model

A musculoskeletal model of the lower extremity was developed. The model consists of rigid foot, leg, thigh, and pelvis segments linked by hinge joints representing the ankle, knee, and hip, and driven by a system of 10 musculotendon actuators (Table 2.1), designed to simulate flexion-extension strength. The system includes seven uniarticular actuators: soleus, dorsiflexors, vastii, biceps femoris (short head), gluteals, adductor magnus, and iliopsoas, and three biarticular actuators: gastrocnemius, hamstrings, and rectus femoris (Figure 2.1). Three of these actuators: soleus, biceps femoris, and rectus femoris, represent individual muscles, whereas the other seven are equivalent actuators that represent multiple muscles.

2.2.2 Geometry of the Musculoskeletal Model

Four rigid segments: foot, leg, thigh, and pelvis, comprise the two-dimensional musculoskeletal model of the lower extremity. One degree-of-freedom hinge joints, representing the ankle, knee, and hip joints, link the rigid segments. We modeled a young or older adult whose height and mass were 1.8m and 75kg, respectively. This study used data of musculoskeletal geometry provided by Delp (1990). Local talus, tibia, femur, and pelvis coordinate systems are attached to the foot, leg, thigh, and pelvis segments, respectively, with the Y axis aligned with the longitudinal axis of the segment and the X axis directed anteriorly relative to the segment (Figure 2.1). The ankle joint center is located at (X,Y) coordinates of (0, 0) in the talus coordinate system and (0, -0.43m) in the tibia coordinate system; the knee joint center is at (-0.0043m, -0.0148m) in the tibia coordinate system and (-0.0104m, -0.4088m) in the femur coordinate system; the hip joint center is at (0, 0) in the femur coordinate system and (-0.0707m, -0.0661m) in the pelvis coordinate system (Delp, 1990). Ankle dorsiflexion, knee extension, and hip flexion are defined as positive joint rotations at the ankle, knee, and hip joints, respectively (Delp, 1990).

The musculotendon lengths (l^{MT}) and lines of action were determined using a method similar to Delp (1990). The l^{MT} consisted of a variable-length segment across each joint spanned, plus a constant-length segment. The variable-length segment is a straight muscular path segment between endpoints and spans one or more joints. The constant-length segment crosses no joints. Endpoints for the variable-length segment and the length of the constant-length segment were derived from the attachment points and wrapping points of Delp (1990). Delp supplied the three-dimensional coordinates of these points in the pelvis, femur, tibia, patella, talus, calcaneus, and toes reference frames. Because we developed a two-dimensional musculoskeletal model, two-dimensional coordinates of attachment (i.e. origin and insertion) and wrapping points were used in this study. Points in the calcaneus and toes frames were transformed into the talus frame, assuming that the foot was in its neutral position. For further simplification, we derived a best-fit fixed knee joint center location within the tibia and femur coordinate systems from the translational functions between the tibia and femur frames provided by Delp (1990). The attachment points of the single musculotendon actuators were the points presented by Delp (1990). Since the production of active muscle force is proportional to muscle physiological cross-sectional area (PCSA) (An et al., 1981; Kaufman et al., 1991), we averaged the attachment points of the component muscles of the equivalent musculotendon actuators based on PCSA. Typically, the origin and insertion of a component muscle were used as its attachment points; however, where the path of a component muscle wrapped over bones, wrapping points were used instead. Except in the cases of the gastrocnemius and vastii actuators, we used the endpoint or wrapping point from Delp (1990) closest to the joint for each component muscle in computing the proximal and distal PCSA-weighted endpoints of the variable-length segment. Similarly, the length of the constant-length segment was computed as the PCSA-weighted average of the constant-length segments of the component muscles, as computed from the attachment and wrapping points of Delp (1990). For the gastrocnemius and vastii actuators, we transformed the angle-dependent wrapping/attachment points at the knee into the corresponding PCSA-weighted endpoints of the variable-length segments by fitting the l^{MT} and moment arm (r_i) to those developed by Delp (1990). Muscular geometry data of the 10 musculotendon actuators are presented in Table 2.2.

The r_i, except those of the knee extensors, are calculated geometrically from the line of action of the variable-length segment across a joint. Yamaguchi and Zajac (1989) developed a quadriceps effective moment arm model (r_{eff}) for the knee extensor moment. r_{eff} is expressed by the following equation (Yamaguchi & Zajac, 1989):

$$r_{eff} = \left(\frac{F_{pl}}{F_q}\right) r_{act} \tag{2.1}$$

where F_{pl} is tensile force generated by the patella ligament, F_q is the net force applied by the quadriceps, and r_{act} is actual moment arm of the patella ligament with respect to the contact point of the tibia and femur. We used the ratio of F_{pl} to F_q provided by Yamaguchi and Zajac (1989). The best fit moment arm equation of the vastii was derived and used as r_{act} in this musculoskeletal model:

$$r_{act} = -0.0044\theta_{KNE}^2 + 0.0045\theta_{KNE} + 0.0526 \tag{2.2}$$

where θ_{KNE} is knee extension angle (rad). At the knee joint, the anatomical position corresponds to an angle of zero. r_{eff} is used as the moment arm of the knee extensors (i.e. vastii and rectus femoris).

2.2.3 Musculotendon Actuators

Hill-type musculotendon models were used to drive the two-dimensional musculoskeletal model. The Hill-type model consists of a contractile (*CE*) element, a passive elastic (*PE*) element, and a series elastic (*SE*) element (Figure 2.2) (Audu & Davy, 1985; Zajac, 1989). The *CE* element represents the active force-generating properties of muscle, whereas the *PE* and *SE* elements represent the passive force-generating properties of muscle and tendon, respectively. The *CE* and *PE* elements are aligned in parallel, and both elements act in series with the *SE*. Thus, the muscle force (F^M) is the sum of the active force generated by the *CE* element (F^{CE}) and the passive force generated by the *PE* element (F^{PE}), and is equal to the force acting in the *SE* element (F^{SE}):

$$F^{M} = F^{CE} + F^{PE} = F^{SE} \tag{2.3}$$

The *CE* element functions as the active-force generator in the musculotendon actuator. The F^{CE} is determined by its time-dependent activation level [$a(t)$], its length-tension relationship [$f^{LT}(i_a, \varepsilon^M)$], its force-velocity relationship [$f^{FV}(\dot{\eta})$], and its maximum isometric force (f_o^{MAX}).

A dimensionless neural excitation signal (*n*), ranging between 0 (no excitation) and 1 (maximum excitation), is used as the control input of each musculotendon actuator. *n* determines $a(t)$ of the *CE*. The relationship governing $a(t)$ is simulated by a non-linear first-order differential equation (Zajac et al, 1986; Thelen, 2003):

$$\dot{a}(t) + \left(\frac{1}{\tau}\right)a(t) = \left(\frac{1}{\tau}\right)n \qquad \tau = \begin{cases} \tau_a & if \quad n > a(t) \\ \tau_d & if \quad n \le a(t) \end{cases} \tag{2.4}$$

where τ_a is the activation time constant, and τ_d is the deactivation time constant.

The generation of F^{CE} is influenced by the length of the muscle (l^M) relative to its optimal muscle length (l_o^M) (Brooks et al., 2005). The maximum isometric active force can be generated when the muscle contracts at its l_o^M. Isometric active force decreases as l^M becomes greater or smaller than l_o^M. The shape of the relationship between muscle length and isometric force is affected by the angle at which the muscle fibers are aligned with respect to the line of action of the muscle along its tendons. This dependence can be expressed in terms of the muscle index of architecture (i_a), which is the ratio of fiber length to muscle length at l_o^M (Kaufman et al., 1991). Kaufman et al. (1991) expressed the active $f^{LT}(i_a, \varepsilon^M)$ of muscle by the following equation, which is used in the model:

$$f^{LT}(i_a, \varepsilon^M) = \exp\left[-\left[\frac{\left(\varepsilon^M + 1\right)^{\{0.96343[1-1/i_a]\}} - 1}{0.35327(1-i_a)}\right]^2\right] \text{ for } i_a < 1 \quad (2.5a)$$

$$f^{LT}(i_a, \varepsilon^M) = \exp\left[-(2.727\ln(\varepsilon^M + 1))^2\right] \quad \text{for } i_a = 1 \quad (2.5b)$$

where ε^M is muscle strain, determined as $(l^M - l_o^M)/l_o^M$.

Finally, the production of F^{CE} is associated with the velocity of isotonic muscle contraction (Kaufman et al., 1991). F^{CE} decreases progressively to zero as the muscle shortening velocity increases to its maximum strain rate ($\dot{\varepsilon}_o^M$), whereas F^{CE} increases to approximately 1.33 times its isometric value as the muscle lengthening velocity increases (Kaufman et al., 1991). The $f^{FV}(\dot{\eta})$ can be expressed by the following equation, developed by Hatze (1981) and used in the model:

$$f^{FV}(\dot{\eta}) = \frac{0.1433}{0.1074 + \exp\left[-1.409\sinh(3.2\dot{\eta} + 1.6)\right]} \quad (2.6)$$

where $\dot{\eta}$ is the normalized strain rate of the muscle, determined by $\dot{\varepsilon}^M / \dot{\varepsilon}_o^M$.

The *CE* element force is found by scaling its maximum isometric force by its activation level and the influence of its length-tension and force-velocity relationships. Thus, the F^{CE} is expressed by the following function:

$$F^{CE} = a(t) \cdot f^{LT}(i_a, \varepsilon^M) \cdot f^{FV}(\dot{\eta}) \cdot f_o^{MAX} \quad (2.7)$$

The *PE* and *SE* elements play roles as the passive-force generators in the actuator. Both elements behave as nonlinear springs. F^{PE} in the model is determined by muscle strain, the passive muscle force gain (k_1^M), and the passive muscle stiffness (k_2^M):

$$F^{PE} = k_1^M \exp(k_2^M \varepsilon^M) \tag{2.8}$$

F^{SE} is determined by tendon strain (ε^T) and normalized tendon stiffness ($\widetilde{k}^T$):

$$F^{SE} = \begin{cases} \widetilde{k}^T (\varepsilon^T)^2 & if\ \varepsilon^T \geq 0 \\ 0 & if\ \varepsilon^T < 0 \end{cases} \tag{2.9}$$

where ε^T is given as $(l^T - l_o^T)/l_o^T$ where l^T and l_o^T are tendon length and tendon slack length, respectively, and $\widetilde{k}^T$ was given as $f_o^{MAX}/(\varepsilon_o^T)^2$, where ε_o^T is maximum tendon strain.

The musculotendon actuator introduces length-dependent relationships among the *CE, PE,* and *SE* elements. The length of the muscle, which is the same as the lengths of the *PE* and *CE* elements, depends on that of the *SE* element and that of the musculotendon actuator. Similarly, the shortening velocities of the muscle, *SE* element, and musculotendon actuator are related. This allows equations (2.1) – (2.7) governing the actuator force to be solved.

2.2.4 Musculotendon Actuator Parameters

The behavior of each musculotendon actuator is specified by a set of 11 parameters (Table 2.3). First, we determined the parameter values in the 10 musculotendon actuators to simulate the strength characteristics of young adults.

The procedure used to determine the musculotendon parameter values for young adults was as follows. The values of τ_a and τ_d were those developed by Thelen (2003). The values of l_o^M were determined by modifying the following equation (Hoy et al., 1990):

$$dl^{MT} = r_i d\theta_i \tag{2.10}$$

where dl^{MT} and $d\theta_i$ are changes in musculotendon actuator length (l^{MT}) and joint angle (θ_i) and r_i is moment arm. Making the simplifying assumption that the tendon length remains constant, the equation was converted into:

$$\frac{dl^M}{l_o^M} = \frac{r_i}{l_o^M} d\theta_i \tag{2.11}$$

l_o^M was determined by taking an average of r_i for the actuator over a physiological range of motion and dividing this by the PCSA-weighted average r_i / l_o^M of the component muscles. The values of l_o^M of the component muscles were averages of l_o^M reported by Wickewicz et al. (1983) and Friederich and Brand (1990). Wickewicz et al. assumed that an optimal sarcomere length is 2.2 μm; Friederich and Brand assumed 2.8 μm. In this study, an optimal sarcomere length was assumed to be 2.8 μm. Thus, l_o^M provided by Wickewicz et al. was multiplied by 2.8/2.2 and used in this model.

The method to determine values of f_o^{MAX} , i_a , k_1^M , k_2^M , and l_o^T was as follows. We used a least-squares optimization technique and found the five parameter values at one time by fitting the simulated isometric joint moments to corresponding experimental data from multiple joint angles. The optimization to find the parameter values was conducted in four separate optimizations: 1) dorsiflexion optimization that found the parameter values of the dorsiflexor actuator, 2) plantarflexion optimization of the gastrocnemius and soleus actuators, 3) knee extension-hip flexion optimization of the vastii, rectus femoris, and iliopsoas actuators, 4) knee flexion-hip extension optimization of the hamstrings, biceps femoris, gluteals, and adductor magnus actuators. Since experimental data were chosen from different studies, we normalized the experimental joint moments to the subjects' body height and mass to compensate for differences in subject strength between studies. In the case where subjects' height and mass or weight were not provided in the study, we scaled the experimental data to the joint moment, normalized to body height and weight, at the corresponding joint angle in a different study. In the knee extension-hip flexion and knee flexion-hip extension optimizations, the experimental knee flexion and extension moments at two different positions, supine and seated (i.e. hip angles of 0º and 90º), were used as reference data to take into account the properties of biarticular muscules (i.e. hamstrings and rectus femoris).

In the dorsiflexion optimization, the values of f_o^{MAX} , i_a , k_1^M , k_2^M , and l_o^T in the dorsiflexor actuator were determined by fitting the active component of the simulated dorsiflexion moment to the corresponding experimental data provided by Koh and Herzog (1995), measured with the knee flexed 90º. In the plantarflexion optimization, the experimental ankle plantarflexion moments of Bobbert and Van Ingen Schenau (1990), measured with the knee flexed 0º and consisting of the summed active and passive moments, were used to determine the parameter values in the soleus and the gastrocnemius actuators simultaneously. In the knee extension-hip flexion optimization, we used the experimental knee extension moments provided by Houtz et al. (1958), measured with the hip flexed 0º and 90º; however, the magnitudes were adjusted by the value of the knee extension moment at 60º of knee flexion and the subjects' height and weight in the study of Knapik et al. (1983). The experimental knee extension moments comprised only an active moment. Reference data for the hip flexion moment were based on the experimental hip flexion moments, in which the hip and knee flexed equal angles, provided by Williams and Stutzman (1959). The magnitudes of the experimental hip flexion moment were adjusted by the value of the hip flexion moment at 10º of hip flexion and the subjects' height and weight in the study of Cahalan et al. (1989). The experimental hip flexion moment consisted of only an active moment. We determined the parameter values of the vastii, rectus femoris, and iliopsoas actuators at the same time by fitting the simulated knee extension and hip flexion moments simultaneously to the experimental data. In the knee flexion-hip extension optimization, the reference data for the active component of the knee flexion moments were obtained from the ratio between the knee extension and flexion moments reported by Houtz et al (1958). The reference data for the hip extension moment were the experimental hip extension moments provided by Németh et al. (1983). The experimental hip extension moment was measured with the knee flexed 30º and comprised only an active moment. The parameter values of the hamstrings, biceps femoris, gluteals, and adductor magnus actuators were determined at the same time by fitting the simulated knee flexion and hip extension moments simultaneously to the corresponding experimental data. In each set of optimizations for f_o^{MAX}, i_a, k_1^M, k_2^M, and l_o^T, we also included reference data for the corresponding passive joint moment. These reference data for the passive joint moments were taken from the equations developed by Riener and Edrich (1999).

Values of $\dot{\varepsilon}_o^M$ were based on muscle fiber-type distributions (Kaufman et al., 1991). We used 2.0 and 7.5 muscle lengths/s provided by Kaufman et al. (1991) as the unloaded shortening rates for slow and fast muscle fibers, respectively. Values of $\dot{\varepsilon}_o^M$ were determined as a weighted sum of the unloaded shortening rates for slow and fast muscle fibers, with the weights based on the corresponding fiber-type proportions. We used muscle fiber-type distributions provided by Johnson et al. (1973). Reported fiber-type distributions were used directly for musculotendon actuators representing individual muscles. For equivalent actuators, the fiber-type distribution was averaged across the component muscles.

Finally, Zajac (1989) assumed that F^{SE} attains the magnitude of f_o^{MAX} when the tendon is stretched by 3.3 %. Thus, we used 0.033 for the values of ε_o^T.

2.2.5 Simulating the Effects of Aging on Muscle Strength

To simulate the effect of aging on muscle strength, we adjusted the parameter values of the young adult actuators. Based on the results of previous studies, we adjusted the muscle deactivation time constant (τ_d) (Thelen, 2003), maximum isometric force (f_o^{MAX}) (Larsson et al., 1979), maximum muscle strain rate ($\dot{\varepsilon}_o^M$) (Larsson et al., 1978), and passive muscle stiffness (k_2^M) and maximum tendon strain (ε_o^T) (Porter et al., 1997; Gao et al., 2008). We assumed that all of the musculotendon actuators were equally influenced by aging.

We employed the same approach as the study of Thelen (2003) in determining τ_a and τ_d. Thelen assumed that aging had no effect on the rate of muscle activation, and that aging caused a 20% increase in muscle deactivation time. In the present study, the same τ_a (15 ms) was used for young and older adult actuators, but different τ_d were used for the young and older adult actuators (young adults: τ_d = 50 ms: older adults: τ_d = 60 ms). We assumed that f_o^{MAX} decreases by 25% by the seventh decade of life (Larsson et al., 1979). Thus, values of f_o^{MAX} in the older adult actuators were defined to be as 75% of f_o^{MAX} in the young adult actuators. Larsson et al. (1978) reported that the ratio of type II to type I fiber areas decreases by 30% with older age. Based on the ratio reported by Larsson et al., we adjusted the muscle fiber-type distributions from those computed for young adults and determined values of $\dot{\varepsilon}_o^M$ in the older adult actuators from the assumed unloaded shortening rates. Porter et al. (1997) reported that older adult women showed a 25% increase in passive dorsiflexion moment, and Gao et al. (2008) found that there was a 8% increase in tendon stiffness in rats. Thus, we assumed that there is an increase in the forces generated by the passive elastic (*PE*) and the series elastic (*SE*) elements with age. To achieve this, we increased the values of k_2^M in the older adult actuators by 8%, and decreased the value of ε_o^T by 8%, from the values used in the young adult actuators.

2.2.6 Tissue Joint Moment Models

We developed tissue joint moment models that simulated the passive joint moments developed by connective tissue components at the ankle, knee, and hip joints. The tissue joint moment models are used to constrain joint rotations within anatomical range of motions. Riener and Edrich (1999) developed equations for the passive elastic ankle plantarflexion, knee flexion, and hip flexion moments. We modified them and developed the following ankle plantarflexion (ANK), knee flexion (KNE), and hip flexion (HIP) tissue moment models:

$$M_{ANK} = \exp(\alpha_1 + \beta_1 \phi_{ANK}) - \exp(\alpha_2 + \beta_2 \phi_{ANK}) + \gamma_1 \tag{2.12}$$

$$\begin{aligned} M_{KNE} = {} & \exp(\alpha_1 + \beta_1 \phi_{KNE}) - \exp(\alpha_2 + \beta_2 \phi_{KNE}) + \gamma_1 \\ & + \exp(\alpha_3 + \beta_3 \phi_{KNE}) \end{aligned} \tag{2.13}$$

$$M_{HIP} = -\exp(\alpha_1 + \beta_1 \phi_{HIP}) + \gamma_1 \tag{2.14}$$

where ϕ_{ANK}, ϕ_{KNE}, and ϕ_{HIP} are ankle dorsiflexion, knee extension, and hip flexion angles in radians, respectively. At each joint, the anatomical position corresponds to an angle of zero. We determined the distinct coefficients α, β, and γ at each joint by fitting the total passive joint moment generated by the musculotendon actuators and our passive tissue moment equation to the corresponding model of Riener and Edrich (1999) at the extremes of the joint range of motion.

2.3 Results

Two sets of parameter values were found for the musculoskeletal model of young and older adults. Parameter values for the contractile (*CE*) element and for the passive elastic (*PE*) and series elastic (*SE*) elements of the 10 musculotendon actuators are presented in Table 2.4 and 2.5, respectively. The activation time constant (τ_a) for all young and older adult actuators is 15 ms, while the deactivation time constants (τ_d) for all young and older adult actuators are 50 ms and 60 ms, respectively. The maximum tendon strain (ε_o^T) is 0.033 for all young adult actuators and 0.0265 for all older adult actuators. Finally, parameter values for the ankle plantarflexion, knee flexion, and hip flexion tissue moments are presented in Table 2.6.

To provide face validity for the two-dimensional musculoskeletal model, the simulated joint moments of the young adult actuators were compared to the reference data. The peak joint moments and the moment-angle relationships developed by the young adult actuators were similar to the reference data for ankle dorsiflexion and plantarflexion moments (Figure 2.3), knee flexion and extension moments (Figure 2.4), and hip flexion and extension moments (Figure 2.5). The behaviors of the passive joint moments were consistent with the passive joint models developed by Riener and Edrich (1999) (Figures 2.8 – 2.10).

In response to step changes in *n*, the model-predicted active joint moments produced by the dorsiflexors of young adults needed a contraction time (CT) of 210 ms to reach the maximum values and a relaxation time (RT) of 260 ms to return to the relaxed state (Figure 2.6). Comparable experimental CT and RT of the dorsiflexion moment produced by twitch stimulation were 200 and 120 ms, respectively (Koh & Herzog, 1995). We found that the CT of the model was consistent with the experimental CT, and that the RT of the model was 117% slower than the experimental RT.

To validate the force-velocity relationship, we compared the simulated maximum concentric isokinetic knee extension moments of young adults at three different knee extension velocities: 36, 108, and 180 deg/sec [Figure 2.7 (a)] with the experimental data measured by Knapik et al. (1983) at a knee flexion angle of 70º during isokinetic knee extension. The simulated concentric knee extension moments showed 40% decreases as knee extension velocity increased from 36 to 108 deg/sec and from 108 to 180 deg/sec. The comparable experimental knee extension moments decreased 17% and 21%, respectively, as knee extension velocity increased from 36 to 108 deg/sec and from 108 to 180 deg/sec. Thus, the simulated knee extension showed around 20% larger decreases than the experimental data between knee extension velocities of 36 – 180 deg/sec.

The active and passive isometric ankle, knee, and hip joint moments, and the concentric and eccentric isokinetic knee extension moments developed by young and older adults are presented in Figures 2.7 – 2.10. As expected, we found that the older adult actuators generated smaller isometric and isokinetic joint moments than the young adult actuators. In contrast, the older adult actuators produced larger isometric passive joint moments than young adult actuators at the extremes of the range of motion (Figures 2.8 – 2.10). Finally, older adult actuators showed slightly slower rates of muscle deactivation than young adult actuators (Figure 2.6).

2.4 Discussion

This study focused on developing a two-dimensional musculoskeletal model of the lower extremity that can be used for forward dynamic simulations, and on determining parameter values to model the behaviors of young and older adult muscles. First, we developed the models of the 10 musculotendon actuators and the tissue joint moments. Second, we determined the parameter values of the young actuators and the coefficients of the tissue joint moment models. Third, based upon a review of the previous literature, we determined the parameter values of the older adult actuators.

2.4.1 Musculoskeletal Model

Aging causes changes in muscle structure, and older adults lose muscle strength (Larsson et al, 1978; Frontera et al., 1991, 2000) and show different movement patterns from young adults (Madigan & Lloyd, 2005). Thus, there was a need to develop musculoskeletal models of young and older adults to understand the theoretical effects of aging on different movement patterns. Another need was that these models contain a small number of musculotendon actuators, because the determination of an optimal control signal in forward dynamic simulations requires high computational expenses. We were able to develop a two-dimensional musculoskeletal model that can simulate the behaviors of young and older adult muscles successfully, while also minimizing the number of the actuators to save the computational expenses.

In designing the two-dimensional musculoskeletal model, a Hill-type muscle model was chosen. A Hill-type muscle model consists of a contractile

(*CE*) element, a passive elastic (*PE*) element, and a series elastic (*SE*) element, and parameter values. The *CE* and *PE* elements represent the active and passive force-generating properties of muscle; the *SE* element models the passive force-generating properties of tendon. The properties of the *CE*, *PE*, and *SE* elements are determined by parameter values. We could model the effects of aging on the active and passive components in muscle and tendon by adjusting parameter values.

When selecting the actuators to include in the model, it was important to include not just uniarticular but also biarticular muscle groups. In the presence of biarticular muscles, the strength at one joint depends on the position of adjacent joints and the joints moments generated at adjacent joints are not independent of each other. Based on their function, we chose the 10 actuators: gastrocnemius, soleus, dorsiflexor, hamstrings, biceps femoris, vastii, rectus femoris, gluteals, iliopsoas, and adductor magnus. We assumed that the lower extremity strength of young and older adults could be modeled by the 10 musculotendon actuators. Seven musculotendon actuators were equivalent musculotendon actuators that were composed of multiple muscles. The advantages of using equivalent musculotendon actuators were to decrease the number of the actuators and save computational expenses in determining optimal control signals. A disadvantage is that one cannot determine the contribution of individual muscles to a movement of interest. In addition to the 10 musculotendon actuators, we developed the tissue joint moment models to constrain joint rotations within anatomical ranges of motion.

We found that the young adult actuators could generate active and passive isometric joint moments that are similar to the experimental data reported in the literature (Figures 2.3 – 2.5 and 2.8 – 2.10), and confirmed that the activation-deactivation dynamics and isokinetic properties are physiologically reasonable (Figures 2.6 – 2.7). However, there is a limitation in our two-dimensional musculoskeletal model. The simulated hip extension moment was not consistent with the experimental hip extension moment with the hip flexed 90º. This may be due to muscle lines of action in the hamstrings, gluteals, and adductor magnus actuators. As the hip flexes greatly, the moment arms (r_i) of the hamstrings, gluteals, and adductor magnus actuators become small due to the straight-line approximation used for the actuator lines of action. As a result, the simulated hip extension moment could not match the experimental data with the hip flexed 90º. However, the simulated hip extension moment was consistent with the experimental hip extension moment at hip flexion of 0º through 60º. In addition, we developed the hip flexion tissue moment model. The hip flexion tissue moment model can generate passive hip joint moments that compensate for the discrepancy. Thus, the simulated total hip extension moment can support and control the upper body. Although not shown here, the simulated joint moments of young adults were compared with the musculoskeletal models developed by Delp (1990) and Hoy et al. (1990). We confirmed that the current simulated ankle, knee, and hip joint moments closely matched those simulated by the models of Delp and Hoy et al. There was also the same limitation in the hip extension moments in the models of Delp and Hoy et al.

Our musculoskeletal model can generate isometric joint moments similar to experimental data. In addition, the model shows physiologically reasonable activation-deactivation dynamics and isokinetic properties. Therefore, we believe our two-dimensional musculoskeletal model can predict the strength of the lower extremity in the sagittal plane with sufficient accuracy for use in forward dynamics simulations. Also, with the two sets of parameter values provided, the model can be used to predict the behaviors of both young adult muscles and older adult muscles.

2.4.2 Aging-Related Loss in Muscle Strength

Because of aging-related losses in muscle strength, there are differences in movement patterns between young and older adults (Madigan & Lloyd, 2005). To better understand the theoretical effects of aging on different movement patterns, we developed the two-dimensional musculoskeletal models of young and older adults. To simulate aging-related losses in muscle strength, we adjusted the parameter values of the young adult actuators. We assumed that the same age-related changes in musculotendon properties occurred in all actuators. We increased the deactivation time constant (τ_d) by 20% (Thelen, 2003), and decreased the maximum isometric active force (f_o^{MAX}) by 25% (Larsson et al., 1979). The aging-related changes in maximum muscle strain rates ($\dot{\varepsilon}_o^M$) resulted from the assumption that there was a 30% decrease in the ratio of type II fibers to type I fibers (Larsson et al., 1978). We also assumed that the forces generated by the *PE* and the *SE* elements increased by 25% per a given amount of strain in older adults (Porter et al., 1997). We therefore increased passive muscle stiffness (k_2^M)

(Gao et al., 2008) and decreased maximum tendon strain (ε_o^T) by 8%, respectively. As a result of the adjustments, the older adult actuators generated smaller isometric active joint moments and smaller concentric and eccentric isokinetic joint moments than the young adult actuators, whereas the simulated passive joint moments of older adults were larger than those of young adults. The older adult actuators showed slightly slower deactivation times than the young adult actuators. These differences were consistent with our expectation and the literature.

A potential limitation of the adjustment in the parameter values was that we did not take into account the fact that eccentric strength tends to be preserved to a greater extent than concentric strength in older adults (Alnaqeeb et al., 1984). Another possible limitation was the assumption that the parameter values of the 10 actuators were equally altered with older age. Aging could individually affect the capacity to produce the muscle and tendon forces in each muscle. For example, Cahalan et al. (1989) reported no significant differences between the hip joint moments developed by young and older adults. In determining the parameter values of the older adult actuators, we employed the results of multiple studies of the effects of aging on muscle. Thus, we could not adjust the parameter values of each musculotendon actuator differently. As a result, the current musculoskeletal model of older adults simulates general aging-related losses in lower extremity strength rather than muscle-specific losses in strength.

Nevertheless, as was stated, the effects of aging shown in Figures 2.6 – 2.10 are consistent with the aging-related changes described in the literature. Thus, we believe that the current musculoskeletal model of older adults can be validly used to investigate the effects of aging-related losses in lower extremity strength on movement patterns of older adults.

2.4.3 Conclusion

A musculoskeletal model can simulate muscle and tendon forces and is used to simulate human motion in forward dynamic simulations. The disadvantage of forward dynamic simulation is the computational expense to determine optimal control signals. The cost is proportional to the number of musculotendon actuators. Thus, we minimized the number of musculotendon actuators and designed a two-dimensional musculoskeletal model that was composed of 10 musculotendon actuators together with ankle plantarflexion, knee flexion, and hip flexion tissue moment models. The two-dimensional musculoskeletal model produces a reasonable approximation to the strength characteristics of the lower extremity. The particular advantage of our musculoskeletal model is the lower cost to find an optimal control signal. For example, Anderson and Pandy (2001) developed a three-dimensional musculoskeletal model that consisted of 54 musculotendon actuators and required 54 control signals. In contrast, the 10 musculotendon actuators can be governed by six control signals. A single biarticular muscle or two antagonist uniarticular muscles can be controlled by one control signal. In addition, the gluteals and adductor magnus can be controlled as a single uniarticular muscle. Thus, six control signals are sufficient to govern seven uniarticular muscles (dorsiflexor – soleus, biceps femoris – vastii, iliopsoas – gluteals/adductor magnus) and three biarticular muscles (gastrocnemius, rectus femoris, and hamstrings).

This study also supplies the parameter values to predict the behaviors of young and older adult muscles. Therefore, the two-dimensional musculoskeletal model and the parameter values can be used with forward dynamic simulations to simulate the motion of young and older adults and investigate the effects of the aging on these motions.

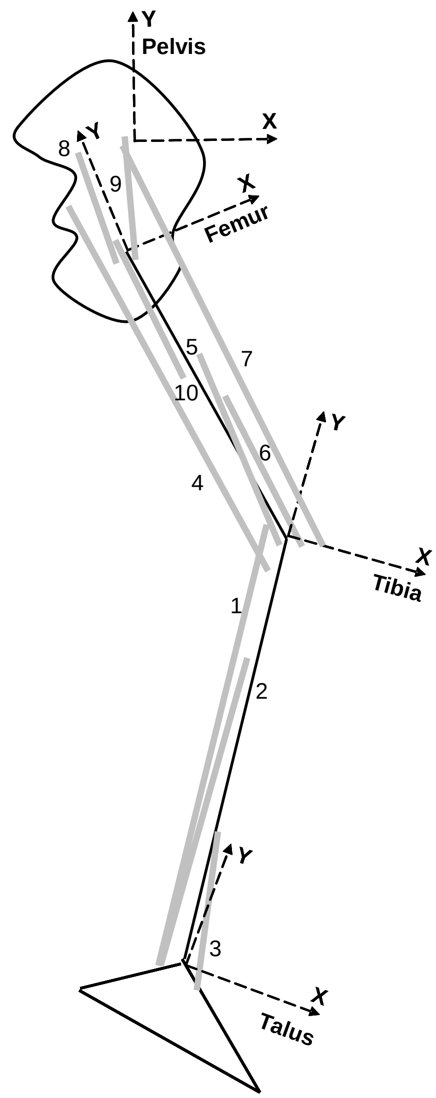


Figure 2.1. Ten musculotendon actuators were developed within the talus, tibia, femur, and pelvis reference frames (1: gastrocnemius, 2: soleus, 3: dorsiflexor, 4: hamstrings, 5: biceps femoris, 6: vastii, 7: rectus femoris, 8: gluteals, 9: iliopsoas, 10: adductor magnus).

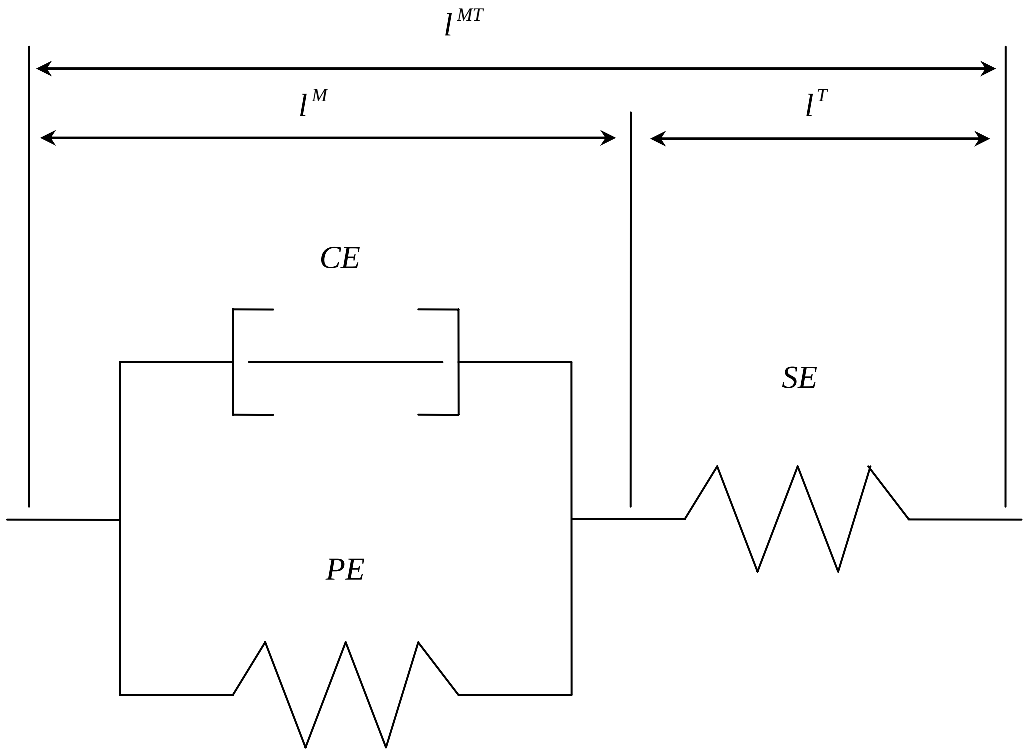


Figure 2.2. The contractile (*CE*) element, passive elastic (*PE*) element, and series elastic (*SE*) element that comprise the musculostendon model. The *CE* and *PE* represent the active and passive force-generating properties of muscle, respectively; the *SE* element represents the passive force-generating properties of tendon. The l^{MT} represents the musculotendon actuator length which consists of the muscle length (l^{M}) and the tendon length (l^{T}).

Figure 2.3. Reference data and predictions of the young adult model for the maximum isometric (a) ankle dorsiflexion moment and (b) ankle plantarflexion moment, as a function of ankle angle. The ankle dorsiflexion moments in (a) consist of only an active moment with the knee flexed 90º; reference data are from Koh and Herzog (1995). The squares and solid line represent the reference data and the model-predicted total dorsiflexion moment, respectively. The ankle plantarflexion moments in (b) consist of the summed active and passive moments with the knee flexed 0°; reference data are from Bobbert and Van Ingen Schenau (1990). The squares, solid line, dashed line, and dotted line represent the reference data and the model-predicted total plantarflexion moment, moment developed by the soleus (SOL), and moment by the gastrocnemius (GAS), respectively. Bw = body weight; Bh = body height.

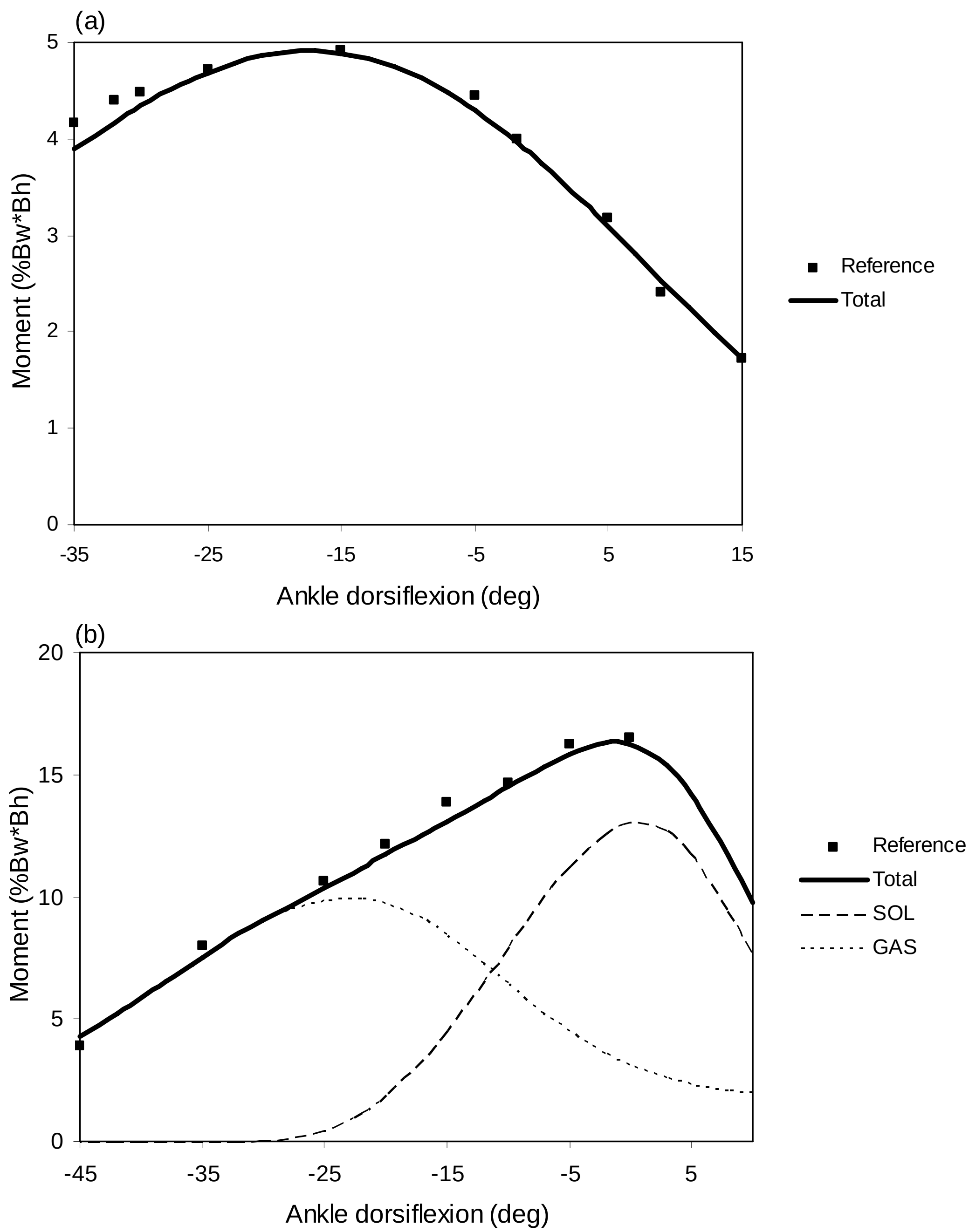
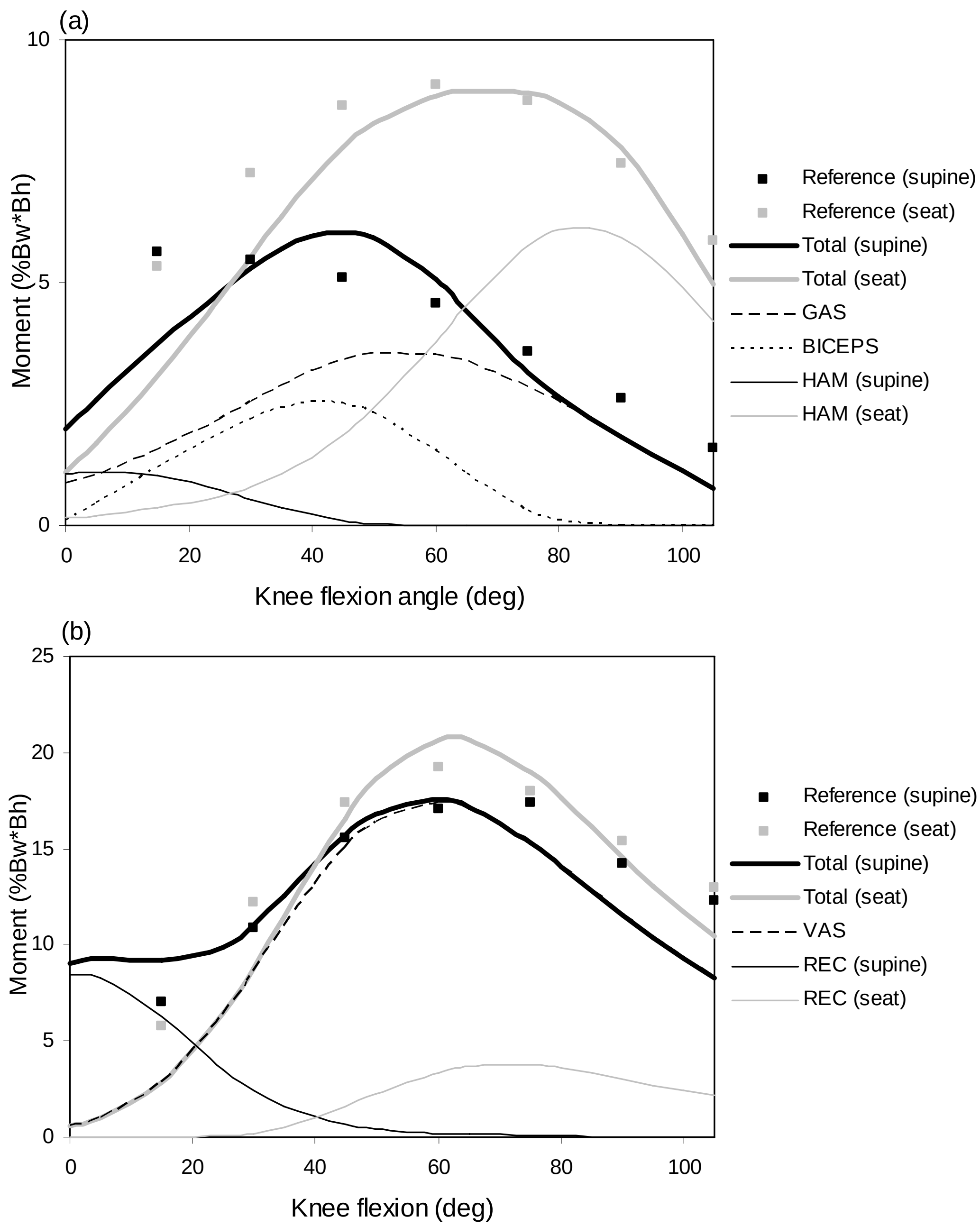

(a)
Moment (%Bw*Bh)
0
1
2
3
4
5
-35
-25
-15
-5
5
15
Ankle dorsiflexion (deg)
Reference
Total
(b)
Moment (%Bw*Bh)
0
5
10
15
20
-45
-35
-25
-15
-5
5
Ankle dorsiflexion (deg)
Reference
Total
SOL
GAS

Figure 2.3.

Figure 2.4. Reference data and predictions of the young adult model for the maximum isometric (a) knee flexion moment and (b) knee extension moment, as a function of knee angle. The isometric knee flexion and extension moments in supine (black) and seated (gray) positions provided by Houtz et al. (1958) and Knapik et al. (1983) were used for the angle-dependent behavior and magnitude, respectively, of the reference. The knee flexion and extension moments consist of only an active moment. (a) The thick solid, dashed, dotted, and thin solid lines and squares represent the total knee flexion moment, the moment developed by the gastrocnemius (GAS), the moment by the biceps femoris (BICEPS), the moment by the hamstrings (HAM), and the reference data, respectively. (b) The thick solid, dashed, and thin solid lines and squares represent the total knee extension moment, the moment developed by the vastii (VAS), the moment by the rectus femoris (REC), and the reference data, respectively. Bw = body weight; Bh = body height.



(a)
Moment (%Bw*Bh)
Knee flexion angle (deg)
Reference (supine)
Reference (seat)
Total (supine)
Total (seat)
GAS
BICEPS
HAM (supine)
HAM (seat)
(b)
Moment (%Bw*Bh)
Knee flexion (deg)
Reference (supine)
Reference (seat)
Total (supine)
Total (seat)
VAS
REC (supine)
REC (seat)

Figure 2.4.

Figure 2.5. Reference data and predictions of the young adult model for the maximum isometric (a) hip flexion moment and (b) hip extension moment, as a function of hip angle. (a) The isometric hip flexion moment with the knee and hip flexed by the same angles provided by Williams and Stutzman (1959) and Cahalan et al. (1989) was used as the behavior and magnitude, respectively, of the reference. The hip flexion moment consisted of only an active moment. The thick solid, dashed, dotted lines and squares represent the total hip flexion moment, the moment developed by the rectus femoris (REC), the moment by the iliopsoas (ILIO), and the reference data. (b) The hip extension moment with the knee flexed 30° provided by Nemeth et al. (1983) was used as reference data. The hip extension moment consisted of only an active moment. The thick solid, dashed, dotted, thin solid lines and squares represent the total hip extension moment, the moment developed by the hamstrings (HAM), the moment by the gluteals (GLU), the moment by the adductor magnus (MAG), and the reference data. Bw = body weight; Bh = body height.

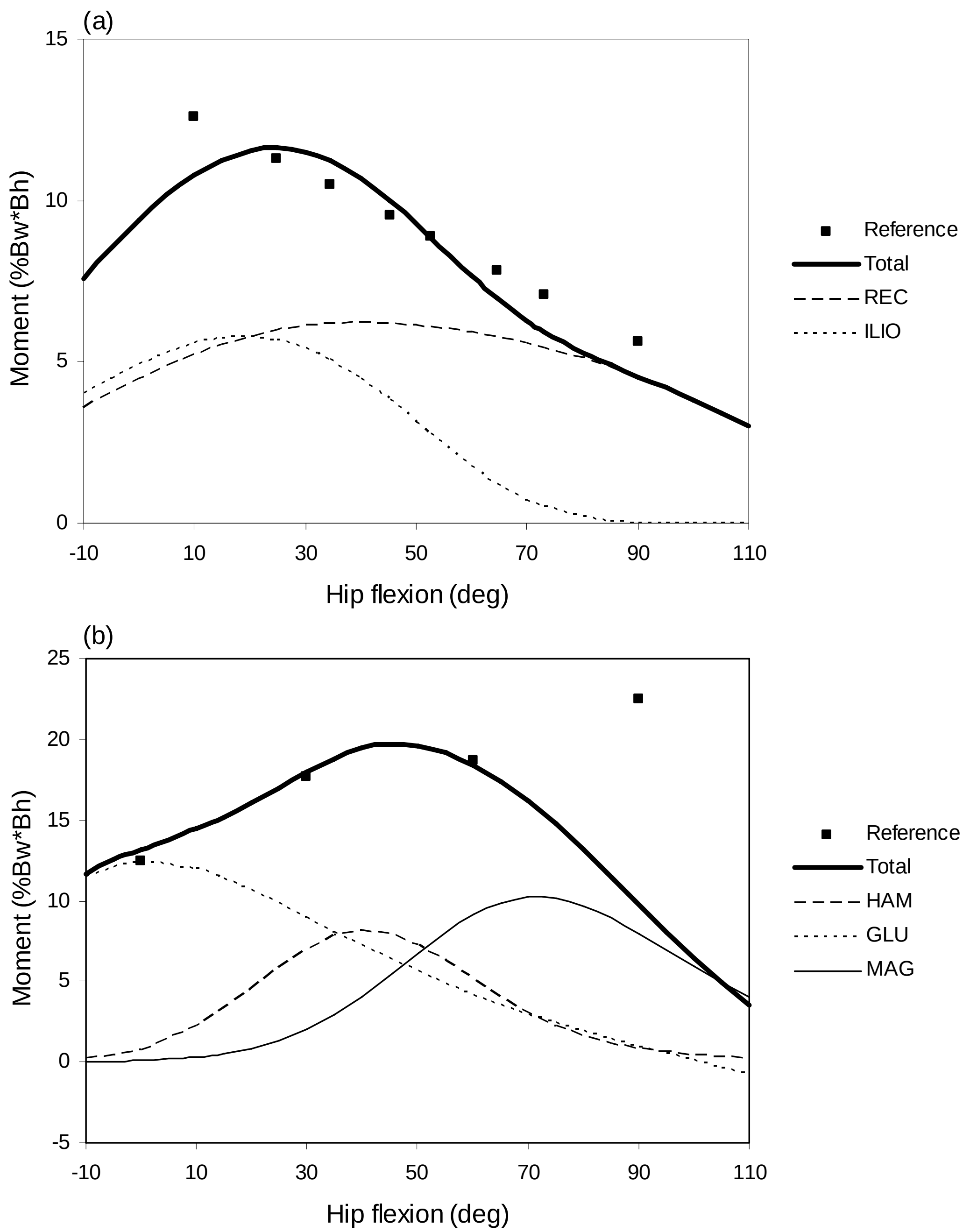

(a)
15
10
5
0
-10
10
30
50
70
90
110
Moment (%Bw*Bh)
Hip flexion (deg)
Reference
Total
REC
ILIO
(b)
25
20
15
10
5
0
-5
-10
10
30
50
70
90
110
Moment (%Bw*Bh)
Hip flexion (deg)
Reference
Total
HAM
GLU
MAG

Figure 2.5.

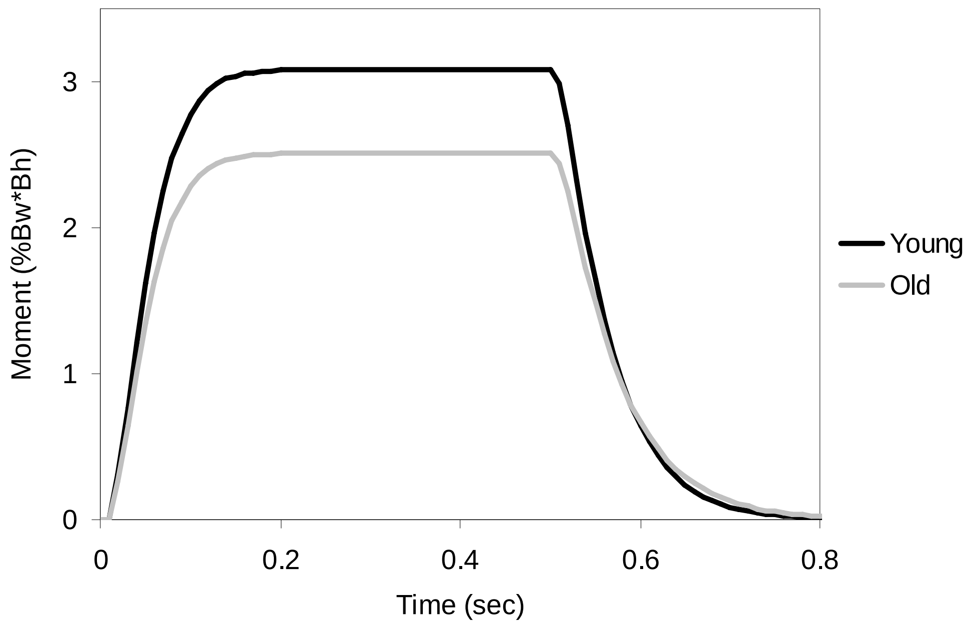


Figure 2.6. Muscle activation and deactivation dynamics of the isometric active dorsiflexion moment by the young (black) and old (gray) dorsiflexor actuator. The neural excitation signal was altered from 0 (none) to 1 (maximum) at t = 0.01 s and from 1 to 0 at t = 0.5 s. The ankle joint angle was held constant at 5º of dorsiflexion. Bw = body weight; Bh = body height.

Figure 2.7. Maximum model-predicted (a) concentric and (b) eccentric isokinetic knee extension moments over the range of 0-90 degrees of knee flexion, simulated with the hip flexed 90º. The isokinetic moments developed by young (black) and older (gray) adults are shown for three different knee extension or flexion velocities. The thick solid, thin solid, and dashed lines represent 36, 108, and 180 deg/sec, respectively. In the corresponding simulations, the knee extensors were maximally activated throughout and were initialized to their isometric steady-state. Bw = body weight; Bh = body height.

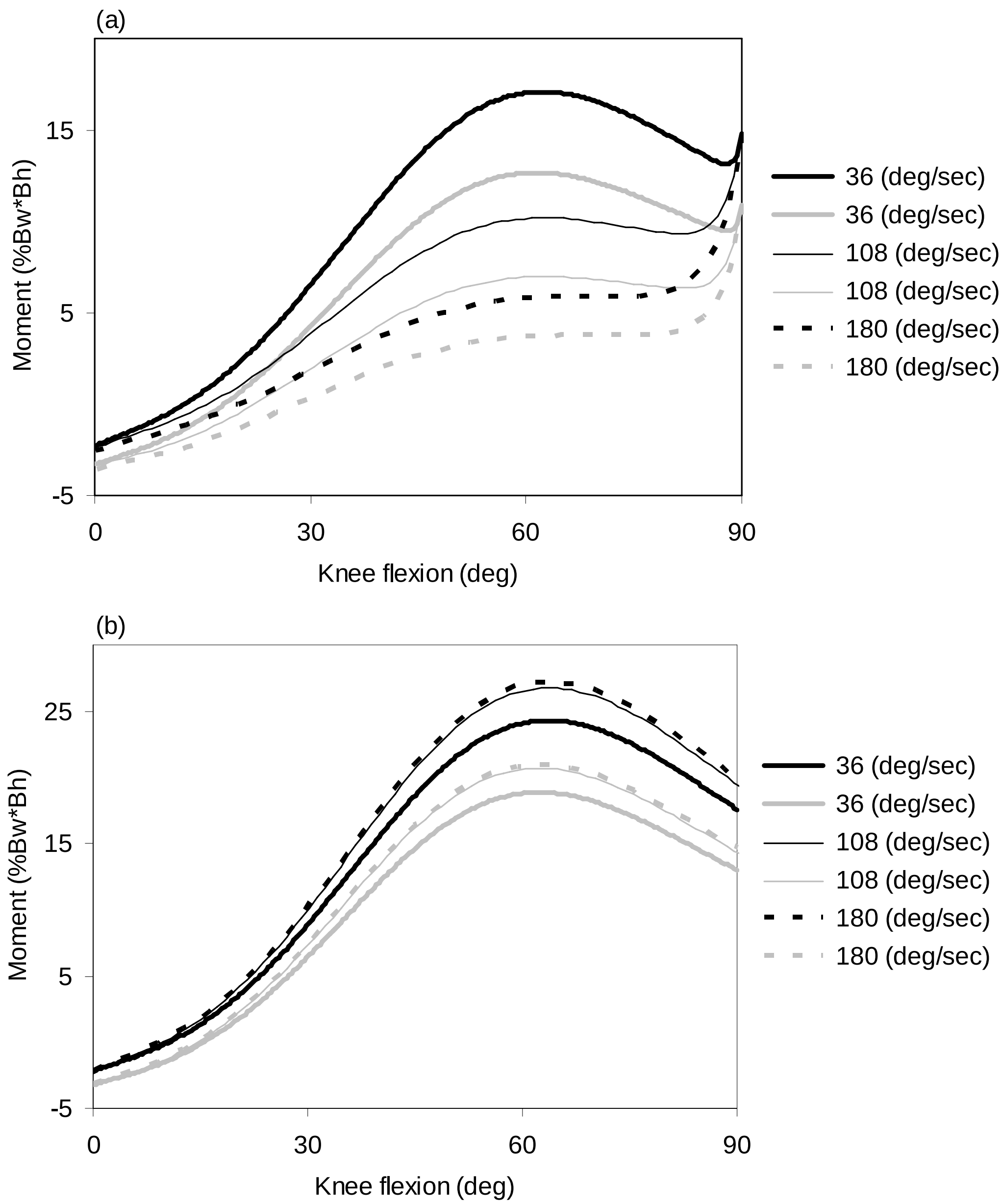

(a)
15
5
-5
0
30
60
90
Moment (%Bw*Bh)
Knee flexion (deg)
36 (deg/sec)
36 (deg/sec)
108 (deg/sec)
108 (deg/sec)
180 (deg/sec)
180 (deg/sec)
(b)
25
15
5
-5
0
30
60
90
Moment (%Bw*Bh)
Knee flexion (deg)
36 (deg/sec)
36 (deg/sec)
108 (deg/sec)
108 (deg/sec)
180 (deg/sec)
180 (deg/sec)

Figure 2.7.

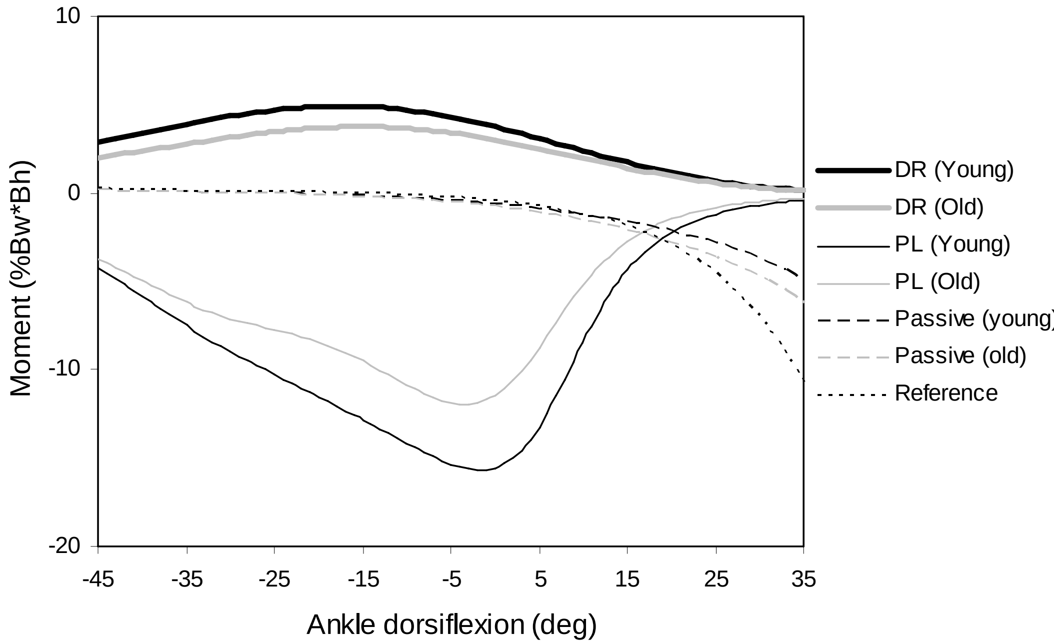


Figure 2.8. Ankle dorsiflexion (+) and plantarflexion (-) moments with the knee flexed 0° developed under isometric conditions by the models of young (black) and older (gray) adults. The thick solid, thin solid, dashed, and dotted lines represent maximum active dorsiflexion (DR) moments, maximum active plantarflexion (PL) moments, passive moments, and reference data, respectively. The reference data are from the passive elastic ankle joint model developed by Riener and Edrich (1999). Bw = body weight; Bh = body height.

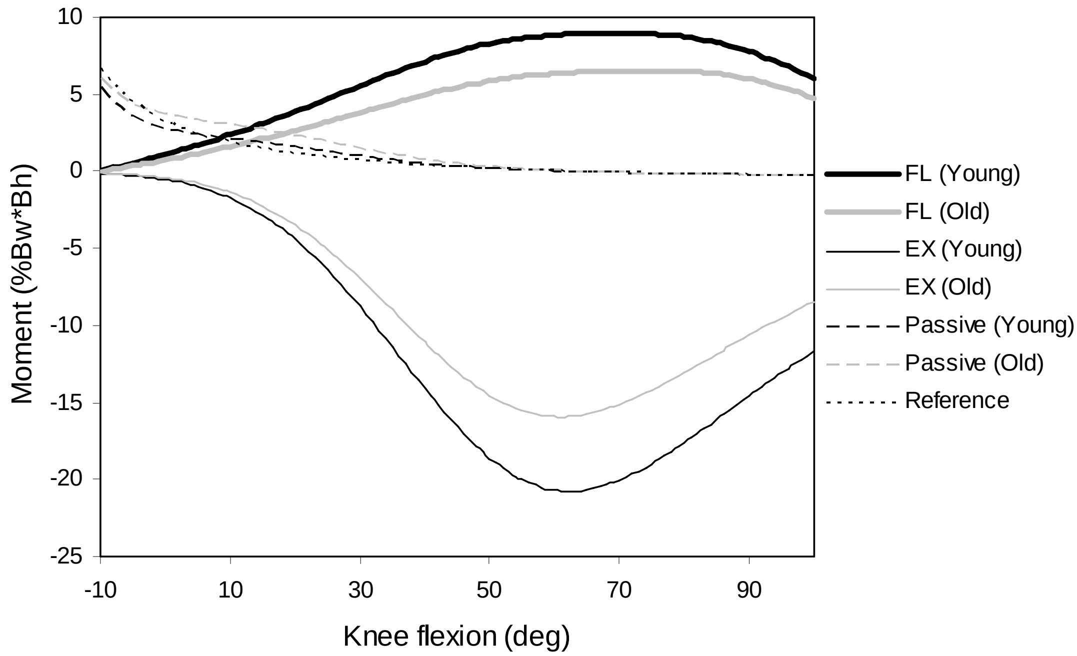


Figure 2.9. Knee flexion (+) and extension (-) moments with the hip flexed 90° developed under isometric conditions by the models of young (black) and older (gray) adults. The thick solid, thin solid, dashed, and dotted lines represent maximum active knee flexion (FL) moments, maximum active extension (EX) moments, passive moments, and reference data, respectively. The reference data are from the passive elastic knee joint model developed by Riener and Edrich (1999). Bw = body weight; Bh = body height.

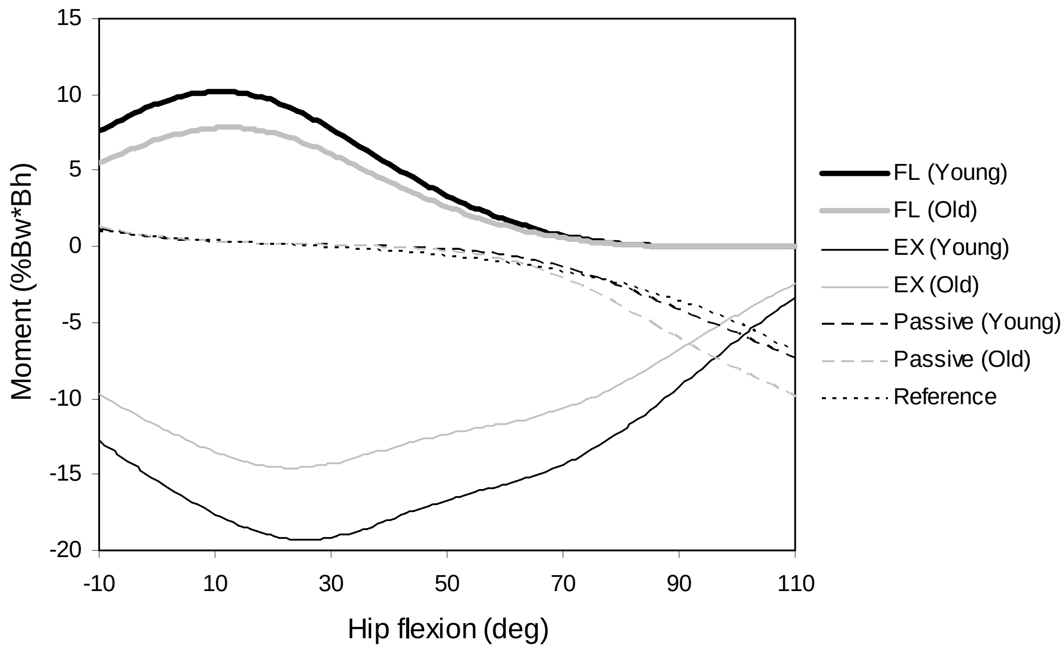


Figure 2.10. Hip flexion (+) and extension (-) moments with the knee flexed 0° developed under isometric conditions by the models of young (black) and older (gray) adults. The thick solid, thin solid, dashed, and dotted lines represent the maximum active hip flexion (FL) moments, maximum active extension (EX) moments, passive moments, and reference data, respectively. The reference data are from the passive elastic hip joint model developed by Riener and Edrich (1999). Bw = body weight; Bh = body height.

Table 2.1. Musculotendon actuators

Actuator	Component Muscles
Gastrocnemius	Gastrocnemius lateralis, Gastrocnemius medialis
Soleus	Soleus
Dorsiflexor	Tibialis anterior, Extensor digitorum longus, Extensor hallucis longus
Hamstrings	Biceps femoris long head, Semitendinosus, Semimembranosus
Biceps femoris	Biceps femoris short head
Vastii	Vastus lateralis, Vastus medialis, Vastus intermedius
Rectus Femoris	Rectus femoris
Gluteals	Gluteus minimus, Gluteus medius, Gluteus maximus
Iliopsoas	Iliacus, Psoas major
Adductor Magnus	Adductor magnus inferior, Adductor magnus middle, Adductor magnus inferior

Table 2.2. Geometry data of the musculotendon actuators

Actuator	Constant-Length Component (m)	Variable-Length Component Endpoints	
		Frame	(X,Y) Coordinates (m)
Gastrocnemius	0.0108	Femur	(-0.0236, -0.397)
		Tibia	(-0.0283, -0.0985)
		Talus	(-0.0444, -0.011)
Soleus	0.	Tibia	(-0.0024, -0.1533)
		Talus	(-0.0444,-.01095)
Dorsiflexor	0.3023	Tibia	(0.0317, -0.3972)
		Talus	(0.0579, -0.015)
Hamstrings	0.009	Pelvis	(-0.1216, -0.102)
		Tibia	(-0.0203, -0.06)
Biceps femoris	0.	Femur	(0.005, -0.2111)
		Tibia	(-0.0101, -0.0725)
Vastii	0.0552	Femur	(0.0316, -0.2478)
		Tibia	(0.0296, 0.029)
Rectus femoris	0.	Pelvis	(-0.0295, -0.0311)
		Tibia	(0.0534, 0.0316)
Gluteals	0.0724	Pelvis	(-0.1166, -0.0342)
		Femur	(-0.0279, -0.0272)
Iliopsoas	0.1447	Pelvis	(-0.0226, -0.0558)
		Femur	(0.0017, -0.053)
Adductor magnus	0.	Pelvis	(-0.0772, -0.1181)
		Femur	(0.0015, -0.2223)

Each musculotendon actuator consists of a constant-length component that passes no joints plus a variable-length component (or components) that acts in a straight line between the endpoints listed and spans one or more joints.

Table 2.3. Musculotendon actuator parameters

Parameter	Definition
τ_a	Activation time constant
τ_d	Deactivation time constant
f_o^{MAX}	Maximum isometric active force
l_o^M	Optimal muscle length
i_a	Muscle index of architecture
$\dot{\varepsilon}_o^M$	Maximum muscle strain rate
k_1^M	Passive muscle force gain
k_2^M	Passive muscle stiffness
l_o^T	Tendon slack length
$\widetilde{k}^T$	Normalized tendon stiffness
ε_o^T	Maximum tendon strain

Table 2.4. Parameter values for the contractile (*CE*) element of young adult and older adult actuators.

Actuator	f_o^{MAX} (N)		l_o^M (m)	$\dot{\varepsilon}_o^M$ (l_o^M/s)		i_a
	Young	Old		Young	Old	
Dorsiflexor	1683	1262	0.0901	3.482	3.129	0.787
Soleus	4115	3086	0.0276	2.677	2.492	1
Gastrocnemius	2904	2178	0.047	4.849	4.361	1
Biceps femoris	1285	964	0.1531	3.821	3.415	0.453
Hamstrings	1764	1323	0.0838	3.821	3.415	1
Vastii	8144	6108	0.0816	4.889	4.4	1
Rectus femoris	2016	1512	0.0745	5.405	4.927	1
Gluteals	4911	3683	0.1308	4.618	4.138	0.672
Adductor magnus	2600	1950	0.1045	4.288	3.997	0.932
Illiopsoas	2077	1558	0.1047	4.794	4.307	1

f_o^{MAX} = maximum isometric active force; l_o^M = optimal muscle length; i_a = muscle index of architecture; $\dot{\varepsilon}_o^M$ = maximum muscle strain rate. Values of l_o^M and i_a apply to both young and old actuators.

Table 2.5. Parameter values for the parallel elastic (*PE*) and series elastic (SE) elements of young adult and older adult actuators.

Actuator	k_1^M (N)	k_2^M (m^{-2})		l_o^T (m)	$\widetilde{k}^T$ (kN)	
		Young	Old		Young	Old
Dorsiflexor	14.142	10.439	11.274	0.2738	1546	1798
Soleus	10.784	4.297	4.64	0.2557	3779	4395
Gastrocnemius	12.747	4.479	4.837	0.3719	2667	3101
Biceps femoris	51.338	46.678	50.412	0.0984	1180	1373
Hamstrings	0.176	12.407	13.4	0.3607	1620	1884
Vastii	0.075	0.147	0.159	0.1485	7478	8698
Rectus femoris	7.804	2.982	3.22	0.3284	1851	2153
Gluteals	9.518	18.78	20.283	0.0092	4510	5245
Adductor magnus	14.935	12.359	13.347	0.1109	2387	2777
Illiopsoas	42.922	21.245	22.945	0.1109	1907	2218

PE parameters: k_1^M = passive muscle force gain; k_2^M = passive muscle stiffness.
SE parameters: l_o^T = tendon slack length; $\widetilde{k}^T$ = normalized tendon stiffness.
Values of k_1^M and l_o^T apply to both young and old actuators.

Table 2.6. Coefficients for the ankle plantarfexion (ANK), knee flexion (KNE), and hip flexion (HIP) tissue moment models

	α_1 (N·m)	α_2 (N·m)	α_3 (N·m)	β_1 (1/rad)	β_2 (1/rad)	β_3 (1/rad)	γ_1 (N·m)
M_{ANK}	- 3.1231	- 9.9518		9.906	-12.805		- 0.7762
M_{KNE}	1.8937	-4.211	2.5085	0.643	-2.587	8.384	- 4.811
M_{HIP}	-3.5895			3.6			2.0476

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The effects of aging-related loss in muscle strength in the lower extremity on the feasible region for balance recovery

Norio Kadono and Michael J Pavol

3.1 Introduction

Falls can have severe impacts on an older adult's life, as well as on society. In 2000, 10,000 people aged 65 years or older died from falls, and 2.6 million older adults received medical treatments for fall-related injuries (Stevens et al., 2006). Approximately $19.2 billion were spent in 2000 for fatal and non-fatal fall-related injuries (Stevens et al., 2006). One injury of major concern is hip fracture. Nevitt et al. (1993) reported that 1% of older women who experienced a fall suffered a hip fracture. However, older adults who experienced hip fractures showed a 10% higher mortality rate than those who had no hip fractures (Leibson et al., 2002). Hip fractures also lead to limitations in activities of daily living and in lower-body function among older adults (Wolinsky et al., 1997). A hip fracture is most likely to result from a backward or sideways fall (Smeesters et al., 2001), of which falls from a backward balance loss are the hardest to prevent (Hsiao & Robinovitch, 2001). Prevention of backward falls by older adults is thus a critical subject to be studied.

Declines in balance and in the ability to prevent a backward fall may be associated with aging-related losses in muscle strength. As age advances, the structure of muscle undergoes alterations (Larsson et al., 1978; Sato et al., 1984; Frontera et al., 1991; Frontera et al., 2000; Thompson, 2002). Older adults lose 20 to 60% of peak muscle torque and power and muscle shortening velocity that they had at young ages (Thom et al., 2007). Daubney and Culham (1999) reported that ankle strength was significantly associated with balance among older adults and that non-fallers showed greater ankle and hip strength than fallers. Orr et al.

(2006) reported that healthy older adults who underwent 10 weeks of knee power training increased their knee strength and improved balance. Thus, muscle strength in the lower extremity could be one of the factors contributing to balance and the ability to prevent a backward fall. In particular, aging-related losses in muscle strength might influence the ability to support the weight of the body and stop its downward motion during the stance phase that follows a recovery step. Madigan and Lloyd (2005) studied the patterns of ankle, knee, and hip moment development by young and older adults during the stance phase following a recovery step from a forward balance loss. They found that the older adults showed different patterns of joint moments than the young adults and that older adults relied more on ankle and hip strength than knee strength to compensate for aging-related losses in muscle strength. In contrast to the findings of these studies, however, Pavol et al. (2002) reported that older adults who fell following an induced trip showed greater ankle and knee strength and faster walking speed than non-fallers. Thus, it still remains unknown to what extent the aging-related loss in muscle strength affects the ability to recover balance during the stance phase following a recovery step from a backward balance loss.

One approach to answer the question is to describe a feasible region of balance recovery (Pai & Patton, 1997; Pai & Iqbal, 1999). Pai and Patton (1997) defined the feasible region as the ranges of initial positions and velocities of the body center of mass (COM) for which the COM can be brought to rest above the base of support (BOS) without taking a step. The feasible region has been computed using musculoskeletal models and used to predict the ranges of successful dynamic balance recovery without a step from a balance loss in a standing position (Pai & Patton, 1997). A musculoskeletal model can model the muscle and tendon force-generating characteristics of young and older adults. Thus, dynamic simulation can be used to determine whether the aging-related loss in muscle strength decreases the ability to prevent a fall by decreasing the feasible region of balance recovery.

The feasible region has been used in this way to examine the effects of aging-related losses in muscle strength on balance control (Pai & Patton, 1997). Pai and Patton (1997) described the feasible region in which a person can maintain balance in a standing position without taking any steps. They found that normal aging-related losses in ankle muscle strength had no effects on the feasible region. However, a person who loses balance typically takes a step to regain balance, and can exhibit considerable hip and knee flexion at step touchdown (Pavol & Pai, 2007). The model of Pai and Patton could not take this hip and knee motion into account. Thus, it still remains unknown what effect the aging-related loss in muscle strength will have on the ability to bring the COM to rest above the BOS during the stance phase following a recovery step from a backward balance loss.

In this study, we will simulate balance recovery motions of young and older adults following a backward step after a backward balance loss and will describe the feasible regions of balance recovery. The effects of both the backward motion of the COM and the downward motion of the hips on the ability to regain balance will be taken into account. This study will thus use a modified version of the feasible region in which the initial COM horizontal positions

(X_{COM}) and hip heights (Z_{HIP}) will be determined from which a person can bring his or her COM to rest in a stable position for a given initial backward and downward velocity. By modeling the behaviors of young and older adult muscles, we will compare the feasible regions of young and older adults and determine the extent to which the aging-related loss in muscle strength influences the ability to regain balance using a backward recovery step.

3.2 Methods

A mathematical modeling approach was used to investigate the effects of aging-related losses in lower extremity strength on the ability to recover balance following a backward step after a backward balance loss. A sagittal-plane model was developed that could simulate the motion of the body during the stance phase after a backward step. Two sets of parameter values were determined that approximated the properties of the lower-extremity muscles of young and older adults, respectively. The two-dimensional model and parameter values were used to model the backward balance recovery motions of young and older adults.

3.2.1 Six-Link Model Formulation

A six-link model was developed to simulate the recovery motions of young and older adults in the sagittal plane after a backward step (Figure 3.1). We assumed that the rear limb (i.e. the limb that took the step) in the model is actively controlled to terminate the motion of the body, while the front limb provides passive weight support. The model thus had the following design. The links of the model represented the front and rear feet, rear leg, rear thigh, head-arms-torso, and front thigh-and-leg. The lengths, masses, and moments of inertia of the segments and the locations of their COMs were determined based on the data provided by Winter (2005) for an individual with a body height of 1.8 m and body mass of 75 kg. The ankle, knee, and hip joints were modeled as frictionless hinge joints in the sagittal plane. The rear foot was fixed to the ground. The front foot was constrained to slide along the ground. The coefficient of friction beneath the front foot was modeled as a sigmoid function of velocity (van den Bogert et al., 1989); the friction coefficient reached 90% of its maximum value of 0.9 when the velocity of the front foot attained 0.05 m/s, and 95% of its maximum value when the velocity attained 0.1 m/s.

A two-dimensional musculoskeletal model was developed to actively control the rear limb. The musculoskeletal model was described in the first study, thus will only be briefly explained in this study. The rear limb was actively controlled through a set of 10 musculotendon actuators acting across the ankle, knee, and hip (Figure 3.2). The 10 musculotendon actuators consisted of three single-muscle and seven equivalent-muscle actuators. The single-muscle actuators consisted of soleus, rectus femoris, and biceps femoris (short head); the equivalent-muscle actuators that represented multiple muscles consisted of gastrocnemius, dorsiflexor, hamstrings, vastii, gluteals, iliopsoas, and adductor magnus. Musculoskeletal geometry was computed using the data provided by Delp (1990). The force-generating properties of the actuators were simulated using Hill-type musculotendon models. The musculotendon models consisted of contractile (*CE*) and passive elastic (*PE*) elements that represented the active and passive components of muscle force generation and a series elastic (*SE*) element that represented the passive behavior of tendon force (Audu & Davy, 1985; Zajac, 1989).

The muscle force generated by the *CE* element (F^{CE}) depends on its time-dependent activation level [$a(t)$], its length-tension relationship [$f^{LT}(i_a, \varepsilon^M)$], its force-velocity relationship [$f^{FV}(\dot{\eta})$], and its maximum isometric force [f_o^{MAX}]. A dimensionless neural excitation signal (n), ranging between 0 (no excitation) and 1 (maximum excitation), was used to determine $a(t)$. The relationship between n and $a(t)$ was modeled by a non-linear, first-order differential equation (Zajac et al, 1986; Thelen, 2003). $f^{LT}(i_a, \varepsilon^M)$ represents the relationship between the length of the muscle (l^M) and its force output. The *CE* element is able to produce the maximum isometric force when the muscle contracts at its optimal muscle length (l_o^M), whereas the *CE* element generates less force as l^M becomes greater or smaller than l_o^M. The changes in force output with changes in l^M depend on the muscle index of architecture (i_a). $f^{FV}(\dot{\eta})$ represents the relationship between F^{CE} and the velocity of isotonic muscle contraction. As the muscle concentric contraction-velocity increases to its maximum strain rate ($\dot{\varepsilon}_o^M$), F^{CE} decreases progressively to zero; as the muscle eccentric contraction velocity increases, F^{CE} increases to approximately 1.33 times its isometric value (Kaufman et al., 1991). Finally, F^{CE} is proportional to f_o^{MAX}.

The *PE* and *SE* elements are assumed to behave as nonlinear springs. F^{PE} and F^{SE} represent the passive forces generated by the *PE* and *SE* element, respectively. F^{PE} is determined by muscle strain (ε^M), the passive muscle force gain (k_1^M), and the passive muscle stiffness (k_2^M); F^{SE} is determined by tendon strain (ε^T) and normalized tendon stiffness ($\widetilde{k}^T$).

The characteristics of the *CE*, *PE*, and *SE* elements of the musculotendon actuators were determined by their associated parameter values. First, the parameter values to model the behavior of young adult muscles were determined by reviewing previous experimental data and studies of musculoskeletal models and using an optimization method to fit simulated isometric joint moments to corresponding experimental data. Second, the behavior of older adult muscles was simulated by adjusting the young adult parameter values that governed muscle deactivation, maximum isometric force, muscle force-velocity behavior, and passive force developed by the *PE* and *SE* elements. For each actuator, we increased the deactivation time constant by 20%, decreased the maximum isometric force by 25%, decreased the ratio of type II to type I fiber areas by 30% and re-calculated maximum muscle strain rate, and increased the passive muscle stiffness and tendon stiffness by 8% relative to the values for young adults.

Passive elastic joint moments were used to enforce the anatomical range of motion at each joint (Audu & Davy, 1985; Riener & Edrich, 1999). The front limb of the six-link model moved passively under the influence of external forces and the passive elastic joint moments at the ankle and hip.

3.2.2 Equations of Motion & Neural Excitation

The six-link model was implemented to allow dynamic simulations to be performed. The equations of motion of the six-link model were derived using the Lagrangian algorithm. The basic equations of motion were of the form:

$$J(\vec{\theta}_i)\ddot{\vec{\theta}} = R(\vec{\theta}_i, \dot{\vec{\theta}}_i) + \vec{M} \tag{3.1}$$

where J is the moment of inertia matrix, $\vec{\theta}_i$ are the angular positions of the segments, $\dot{\vec{\theta}}_i$ are the angular velocities of the segments, $\ddot{\vec{\theta}}_i$ are the angular accelerations of the segments, R is the gravitational and Coriolis effects matrix, and $\vec{M}$ is the joint moment vector. The equations of the motion of the model were implemented in C++. The state of the model was represented by a vector *q*:

$$q = [\theta_i, \dot{\theta}_i, a_j, f_k] \tag{3.2}$$

where a_j was a muscle activation and f_k was F^{SE}. A fourth-order Runge-Kutta method with adaptive step sizes was used to integrate the vector *q* during forward dynamic simulations.

During the dynamic simulations of the motion of the six-link model, the model was controlled by neural excitation signals to the 10 musculotendon actuators that generated the dynamic motion of the rear limb of the model. Each musculotendon actuator was given a dimensionless neural excitation signal (n_l) ranging between 0 and 1 as an input. However, the neural excitation signals to the 10 musclotendon actuators were generated by only six controllers. Each biarticular actuator (i.e. gastrocnemius, rectus femoris, hamstrings) was controlled individually. Antagonist uniarticular actuators were controlled by a single control signal, such that only one of the two actuators received the neural excitation at any given time. That is, the soleus and dorsiflexors shared the same controller, as did the vastii and biceps femoris, as well as the gluteals and iliopsoas. In addition, to decrease the number of control signals, adductor magnus received the same control signal as gluteals. The control signals to these actuator pairs ranged from -1 to 1. Positive and negative control signals represented the neural excitation to the extensor and flexor actuators, respectively, with the excitation of the antagonist set to zero.

The n_l were parameterized control signals. Each control signal was parameterized by the start time, magnitude, and stop time of periods of constant excitation. The periods of the constant neural excitation were connected by linear changes in excitation.

3.2.3 Determining the Feasible Region for Balance Recovery

The feasible regions for balance recovery of the six-link model were determined with the rear limb controlled actively and the front limb controlled passively. The feasible region was defined as the range of initial COM horizontal positions (X_{COM}) and initial hip heights (Z_{HIP}) from which the model could successfully be brought to rest for a given initial horizontal and vertical velocity of the COM and hips, respectively. This region was mapped using repeated optimizations. In each optimization, the initial horizontal velocity of the COM and vertical velocity of the hips was specified and the initial COM horizontal position or hip height was chosen. Then, either the maximum or minimum value of the other variable was found for which the model could be brought to rest. The Simulated Annealing algorithm was used for optimization in this study (Corana et al., 1987).

The control variables in the optimization included the initial angles and initial angular velocities of the rear leg (θ_1), rear thigh (θ_2), and head-arms-torso (θ_3), the initial muscle activation (a_j) in each actuator, the initial F^{SE} (f_k) in each actuator, and the parameters of the neural excitation signals (n_l). There were constraints on the initial conditions (Table 3.1). The cost function to be minimized was expressed in the following form:

$$I_{COST} = kf_0 + \sum g_n + \sum h_n \tag{3.3}$$

where the f_0 was an initial X_{COM} or Z_{HIP}, the k was +1 to minimize f_0 and -1 to maximize f_0, the g_n were penalty values associated with violating the penalty functions of the continuous state during the movement, and the h_n were penalty values associated with violating the penalty functions of the final state. The penalty functions of the continuous state during the movement required the six-link model to perform the body-stabilizing motion under conditions in which young and older adults could conduct them in real life (Table 3.2). The penalty functions of the final state required the six-link model to bring itself to a statically stable, near-stationary state at the end of the simulation (Table 3.3). The duration of each simulation used to establish the feasible region was 1 second.

We assumed that each parameterized control signal included five periods of constant neural excitation within the 1-second duration of the simulation, with the first period of constant excitation starting at time 0 and with the last such period ending at 1 second. To aid in the convergence of the optimizations, a 4-stage optimization procedure was used. The duration of the simulation and the complexity of the control signal (i.e. number of periods of constant neural excitation) progressively increased from stage to stage. The optimal control variable values from each stage served as the starting point for the optimizations of the next phase.

To map the boundaries of the feasible region for balance recovery, a systematic search process was used. First, the range of initial values of X_{COM} and

Z_{HIP} that satisfied the constraints on the initial conditions were determined and a set of values of interest was selected for each variable (e.g. X_{COM} = 0, 0.33, 0.67, and 1.0 foot lengths (fl) from the heel). Next, an initial Z_{HIP} was chosen, and the maximum and minimum values of X_{COM} were found from which the model could be successfully brought to near rest with minimal violations of the constraints. This process was repeated for progressively higher and lower initial values of Z_{HIP} until valid maximum and minimum values of X_{COM} could not be found at a given Z_{HIP}. In other optimizations, an initial X_{COM} was chosen, and the maximum and minimum values of Z_{HIP} were found for which the model could be brought to near rest successfully. Again, the process was repeated for progressively larger and smaller initial values of X_{COM} until valid maximum and minimum values of Z_{HIP} could not be found.

3.2.5 Analysis

The six-link model and two sets of actuator parameter values were used to investigate the effects of aging-related changes in strength on the ability to terminate movement after a backward step following a loss of balance. This was examined by computing and comparing the feasible regions for balance recovery between the models of young and older adults for a given initial horizontal and vertical velocity of the COM ($\dot{X}_{COM}$) and hips ($\dot{Z}_{HIP}$), respectively.

The six-link model was given initial $\dot{X}_{COM}$ and $\dot{Z}_{HIP}$ to simulate a loss of balance in a backward direction. Two combinations of initial velocity were used. These combinations were determined from the results of previous experimental studies of backward balance losses by young and/or older adults following induced slips during a sit-to-stand (Pavol & Pai, 2007) or a simulated lifting task (Welsh & Pavol, 2007). One combination was representative of individuals who successfully recovered and the other was representative of individuals who fell. Typical backward and downward velocities at step touchdown of those who recovered were 6 – 25% body height/s (Bh/s) and 8 – 12% Bh/s, respectively. Corresponding velocities were 6 – 30% Bh/s and 30 – 40% Bh/s in those who fell. Therefore, the combinations of initial velocities used were backward and downward velocities of 15% Bh/s (condition 1), and backward and downward velocities of 30% Bh/s (condition 2).

The feasible region was described by the range of initial X_{COM} and Z_{HIP} for which the model could successfully stabilize the body for the given initial velocities. Feasible regions for young and for older adults were determined using their respective sets of the musculotendon actuator parameters. Through qualitative comparison of the resulting feasible recovery regions, the effect of the aging-related decline in muscle strength was examined.

3.3 Results

Minimum and maximum points of X_{COM} and Z_{HIP} were successfully found through optimization, with only minor violations of the penalty function constraints. The feasible regions of balance recovery by young and older adults for conditions 1 and 2 are presented in Figures 3.3 and 3.4, respectively. The simulations (a) – (h), indicated in these figures, were chosen to examine maximum muscle activations and peak joint moments during balance recovery for different initial positions on the boundaries for conditions 1 and 2. Maximum values of muscle activations and joint moments for simulations (a) – (h) are presented in Tables 3.5 – 3.8.

3.3.1 Feasible Regions of Balance Recovery

For condition 1, corresponding to initial backward and downward velocities of 15% *Bh/s*, there were the largest differences in the forward and the rear boundaries between young and older adults at the lower hip positions. We found that the forward boundary for older adults at a hip height of 50% *Bh* was 25% foot length (*fl*) more posterior than that for young adults and that the rear boundary for older adults at a hip height of 47.5% *Bh* was 12.2% *fl* more anterior than that for young adults. For condition 2, corresponding to initial backward and downward velocities of 30% *Bh/s*, both young and older adults showed smaller feasible regions than those for condition 1. There was the largest difference in the rear boundaries between young and older adults at lower hip positions. We found that the rear boundary for older adults at a hip height of 47.5% *Bh* was 12.8% *fl* more anterior than that for young adults. There was a minor difference in the forward boundary at a hip height of 48.5% *Bh*; older adults showed a 6.4% *fl* more posterior forward boundary than that for young adults. There were negligible differences in the lower and upper boundaries between the feasible regions for young and older adults in conditions 1 and 2.

3.3.2 Muscle Activations

In simulations (b), (c), (d), and (f), fewer muscles were activated over 90% for older adults than young adults. For example, in simulation (b), there were no muscles activated over 90% for older adults, whereas two muscles (biceps femoris and rectus femoris) were activated over 90% for young adults. However, in simulation (e), older adults showed a larger number of highly activated muscles than young adults; older and young adults activated five and four muscles over 90% (older adults: vastii, rectus femoris, gluteals, and adductor magnus; young adults: dorsiflexors, hamstrings, vastii, and rectus femoris). In simulations (a), (g) and (h), the same number of muscles was activated over 90% for young and older adults. In simulations (a) – (h), the total numbers of muscles maximally activated (≥99%) for young and older adults were four and one muscles, respectively. Also, the total numbers of muscles activated over 90% for young and older adults across simulations (a) – (h) were 29 and 21, respectively.

3.3.3 Joint Moments

The maximum joint moments were also examined in the four selected simulations for each of conditions 1 and 2 (simulations (a) – (h) in Figures 3.3 and 3.4; Tables 3.5 – 3.8). In simulation (a), the maximum knee extension moment for young adults was 128% larger than for older adults, consistent with the more extreme initial center of mass position for the young adults. In simulation (e), the maximum hip extension moment for young adults was 47% larger than for older adults, consistent with the more posterior initial center of mass position for the young adults. In contrast, in simulation (h), the maximum hip extension moment for older adults was 92% larger than for young adults, consistent with the slightly lower hip height for the older adults. However, in simulations (c), (d), (f), and (g), where the initial states of the COM and hips were similar for young and older adults, the maximum ankle, knee, and hip joint moments for young and older adults were also similar (within 3% Body weight*Body height).

3.4 Discussion

This study investigated the effect of the aging-related loss in lower extremity strength on the ability to recover balance following a backward step after a backward loss of balance. Two conditions were studied. Condition 1 corresponded to experimentally-observed velocities at step touchdown for successful recoveries following a backward balance loss, whereas the initial velocities for condition 2 were associated with falls (Pavol & Pai, 2007; Welsh & Pavol, 2007). The feasible regions for young and older adults showed similar ranges of hip height in both condition 1 and condition 2. Furthermore, there were large overlaps in the feasible regions for young and older adults in the forward-backward direction. However, despite this overlap, there were also large differences in the feasible regions in the horizontal direction between young and older adults in conditions 1 and 2. For condition 1, older adults showed a 25% *fl* more posterior forward boundary at a hip height of 50% *Bh* and a 12.2% *fl* more anterior rear boundary at a hip height of 47.5% *Bh* than young adults. For condition 2, older adults showed a 6.4% *fl* more posterior forward boundary at a hip height of 48.5% *Bh* and a 12.8% *fl* more anterior rear boundary at a hip height of 47.5% *Bh* than young adults. Therefore, the comparisons of the feasible regions suggest that older adults are less able than young adults to regain balance after a backward step from a backward balance loss if older adults take a very short or very long step. Thus, the aging-related loss in muscle strength could impair the ability to recover balance after a backward balance loss.

The effect of the aging-related loss in muscle strength on the ability to recover balance was apparent in the muscle activations and joint moments for older adults. As one of the aging-related effects on the capacity to generate muscle and tendon forces, we decreased maximum isometric force by 25%. This adjustment would require the muscles that contributed to balance recovery to be activated to greater extent for older adults than for young adults in order to produce the same force. Thus, older adults were expected to show a larger number of highly activated muscles (>90%) to recover balance than young adults. However, in half of the simulations examined, the older adult model used a balance recovery strategy that involved highly activating fewer muscles during the balance recovery motion compared to young adults. For example, in simulation (b), there were no highly activated muscles for older adults, but young adults showed two highly activated muscles. Overall, in simulations (a) – (h), older adults highly activated 21 muscles whereas young adults showed 29 highly activated muscles.

Older adults showed the same ability to recover balance as the young in some simulations, despite having weaker and fewer highly activated muscles than the young. Since there are more muscles at a joint than are required to control the joint (Leijnse et al., 2005), it is possible that older adults relied on multiple muscles instead of single muscles in recovering from a backward balance loss. This would cause older adults to show fewer highly activated muscles than young adults. Also, although the muscle and tendon force-generating characteristics of young adults were greater than for older adults, these characteristics could have worked adversely for young adults due to the dynamic coupling created by rectus femoris and hamstrings, that are biarticular muscles. Rectus femoris is a knee extensor and hip flexor; however, dynamic coupling from knee extension acts to increase hip extension. Although hamstrings is a knee flexor and hip extensor, dynamic coupling from hip extension acts to increase knee extension. These opposing effects of biarticular muscles may counteract the strength advantages of young adult muscles. Finally, similarities in the abilities of young and older adults to recover balance could have resulted from the fact that control signals were determined through optimization separately for young and older adults. The optimization was able to find a different set of control signals for older adults that would allow them to recover from a backward balance loss at the same initial X_{COM} and initial Z_{HIP} from which young adults could successfully arrest the body.

The effect of a 25% decrease in maximum isometric force on the ability to recover balance was also seen in the joint moments developed by older adults during the simulations of balance recovery motions. Simulations (a) and (e) corresponded to the simulations that showed the large differences in the rear boundaries of the feasible regions in condition 1 and 2, respectively. In simulation (a), older adults showed a 128% smaller maximum knee extension moment than young adults. In simulation (e), the maximum hip extension moment for older adults was 47% smaller than for young adults. Thus, it appears that the larger joint moments allowed the young adults to recover from more extreme initial center of mass horizontal positions in simulations (a) and (e) than the older adults. However, in the majority of cases, there was very little difference between young and older adults in the maximum joint moments. This is consistent with the fact that young and older adults showed the same ability to recover balance from a backward loss of balance over a wide range of initial states. Thus, the aging-related loss in muscle strength does not always influence the ability to arrest the body from a backward loss of balance. The effect of aging-related losses in lower extremity strength appeared in the forward and backward boundaries, not in the middle of the feasible regions.

The present findings regarding the changes in the feasible region of balance recovery in older adults contrast with those of Pai and Patton (1997). They found that the normal aging-related loss in muscle strength had no effects on the feasible region of balance recovery from a standing position without taking any steps, and that 35 and 50% losses in plantarflexion and dorsiflexion strength, respectively, were required to reduce the feasible region. The present finding that young and older adults showed similar ranges of initial hip heights in the feasible regions for conditions 1 and 2 was consistent with the results of Pai and Patton. However, in the present study, changes in the horizontal dimensions of the feasible stability region were observed for a 25% decrease in maximum isometric force. A difference is that this study allowed motions at the knee and hip joints, whereas the model of Pai and Patton did not. Thus, the six-link model could move the hips in the downward direction. We believe that the downward motion of the hips amplified the effects of aging-related losses in muscle strength on the ability to recover balance and caused the forward and rear boundaries for older adults to be less anterior and posterior, respectively, than those of young adults.

The models of young and older adults showed different feasible regions of balance recovery for conditions 1 and 2. In condition 2, the initial backward and downward velocities were twice as large as in condition 1 and the resulting feasible regions for condition 2 were smaller than those for the condition 1. The differences in the rear boundary of the feasible region indicate that the minimum step sizes for recovery of both young and older adults is 33% *fl* greater in condition 2 than in condition 1. The differences in the forward boundary were only apparent in the model of young adults. Young adults had a 20% *fl* smaller maximum step size for recovery in condition 2 compared to condition 1. Finally, the ranges of recoverable initial hip heights were smaller for condition 2, as the ranges of both young and older adults in condition 2 were between 46% and 50% *Bh*, compared to 43% to 52.5% *Bh* in condition 1. Thus, doubling the speed at step touchdown had an adverse effect on the ability to recover balance. Young and older adults showed much smaller feasible regions for condition 2 than for condition 1. Functionally, this means that there was a much smaller range of step lengths and hip heights at step touchdown that would allow balance recovery at the faster speed. In addition, the age-difference in the size of the feasible region was smaller at the faster speed. For example, the difference in the anterior boundary between young and older adults in condition 1 was 25% *fl* at a hip height of 50% *Bh*, whereas the difference in the anterior boundary between young and older adults in condition 2 was only 5% *fl* at a hip height of 48.5% *Bh*. Thus, doubling the speed at touchdown across conditions 1 and 2 decreased the difference in the forward boundary between young and older adults by 20% *fl*. This suggests that the aging-related loss of muscle strength has a lesser effect on the ability to recover balance at the fast speeds associated with a high likelihood of falls compared to the slower speeds typically associated with successful recovery.

There were some limitations to this study. We assumed that the front leg was kept straight during the simulations and that only the rear leg was actively controlled by the muscular system. This is based on the belief that, to arrest the backward and downward motion of the body after the completion of a backward step, a person will rely primarily on the rear limb rather than the front limb. To decrease the complexity of the simulation, we also assumed that the front foot was only allowed to slide along the ground and that the rear foot was fixed to the ground. This assumption did not allow either foot to move in the upward direction, limiting the magnitudes of the ankle plantarflexion and dorsiflexion moments that could be used during recovery. This may explain why the aging-related loss in ankle muscle strength had no effects on the upper and lower boundaries for recovery.

Aging-related losses in lower extremity strength could impair the ability to regain balance after taking a backward step from a backward loss of balance. Our results suggest that older adults should avoid taking a very short or very long step to prevent themselves from falling backward. However, if a medium step is taken, older adults can show the same ability to recover balance as young adults.

Therefore, our results suggest that training of the stepping response might prove to be more effective than strength training in preventing backward falls.

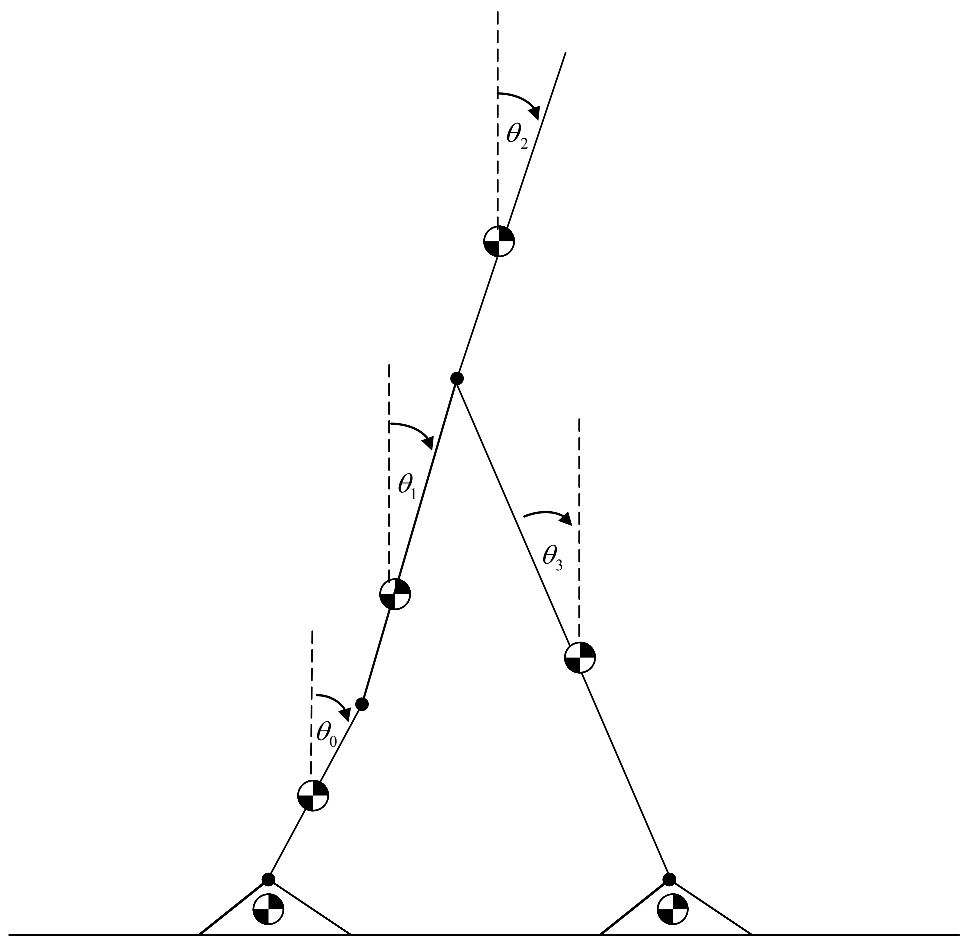

Figure 3.1. Six-link model used in the simulations. The state variables θ_0, θ_1, θ_2, and θ_3 represented the angular positions of the rear leg, the rear thigh, the head-arms-torso, and the front thigh-and-leg segments in the sagittal plane. The ankle, knee, and hip joints were modeled as frictionless hinge joints. The rear foot was fixed to the ground and the front foot was constrained to slide along the ground in the anterior-posterior directions.

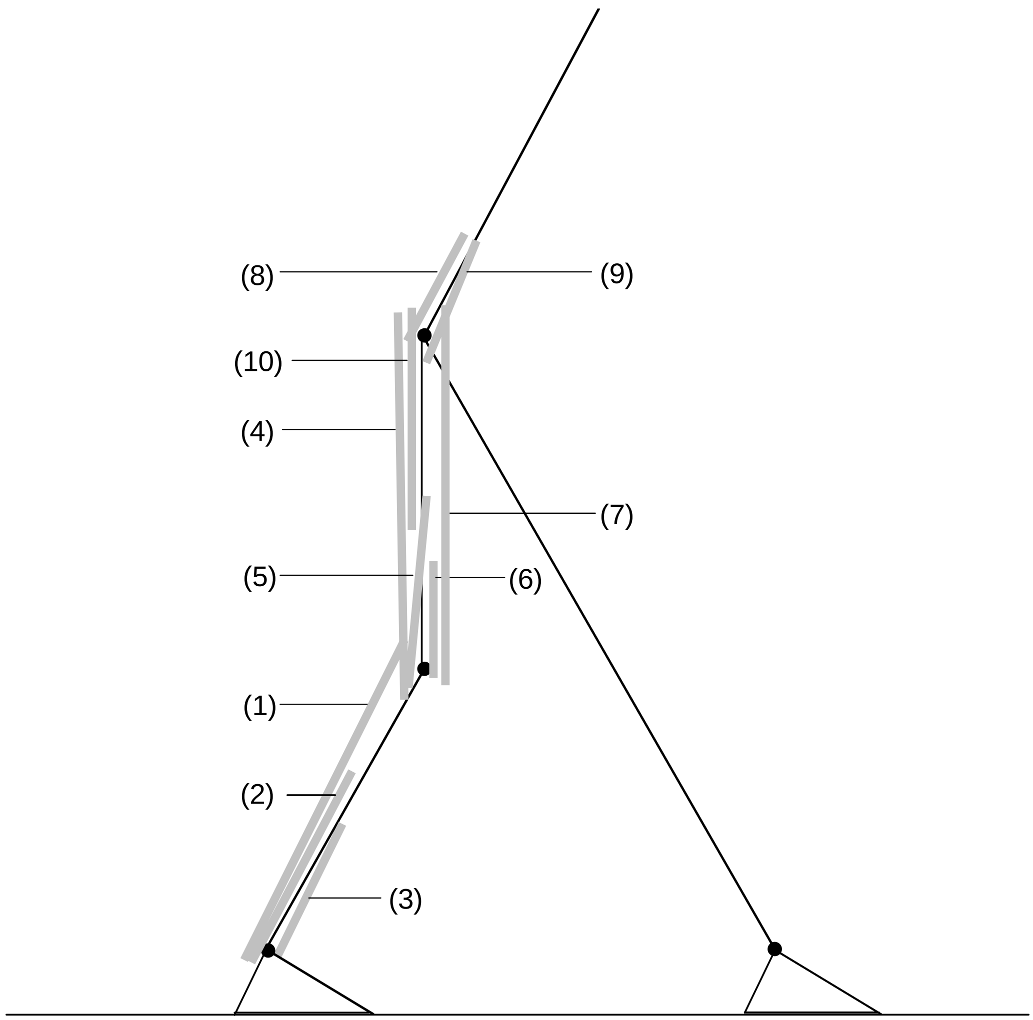

Figure 3.2. Diagram of the musculoskeletal model used to control the rear limb (i.e. the limb that took the step) of the six-link model. Ten Hill-type musculotendon actuators acted across the ankle, knee, and hip joints in the rear limb: (1) gastrocnemius, (2) soleus, (3) dorsiflexor, (4) hamstrings, (5) biceps femoris, (6) vastii, (7) rectus femoris, (8) gluteals, (9) iliopsoas, and (10) adductor magnus.

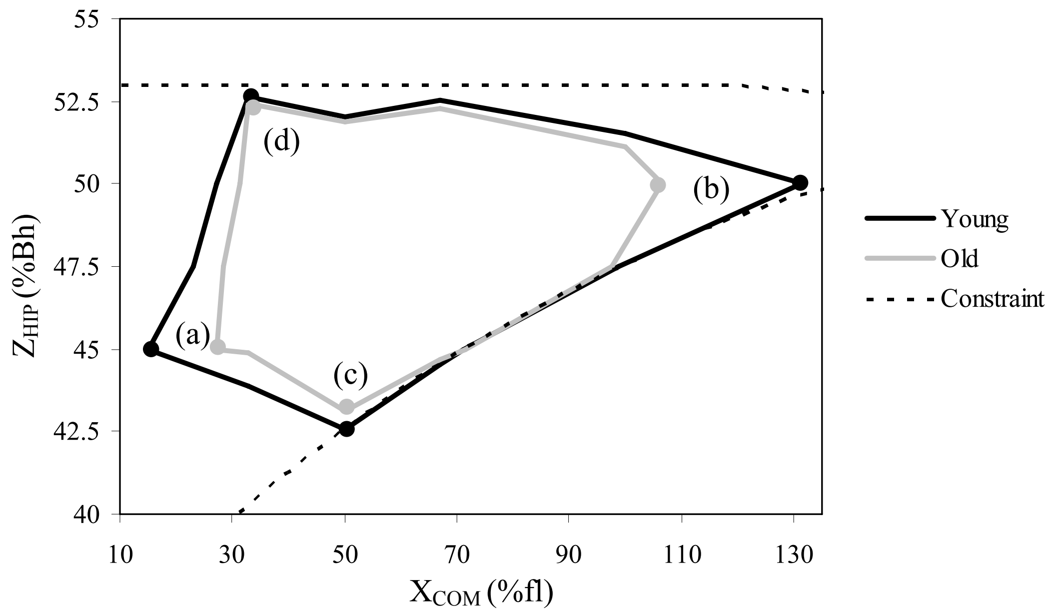

Figure 3.3. The feasible region of balance recovery from which young (black line) and older (gray line) adults could bring their center of mass (COM) and hips to near rest successfully for initial backward and downward velocities of 15% body height/s (*Bh/s*). The dashed lines represent the anatomical constraints on the initial state. (a), (b), (c), and (d) correspond to simulations (a), (b), (c), and (d), respectively. X_{COM} = COM position anterior to the rear heel. Z_{HIP} = hip height. fl = foot length.

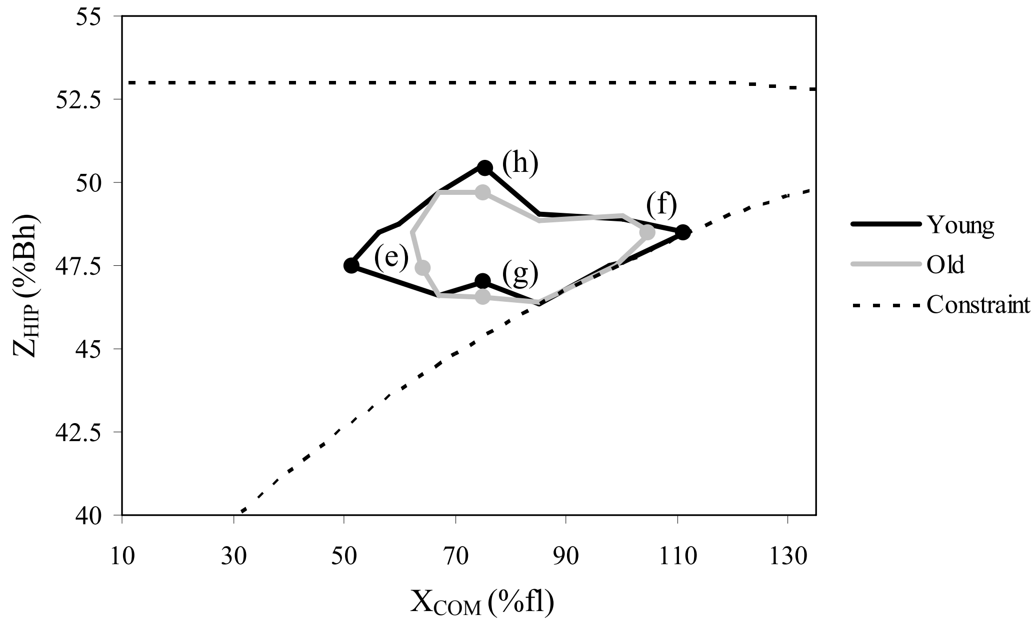

Figure 3.4. The feasible region of balance recovery from which young (black line) and older (gray line) adults could bring their center of mass and hips to near rest successfully for initial backward and downward velocities of 30% body height/s (*Bh/s*). The dashed line represents the anatomical constraints on the initial state. (e), (f), (g), and (h) correspond to simulations (e), (f), (g), and (h), respectively. X_{COM} = COM position anterior to the rear heel. Z_{HIP} = hip height. fl = foot length.

Table 3.1. Constraints on the initial joint angles and joint velocities at the start of the simulations.

Variables	Initial State Conditions
θ_{ANK}	$-0.96 < \theta_{ANK} < 0.44\ rad$
θ_{KNE}	$0 < \theta_{KNE} < 2.53\ rad$
θ_{HIP}	$-0.52 < \theta_{HIP} < 1.57\ rad$
$\dot{\theta}_{ANK}$	$\lvert\dot{\theta}_{ANK}\rvert < 8\ rad/s$
$\dot{\theta}_{KNE}$	$\lvert\dot{\theta}_{KNE}\rvert < 10\ rad/s$
$\dot{\theta}_{HIP}$	$\lvert\dot{\theta}_{HIP}\rvert < 8\ rad/s$

θ_{ANK} = ankle angle, θ_{KNE} = knee angle, θ_{HIP} = hip angle, $\dot{\theta}_i$ = angular velocity of joint i. Positive values correspond to flexion with respect to the anatomical position. Joint angle constraints approximate the anatomical range of motion.

Table 3.2. Penalty functions of the continuous state during the simulations.

Variables	Penalty functions
F_{GZ}^{R}	$F_{GZ}^{R} < 10\,N$
F_{GX}^{R}	$\left(\mu \times F_{GZ}^{R}\right) < F_{GX}^{R}$
COP	Moves outside the middle 95% of the rear foot
F_{GZ}^{F}	$F_{GZ}^{F} < 1\,N$
Z_{KNE}	$Z_{KNE} < 0.3\,m$
Z_{HIP}	$Z_{HIP} < 0.3\,m$

Continuous penalties (g_n) were of the form: $g_n = k_n \int_0^{tf} x_n(t)^2\,dt$, where $x_n(t)$ is the magnitude of the constraint violation (≥ 0) at time t and k_n is a weighting constant. F_{GZ}^{R} = upward ground reaction force on the rear foot, F_{GX}^{R} = horizontal ground reaction force on the rear foot, μ = friction coefficient beneath the rear foot (= 0.7), COP = center of pressure beneath the rear foot, F_{GZ}^{F} = upward ground reaction force on the front foot, Z_{KNE} = knee height of the rear limb, and Z_{HIP} = hip height.

Table 3.3. Penalty functions of the final state at the end of the simulation.

Variables	Constraint violation condition
X_{COM}	$[X_{COM} < 2.5\%\ fl]$ or $[100\%\ fl < X_{COM}]$
Z_{HIP}	$Z_{HIP} < 35\%\ Bh$
$\dot{X}_{COM}$	$\lvert \dot{X}_{COM} \rvert > 0.001\, m/s$
$\dot{Z}_{HIP}$	$\lvert \dot{Z}_{HIP} \rvert > 0.015\ m/s$
$\ddot{X}_{COM}$	$\lvert \ddot{X}_{COM} \rvert > 0.03\ m/s^2$
$\ddot{Z}_{HIP}$	$\lvert \ddot{Z}_{HIP} \rvert > 0.045\ m/s^2$

Final-state penalties (h_n) were of the form: $h_n = k_n x_n^{\ 2}$ where x_n is the magnitude of the constraint violation (≥ 0) at the end of the simulation and k_n is a weighting constant. X_{COM} = center of mass (COM) horizontal position, measured forward from the rear heel, $\dot{X}_{COM}$ = COM horizontal velocity, $\ddot{X}_{COM}$ = COM horizontal acceleration, Z_{HIP} = hip height, $\dot{Z}_{HIP}$ = hip vertical velocity, $\ddot{Z}_{HIP}$ = hip vertical acceleration, fl = foot length, and Bh = body height.

Table 3.4. Maximum muscle activations and joint moments of simulations (a) and (b) for the models of young and older adults.

	Simulation (a)		Simulation (b)	
	Young	Old	Young	Old
X_{COM} (% fl)	15.1	27.3	131.5	106.5
Z_{HIP} (% Bh)	45.0	45.0	50.0	50.0
GAS (%)	21.8	90.6	85.5	84.5
SOL (%)	41.1	83.0	70.0	57.1
DOR (%)	99.6	95.1	69.2	65.1
HAM (%)	42.9	94.0	84.8	67.1
BFM (%)	70.4	38.1	92.3	73.9
VAS (%)	99.0	84.1	62.8	60.3
REC (%)	81.3	97.8	95.0	76.0
GLU/ADD (%)	91.9	69.6	73.5	81.1
ILI (%)	70.5	67.7	56.2	77.4
Ankle PL (% Bw*Bh)	0.8	2.6	6.9	2.7
Ankle DF (% Bw*Bh)	2.9	0.9	—	—
Knee EX (% Bw*Bh)	19.6	8.6	8.7	10.4
Knee FL (% Bw*Bh)	—	—	3.2	3.7
Hip EX (% Bw*Bh)	13.3	16.2	13.3	9.3
Hip FL (% Bw*Bh)	4.8	1.0	—	—

Table 3.4 (continued). In simulations (a) and (b), the initial hip height (Z_{HIP}) and the initial backward and downward velocities of 15% body height/s (*Bh/s*) were given. The respective minimum and maximum center of mass horizontal positions (X_{COM}) with respect to the rear heel were determined from which the six-link model could be brought to near rest. The maximum muscle activations of the 10 musculotendon actuators are shown. GAS = gastrocnemius, SOL = soleus, DOR = dorsiflexor, HAM = hamstrings, BFM = biceps femoris, VAS = vastii, REC = rectus femoris, GLU = gluteals, ILI = iliopsoas, and ADD = adductor magnus. Since GLU and ADD shared the same control signal, the activations of GLU and ADD were identical. In addition, maximum extension and flexion moments during the simulations are shown. PL = plantarflexion, DF = dorsiflexion, EX = extension, FL = flexion. fl = foot length from the heel, Bh = body height, Bw = body weight.

Table 3.5. Maximum muscle activations and joint moments of simulations (c) and (d) for the models of young and older adults.

	Simulation (c)		Simulation (d)	
	Young	Old	Young	Old
X_{COM} (% fl)	50.0	50.0	33.0	33.0
Z_{HIP} (% Bh)	42.6	43.1	52.6	52.4
GAS (%)	89.0	50.9	92.7	53.7
SOL (%)	70.6	89.3	87.7	7.7
DOR (%)	85.5	89.3	70.7	88.3
HAM (%)	92.0	83.3	93.0	89.4
BFM (%)	61.1	87.0	99.3	31.1
VAS (%)	93.6	97.0	63.9	94.4
REC (%)	95.3	83.9	92.4	95.3
GLU/ADD (%)	98.8	89.6	73.4	97.0
ILI (%)	4.2	51.9	93.5	27.5
Ankle PL (% Bw*Bh)	2.5	0.8	0.8	0.7
Ankle DF (% Bw*Bh)	0.7	2.3	2.0	2.6
Knee EX (% Bw*Bh)	12.0	15.0	1.1	1.3
Knee FL (% Bw*Bh)	0.7	—	4.5	3.6
Hip EX (% Bw*Bh)	11.9	11.2	15.7	16.1
Hip FL (% Bw*Bh)	—	0.6	—	—

Table 3.5 (continued). In simulations (c) and (d), the initial center of mass position (X_{COM}) with respect to the rear heel and the initial backward and downward velocities of 15% body height/s (*Bh*/s) were given. In (c) and (d), the respective minimum and maximum initial hip heights (Z_{HIP}) were determined from which the six-link model could be brought to near rest. The maximum muscle activations of the 10 musculotendon actuators and the maximum extension and flexion moments during the simulation are shown. Abbreviations are the same as in Table 3.5.

Table 3.6. Maximum muscle activations and joint moments of simulations (e) and (f) for the models of young and older adults.

	Simulation (e)		Simulation (f)	
	Young	Old	Young	Old
X_{COM} (% fl)	50.9	63.7	111.7	105.3
Z_{HIP} (% Bh)	47.5	47.5	48.5	48.5
GAS (%)	9.6	83.1	82.6	66.6
SOL (%)	8.9	14.3	91.4	88.7
DOR (%)	99.7	91.3	77.5	35.3
HAM (%)	93.1	73.4	85.6	98.2
BFM (%)	11.6	71.4	76.8	53.6
VAS (%)	96.5	99.5	67.0	85.5
REC (%)	91.4	92.6	93.4	60.1
GLU/ADD (%)	86.8	96.8	90.7	59.6
ILI (%)	13.5	18.9	28.1	91.4
Ankle PL (% Bw*Bh)	0.4	1.3	3.1	4.2
Ankle DF (% Bw*Bh)	2.9	2.8	0.5	0.2
Knee EX (% Bw*Bh)	14.5	13.0	10.2	8.5
Knee FL (% Bw*Bh)	—	—	3.7	3.5
Hip EX (% Bw*Bh)	19.5	13.3	10.8	9.3
Hip FL (% Bw*Bh)	4.2	—	—	—

Table 3.6 (continued). In simulations (e) and (f), the initial hip height (Z_{HIP}) and the initial backward and downward velocities of 30% body height/s (*Bh/s*) were given. In (e) and (f), the respective minimum and maximum center of mass horizontal positions (X_{COM}) with respect to the rear heel were determined from which the six-link model could be brought to near rest. The maximum muscle activations of the 10 musculotendon actuators and the maximum extension and flexion moments during the simulation are shown. Abbreviations are the same as in Table 3.5.

Table 3.7. Maximum muscle activations and joint moments of simulations (g) and (h) for the models of young and older adults.

	Simulation (g)		Simulation (h)	
	Young	Old	Young	Old
X_{COM} (% fl)	75.0	75.0	75.0	75.0
Z_{HIP} (% Bh)	47.0	46.5	50.5	49.7
GAS (%)	78.8	66.0	85.2	61.5
SOL (%)	82.3	66.2	51.6	55.4
DOR (%)	85.2	91.4	86.5	89.6
HAM (%)	94.6	78.0	94.6	95.3
BFM (%)	56.9	40.4	2.5	5.1
VAS (%)	94.1	94.6	88.8	89.7
REC (%)	74.7	87.6	75.8	97.4
GLU/ADD (%)	81.5	82.5	97.3	68.2
ILI (%)	59.0	25.6	83.7	98.0
Ankle PL (% Bw*Bh)	2.0	1.9	2.9	2.5
Ankle DF (% Bw*Bh)	0.9	2.0	2.8	—
Knee EX (% Bw*Bh)	14.0	11.6	10.8	8.3
Knee FL (% Bw*Bh)	—	2.2	1.6	—
Hip EX (% Bw*Bh)	14.9	13.3	9.6	18.4
Hip FL (% Bw*Bh)	—	—	—	0.8

Table 3.7 (continued). In simulations (g) and (h), the initial center of mass position (X_{COM}) with respect to the rear heel and the initial backward and downward velocities of 30% body height/s (*Bh/s*) were given. In (g) and (h), the respective minimum and maximum initial hip heights (Z_{HIP}) were determined from which the six-link model could be brought to near rest. The maximum muscle activations of the 10 musculotendon actuators and the maximum extension and flexion moments during the simulation are shown. Abbreviations are the same as in Table 3.5.

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4.0 Conclusion

Falls and fall-related injuries can have severe impacts on an older adult's life. Because backward falls are a major cause of hip fractures in older adults, we need to prevent older adults from falling backwards so that they can maintain a high quality of life. Although the effects of the normal aging-related loss in lower extremity loss have been studied, it still remains unknown to what extent the aging-related loss in lower extremity strength affects the ability to recover balance after the completion of a recovery step from a backward balance loss. To answer this question, two-dimensional musculoskeletal models of young and older adults were developed and backward balance recovery of young and older adults was simulated. The effects of the aging-related loss in lower extremity strength on the ability to recover balance after completing a stepping motion following a backward balance loss were quantified by describing the feasible regions of balance recovery.

In the first study, a two-dimensional musculoskeletal model of the lower extremity was developed that was suitable for forward dynamic simulations. Also, the parameter values of young and older adult muscles were determined. The musculoskeletal model consists of rigid foot, leg, thigh, and pelvis segments linked by hinge joints representing the ankle, knee, and hip, and driven by a system of 10 Hill-type musculotendon actuators. First, the parameter values of young adults were found by reviewing the existing literature and using a least-squares optimization technique. Second, previous studies of the effects of aging on muscle strength were reviewed and the parameter values of older adults were determined by adjusting the parameter values of young adults. It was found that the simulated isometric joint moments for the young adult actuators were consistent with the corresponding experimental data, and that dynamic properties of force-generation for young adults were physiologically reasonable. Finally, the behaviors of older adult muscle matched my expectation. This model thus provides a new tool for investigating the aging-related changes in movement patterns.

In the second study, the effects of the normal aging-related loss in lower extremity strength on the ability to terminate movement after a backward step following a loss of balance were investigated. A six-link model was developed to simulate backward balance recoveries by young and older adults, in order to describe feasible regions of balance recovery for young and older adults. Two combinations (combinations 1 and 2) of initial velocities were given to the six-link model at step touchdown to simulate a backward balance loss. The initial velocities for condition 1 were backward and downward velocities of 15% *Bh/s*, corresponding to experimentally-observed velocities at step touchdown for successful recoveries following a backward balance loss. The initial velocities for condition 2 were backward and downward velocities of 30% *Bh/s*, corresponding to experimentally-observed velocities associated with falls.

Balance recovery for young and older adults was successfully simulated with only minor violations of the penalty function constraints. The feasible regions of young and older adults were described for conditions 1 and 2. Comparing the feasible regions of young and older adults, it was found that the models of young and older adults showed similar ranges of recoverable initial hip heights and large overlaps in the ranges of recoverable initial COM horizontal positions in both conditions 1 and 2. However, there were large differences in the feasible regions in the horizontal direction between young and older adults in conditions 1 and 2. For condition 1, older adults showed a 25% *fl* more posterior forward boundary at 50% *Bh* and a 12.2% *fl* more anterior rear boundary at 47.5% *Bh* than young adults. For condition 2, older adults showed a 6.4% *fl* more posterior forward boundary at 48.5% *Bh* and a 12.8% *fl* more anterior rear boundary at 47.5% *Bh* than young adults.

The comparisons of the feasible regions suggest the following. The effects of aging on balance recovery are apparent in a very short or very long step, whereas the normal aging-related changes in muscles have no effects on the range of recoverable hip heights at step touchdown. If older adults take a very short or very long step, older adults are less able than young adults to regain balance after a backward step from a backward balance loss. Therefore, the aging-related loss in lower extremity strength could impair the ability to recover balance from a backward balance loss during the stance phase after a recovery step. However, if an older adult takes a medium-sized backward step, the normal aging-related loss in muscle strength has no effect on the ability to recover balance and older adults are able to regain balance. Thus, taking an appropriate step length may be a more important factor than lower extremity strength for preventing older adults from falling backward.

The effects of aging-related losses in muscle strength on the ability to recover balance were influenced by the initial backward and downward velocities.

Initial backward and downward velocities of 15% and 30% *Bh/s* were used for conditions 1 and 2, respectively. The models of young and older adults showed smaller feasible regions of balance recovery for condition 2 than for condition 1. Young and older adults showed 33% *fl* less posterior rear boundaries of the feasible regions for condition 2 than condition 1. As far as the forward boundary, only young adults showed a difference between conditions. The forward boundary of young adults in condition 2 was 20% *fl* less anterior than that in condition 1. The ranges of recoverable initial hip heights for young and older adults in condition 1 were between 43% and 52.5% *Bh*; those in condition 2 were between 46% and 50% *Bh*. Thus, doubling the speed at step touchdown had an adverse effect on the ability to recover balance. The results suggest that increments in initial velocities decrease the size of the feasible region. In addition, the effects of aging on the ability to recover balance appeared to be smaller at the faster speed. For example, there was a 25% *fl* difference between young and older adults in the forward boundary of the feasible region at a hip height of 50% *Bh* in condition 1. However, in condition 2, the difference in the forward boundary between young and older adults was only 5% *fl* at a hip height of 48.5% *Bh*. Thus, the difference in the forward boundary between young and older adults decreased by 20% *fl* from condition 1 to 2. Therefore, the effects of the aging-related losses in lower extremity strength on the ability to recover balance appear to decrease at the faster speeds associated with a high likelihood of falls compared to the slower speeds typically associated with successful balance recovery.

The six-link model was developed to investigate the effects of aging-related changes in muscles on the ability to recover balance from a backward balance loss. The models of young and older adults showed aging-associated differences in the forward and rear boundaries of the feasible regions in conditions 1 and 2. However, there were no aging-related differences in the range of recoverable initial hip heights in conditions 1 and 2. This could be due to the limitation that the front and rear feet stayed flat on the ground. This limitation did not allow the heels or toes to move in the upward direction. Because the range of ankle motion was limited, ankle plantarflexion moments could not fully contribute to arresting the descent of the hips. Thus, questions still remain regarding the effects the normal aging-related loss in ankle strength has on the ability to recover balance during the stance phase following a recovery step from a backward balance loss. There is a need to develop a six-link model that can raise its feet in order to further investigate the effects of aging-related losses in ankle strength on the feasible range of initial hip states.

Other issues to be addressed are the properties of older adult muscles. In this study, the effects of normal aging-related losses in muscle strength on the ability to recover balance were investigated. The model of older adults represents healthy older adults. Older adults who tend to fall may be physically weaker than typical older adults. Thus, there is a need to study the effects of extreme aging-related losses in muscle strength on the ability to recover balance. In addition, parameter values of the 10 musculotendon actuators were equally adjusted to simulate the effects of aging. It remains unknown what effects aging-related losses in the strength of an individual muscle has on the ability to recover balance. This point needs to be studied in the future.

In this study, feasible regions of balance recovery were described to investigate the effects of the normal aging-related loss in lower extremity strength on the ability to recover balance during the stance phase following a backward step from a backward balance loss. It was found that the normal aging-related changes in muscle strength could impair the ability to recover balance, particularly when the body is falling slowly at step touchdown. However, if an older adult takes a medium-sized step, an older adult has the same ability to recover balance as young adults. Therefore, this study concludes that stepping-response training might be more effective than strength training in enabling older adults to avoid a backward fall.

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Appendix

Appendix – Literature Review

A.1 Falls Epidemiology

Falls and fall-related injuries are major concerns for older adults (Tinetti et al., 1988). The Centers for Disease Control and Prevention (2006) reported that, in 2004, 15,000 older adults died from fall-related injuries in the United States. There were 10,300 fatal fall injuries and 2.6 million non-fatal fall injuries among people aged 65 years or older in 2000 (Stevens et al, 2006). The estimated medical costs of fatal fall and non-fatal fall injuries were $180 million and $20 billion respectively. Thirty percent of older adults fall at least once every year (Tinetti et al., 1988). Twenty-five percent of falls cause serious injuries and 5% result in bone fracture (Tinetti et al., 1988; Alexander et al., 1992).

Fall-related injuries can have a severe impact on the quality of daily living in older adults. Falls are the main causes of hip fractures (Grisso et al., 1991; Vogt et al., 2002), the consequences of which can be serious. Sixty percent of older adults who lived in the community before hip fracture are admitted to a nursing home within one year (Leibson et al., 2002). Twenty percent of those people who were admitted to a nursing home continue to stay there one year after their hip fracture. Older adults with hip fracture show a 10% increment in mortality rate after the incident (Leibson et al., 2002). Older adults who suffer a hip fracture can thus lose their independence and their lives. In addition, fall-induced hip fracture can decrease the quality of life of older adults. Wolinsky et al. (1997) reported that hip fractures led to greater difficulties in daily activities and limited ranges of motion of the lower extremity among older adults. Older adults who experienced a fall also often show a fear of falling and avoid or limit daily activities (Vellas et al., 1997; Tinetti et al., 1988; Zijlstra et al., 2007). Therefore, preventing fall-related injuries is an imperative.

The mechanisms underlying fall-related injuries, including hip fracture, have been studied. Nevitt et al. (1989) reported that 0.2% of falls resulted in a hip fracture and 3% of falls caused bone fractures other than a hip fracture. The direction of a fall influences the type of bone fracture that will occur (Cummings & Nevitt, 1989). One factor that can affect the direction of a fall is walking speed. Slow walking produces little forward momentum, so the pelvis moves straight down, to the side, or backward during a fall and a hip fracture is more likely to occur; fast walking can generate forward momentum that carries a faller forward on to the knees or wrists, and knee or wrist fractures are more likely to occur (Cummings & Nevitt, 1989). Smeesters et al. (2001) studied whether falling directions and hip impact location were associated with gait speed. They found that a fast gait speed tended to result in forward falls in which the faller landed on the abdominal pelvis, and that a slow gait speed was more likely to cause sideward or backward falls in which the fallers landed on the hip. Thus, a backward fall is a risk factor for hip fracture, as is a sideward fall.

Thus, backward falls appear to be a major cause of hip fracture. These hip fractures can pose a large potential threat to the life or quality of life of an older adult. Prevention of backward falls is therefore a major public health issue in our society.

A.2 Muscle Strength, Balance, and Falls

The literature suggests that balance loss and backward falls could be associated with a decline of muscle strength with advancing age. It has been reported that muscle strength at the ankle, knee, and hip joints play an important role in maintaining balance (Duncan et al., 1993; Wolfson et al., 1995; Daubney & Culham, 1999 Binda et al, 2003; Carter et al., 2002; Karinkanta et al., 2005). Duncan et al. (1993) found that older adults with impaired balance control showed weaker dorsiflexion and plantarflexion strengths at the ankle joints than older adults with non-impaired balance control. Daubney and Culhan (1999) reported that ankle strength was significantly associated with balance. In addition, Carter et al. (2002) and Karinkanta et al. (2005) reported that the knee extensors play a significant role in dynamically controlling balance. Thus, a decline in strength at the ankle, knee, or hip joints appears to result in impaired control of static and dynamic balance, potentially increasing the likelihood of a balance loss that could lead to a backward fall.

Differences in muscle strength and power between fallers and non-fallers among older adults have been studied (Whipple et al., 1987; Melzer et al., 2004; Sieri & Beretta, 2004). Whipple et al. (1987) examined differences in peak torque and power at the ankle and knee joints between fallers and non-fallers among older adults. They reported that the ankle and knee peak torques and powers of fallers were significantly lower than those of non-fallers. In fallers, the ankle torque was 10% of the knee torque whereas, in non-fallers, the ankle torque was 25% of the knee torque. Also, fallers generated ankle power that was 11% of their knee power, while non-fallers generated ankle power that was 22% of their knee power. In addition, the generation of knee and ankle power decreased rapidly at faster velocities in fallers. At high ankle joint velocities, the ankle power of fallers was lower by 80% than that of non-fallers. Daubney and Culham (1999) also studied lower extremity muscle strength and balance of fallers and non-fallers among older adults. They found that the strength of the ankle dorsiflexor and hip extensor muscles of fallers was lower by 50 and 20%, respectively, than that of non-fallers. Therefore, among the characteristics of fallers are weak muscle strength at the ankle, knee, and hip joints (Whipple et al., 1987; Daubney & Culham, 1999). Peak torque and power of the ankles, knees, and hips of fallers are lower than those of non-fallers. In particular, ankle strength is deteriorated to a greater extent than knee strength in fallers, and the amount of ankle power that fallers can generate drops steeply at higher velocities to greater extent than at the knee. Given that weak lower extremity muscles are associated with impaired control of static and dynamic balance, these strength characteristics of fallers have been hypothesized to directly contribute to backward balance losses and falls.

Consistent with this idea, the effects of strength training on balance improvement and falls prevention have been studied (Amiridis et al., 2005; Hauer et al., 2001; Barnett et al., 2003; Orr et al., 2006). Amiridis et al. (2005) examined the effect of strength training of the ankle dorsiflexors on static balance among older adults. They found that an increase in dorsiflexor strength improved the static balance of older adults. Orr et al. (2006) examined the effect of 10 weeks of knee strength training on balance improvement among older adults. After the training, subjects showed significant increases in knee strength. The training also resulted in improvements in static and dynamic balance. Hauer et al. (2001) studied the effect of a 3-month strength and balance training program on falls prevention among older adults. Older adults who underwent the training showed an increase in lower extremity strength and a 25% decrease in fall incidence during the 3 months of training and a 6-month follow-up period. Barnett et al. (2003) reported that physically weak older adults with 6 months of group exercise training showed better static balance than those without group exercise training. They found that, during the 12-month follow-up period, the rate of falls in the weak older adults who had the group exercise training was 40% lower than in those who had no group exercise training. Thus, improvement of muscle strength can contribute to balance control and fall prevention in weak older adults.

In healthy adults, it appears that aging-related declines in muscle strength can contribute to reducing stepping velocity in regaining balance (Thelen et al, 1997, 2000; Wojcik et al., 1999; Madigan & Lloyd, 2005b). Stepping velocities in recovering from forward losses of balance have been examined (Wojcik et al., 1999; Thelen et al., 2000; Madigan and Lloyd, 2005b). Wojcik et al. (1999) found that stepping velocities of older adults were significantly slower than those of young adults. Madigan and Lloyd (2005b) examined the association between the age-related reduction in stepping velocity and the peak joint velocities of the stepping leg in recovering from a forward loss of balance. In recovering from the maximum forward lean angle, peak hip flexion, knee flexion, knee extension, and ankle plantarflexion velocities of older adults were lower by 25 – 30% than those of young adults. These reductions in velocity could be the result of slower muscle activation rates or a slower rate of torque generation. Thelen et al. (2000) studied muscle activation patterns in the lower extremity of young and older adults during stepping to recover from a forward loss of balance. They found that activation of vastus lateralis in the stepping leg of older adults was slower by 40 ms than in young adults, and that older adults showed slower rates in deactivating the muscles of the stance leg and in activating the muscles of the step leg during the stepping movement. Thelen et al. (1996) further reported that older adults generated ankle joint torques at a slower rate than young adults. Thus, a decline in peak joint velocities due to lesser muscle strength or slower muscle activation rates could reduce the stepping velocities of older adults in recovering from a backward loss of balance, resulting in a lesser ability to restore stability by stepping.

However, recovery from a loss of balance also requires supporting the weight of the body and stopping its downward motion during the stance phase that follows the recovery step. This may require a greater amount of muscle strength than is required for stepping. Despite this, few studies have examined the effect of aging-related declines in muscle strength on balance recovery during the stance phase following the recovery step. Madigan and Lloyd (2005a) studied the effect of aging on the joint torques during the stance phase following a step to recover from a forward loss of balance. They found that the peak knee extensor torque of older adults during stance phase was significantly smaller than that of young adults. They reported that older adults produced high peak hip extensor and peak ankle plantarflexor torques to compensate for the loss of knee extensor torque.

Thus, not only could a loss of strength result in a decline in stepping velocity during swing phase, a loss of knee extensor strength could reduce the ability of older adults to support the body in regaining balance during stance phase after the recovery step. However, Pavol et al. (2002) studied the relationship between the muscle strength of older adults and the recovery outcome of an induced balance loss during walking. They found that fallers had greater strength in the lower extremity than non-fallers. Therefore, the literature does not indicate whether normal aging-related declines in muscle strength will have a large or a minor impact on the ability to recover from a backward balance loss.

A.3 Models of the Feasible Region for Balance Recovery

Older adults could fall backward if a decline in peak muscle torque and power impairs the ability to regain balance during stance phase after the recovery step from a backward balance loss. The effect of an aging-related decline in muscle strength on the ability to regain balance can be determined by describing the feasible region for balance recovery in terms of the displacement of the center of mass (COM) in the horizontal direction and of the hips in the vertical direction. The feasible stability region has been defined as the range of initial positions and velocities of the COM for which the COM can be brought to rest within and relative to the base of support (BOS) (Pai & Patton, 1997; Iqbal & Pai, 2000). Dynamic models of balance recovery have been used to predict the feasible region (Yang et al., 1990; Yang et al., 2007; Pai & Patton, 1997; Iqbal & Pai, 2000). A musculoskeletal model is typically incorporated into these dynamic models, and this musculoskeletal model can have the muscle properties of young or older adults. Thus, the effect of aging-related declines in muscle strength on the ability to prevent a backward fall after a recovery step can be examined through determining the associated changes in the corresponding feasible region for balance recovery.

The task of maintaining or recovering balance has been modeled statically and dynamically (Yang et al., 1990; Pai & Patton, 1997; Pai et al., 1998). The validity of dynamic models was evaluated by Pai et al. (1998), who compared static and dynamic model predictions of the feasible stability region with experimental results for a forward perturbation. Sixty-five percent of the dynamic model predictions matched the experimental results, whereas the static model prediction showed only 5% accuracy. Thus, a dynamic model of balance recovery motion is a valid method to predict the feasible region for balance recovery from a backward balance loss.

The feasible regions for balance recovery have been predicted to examine balance control under various conditions (Pai & Patton, 1997; Pai & Iqbal, 1999; Iqbal & Pai, 2000; Yang et al., 2007). Pai and Patton (1997) examined the feasible stability region for balance control during upright standing using the dynamic model of an inverted pendulum. They predicted the feasible region of COM anterior position-velocity combinations for which the dynamic model could maintain balance without stepping for different ankle muscle strengths. The boundary for backward fall initiation of their model of an average male ranged from heel position for an initial velocity of 0 body height/s to 2.7 foot-lengths behind the heel for an initial forward velocity of 0.9 body height/s. If the initial position and velocity of the COM was behind this boundary, the model would fall backwards. Pai and Patton (1997) decreased ankle muscle strength and investigated the effect of the aging-related loss in muscle strength on the feasible stability regions. They concluded that the normal aging-related loss in muscle strength had no effect on the ability of older adults to prevent the initiation of forward and backward falls from a standing position.

As described, dynamic models and the feasible stability region have been used to examine the ability to control balance (Pai & Patton, 1997; Pai & Iqbal, 1999; Iqbal & Pai, 2000; Yang et al., 2007) and, using this approach, Pai and Patton (1997) found that there was no effect of the normal aging-related loss in muscle strength on the ability to maintain balance recovery in a standing position without a step. However, stepping typically does occur following a loss of balance. No studies have previously investigated what effect the aging-related deterioration of muscle strength in the lower extremity has on balance control after a step following a backward loss of balance. Arguably, the findings of Pai and Patton will not be generally applicable, as their model did not allow motion at the knee and hip joints, so it could not take into account the effects of a downward motion of the hips on the ability to regain balance. To address this limitation, a modified version of the feasible stability region can be defined as the range of initial COM anteroposterior positions and hip heights from which the body can be brought to rest in a stable position for a given initial backward and downward velocity. By using an appropriate musculoskeletal model with the muscle properties of young and older adults, such effects of aging-related reductions in muscle strength can be examined.

A.4 Musculoskeletal Models

Musculoskeletal models are used in forward dynamic simulation to simulate human motions and understand mechanisms underlying human motions. In general, a musculoskeletal model consists of rigid links representing segments of the body, joints between these links, musculotendon actuators that simulate the force-generating properties of the muscles and tendons acting across joints, and passive joint moments that simulate passive restraints to joint motion. A set of neural control signals is used to simulate the muscle excitation levels and drive the model.

Different types of musculoskeletal models and control signals approaches have been proposed (Chao & Rim, 1973; Audu & Davy, 1985; Winters & Stark, 1985; Zajac et al. 1986; Zajac, 1989; Hoy et al., 1990; Riener & Edrich, 1999; Thelen, 2003). In particular, there have been many approaches to the modeling of the system of musculotendon actuators within a model. There are two main types of muscle actuators used: Hill-type muscle models and torque generators (Chao & Rim, 1973; Audu & Davy, 1985; King et al., 2006). Hill-type muscle models can be further divided into individual muscle actuators and equivalent muscle actuators (Blemker & Delp, 2005). A muscle model is typically activated by a simulated neural control signal. The validity of muscle models and control signal approaches has been evaluated (Audu & Davy, 1985; Zajac, 1989; Hoy et al., 1990; Perreault et al., 2003; Camilleri & Hull, 2005; Blemker & Delp, 2006; King et al., 2006) and each approach has its benefits and disadvantages in terms of physiological accuracy and computational complexity.

As stated, Hill-type muscle models and torque generators are the most commonly used muscle actuators. The Hill-type muscle model consists of a contractile element (*CE*), a passive elastic element (*PE*), and series elastic element (*SE*) (Audu & Davy, 1985; Zajac, 1989). The *CE* represents the active state dynamics of muscles. The active muscle force that the *CE* generates depends on activation, length-tension and force-velocity relationships. The *PE* and *SE* represent the passive state dynamics of muscle and tendon, respectively. The *PE* and *SE* function as nonlinear-elastic springs to produce passive muscle and tendon forces, respectively. Hill-type muscle models produce joint moments by generating force along a line of action between its origin and insertion points. In contrast, a torque generator directly produces a torque about a joint. It typically comprises a *CE* and a *PE* whose properties are a function of joint angles and angular velocities. The approaches of using Hill-type muscle models and torque generators have been reviewed (Audu & Davy, 1985; Blemker & Delp, 2006; King et al., 2006).

Hill-type muscle models can be divided into two types: equivalent muscle actuators and individual muscle actuators. Equivalent muscle actuators model multiple muscles that have the same action on a segment as one muscle group, whereas individual muscle actuators model a single muscle. Musculoskeletal models employing equivalent muscle actuators have less complex structures and are less computationally intensive than those using individual muscle actuators

(Audu & Davy, 1985; Blemker & Delp, 2006). On the other hand, individual muscle actuators can be used to examine the contributions of single muscles to a movement task, but equivalent muscle actuators can only indicate the role of a muscle group. Torque generators have the least computational difficulty among the three types of muscle models. But torque generators cannot evaluate the contribution of single muscles, nor can they model the behaviors of biarticular muscles. The general structure of Hill-type muscle models is also better suited for use in forward dynamic simulation to examine how changes in muscle properties affect joint motions and accomplishment of a motor task. Regardless of what muscle actuators are used, another important consideration is that the elasticity of the other soft tissues at a joint can generate passive moments that provide restraints to joint ranges of motions (Riener & Edrich, 1999). A model of these passive joint moments is also required in a musculoskeletal model.

Production of *CE* force in a muscle model depends on the assumed activation level of the muscle (Zajac, 1989). This activation is governed by a neural control signal (Audu & Davy, 1985; Zajac, 1989; Hoy et al., 1990; Perreault et al., 2003; Camilleri & Hull, 2005; King et al., 2006). There are two types of control signal approaches: constant activation and dynamic activation (Zajac, 1989; Hoy et al., 1990). In the former approach, the activation remains constant throughout the simulation, whereas, in the latter approach, the activation level changes over the course of the simulation. Hoy et al. (1990) employed the constant maximal activation approach, which is the simplest approach. The dynamic activation approach approximates the muscle activation-deactivation mechanism by first- or second-order differential equations (Zajac, 1989) and is more realistic. However, this dynamic activation approach needs to be governed by a neural control signal. Finding an optimal neural control signal usually requires a non-linear optimization routine and very large computational expense (Davy & Audu, 1987; Pandy et al., 1990; Neptune, 1999; Anderson & Pandy, 2001; Thelen & Anderson, 2006).

Forward dynamic simulation has the potential to contribute to a greater understanding of the relationship between characteristics of musculotendon contractile properties and human movements. In particular, the effects of aging-related changes in muscle strength can be simulated by adjusting parameter values within a musculoskeletal model (Thelen, 2003). However, a major disadvantage of forward dynamic simulation is that the determination of optimal neural control signals requires large computational expenses. One of the solutions to this problem is to reduce the number of muscle actuators contained within the musculoskeletal model, and hence the number of neural control signals needed. A general approach to developing musculoskeletal models for forward dynamic simulation would thus be to include the minimum number of equivalent muscle actuators and individual muscle actuators needed to generate physiologically reasonable muscle strength characteristics. Such a model could then be used to investigate the effects of aging-related changes in muscle properties on human movements, such as balance recovery.

A.5 Aging-Related Changes in Muscles

The structure of muscles undergoes alterations with advancing age (Larsson et al., 1978; Sato et al., 1984; Frontera et al., 1991, 2000; Thompson, 2002). Older adults lose 10% of the peak muscle mass that they had at young ages by age 65 years (Lindle et al., 1997). In addition, the distribution of muscle fibers changes with advancing age (Larsson et al., 1978). These age-related changes in muscle structure cause deteriorations in muscle contractile properties (Lindle et al., 1997; Thompson, 2002). Peak muscle torque, power, and shortening velocity decrease by 20 to 60% as age increases from 20 to 85 years (Thom et al., 2007). After 65 years of age, an older adult loses approximately 2.5% of muscle strength in the lower extremities every year (Frontera et al. 2000).

A main contributor to aging-related decreases in muscle strength is a loss of muscle mass. Muscle mass declines with advancing age, accompanied by a proportional decrease in muscle volume (Thom et al., 2005). Thom et al. (2005) reported that older adults showed a 20% decrease in muscle volume, and that muscle power in older adults was smaller by 60% than that in young adults. Muscle power is a function of peak muscle torque and angular velocity. Thus, muscle torque and contractile velocity for a given joint velocity and load, respectively, are reduced by a loss of muscle mass.

The aging-related loss of muscle mass results from reductions both in the sizes and numbers of single muscle fibers (Sato et al., 1984; Aniansson et al., 1986; Häkkinen & Häkkinen, 1991). In general, muscle fibers are categorized into three types: type I, type IIA, and type IIB (Thompson, 1994). Contraction velocity in Type I fiber is slower than those in type IIA and IIB, whereas type I has more fatigue resistance than type IIA and IIB, and type IIA fiber is more fatigue-resistant than type IIB fiber (Thompson, 1994). Häkkinen and Häkkinen (1991) studied properties of the cross-sectional areas of muscles in women aged approximately 30, 50, and 70 years. Cross-sectional area of muscles in 70 year-old women was significantly smaller than that in those aged 30 and 50 years. Aniansson et al. (1986) investigated the effect of age after the sixth decade of life on a fiber cross-sectional area. They found that, after age passed 65 years, the cross sectional areas in type IIA and B fibers were reduced by 15 – 25%, and muscle fiber types and distributions were significantly correlated with the maximal voluntary forces. Consistent with this, Trappe et al. (2003) reported that there was no difference in the maximal forces of single type I muscle fibers between persons aged 25 and 80 years, but that 80 year-olds showed a 20% decline in the maximal force of type II fibers compared to 25 year-olds. Ochala et al. (2007) found that the maximal forces of single type I and IIA muscle fibers in persons aged 65 years were significantly lower by 25 and 35%, respectively, than those in persons aged 30 years. Thus, the maximal forces of type I and II muscle fibers in older adults can decrease by 20 to 30% compared to young adults, causing a decline in muscle strength.

In addition to the changes in muscle fiber cross-sectional area and force, a loss of motor units with older age decreases the number of muscle fibers (Edström & Larsson, 1987). Sato et al. (1984) studied the number of muscle fibers in the pectoralis minor among different age-groups of women. Total numbers of muscle fibers of women in the 30's, 50's, 60's, and 70's year-old groups were 156, 158, 131, and 118, respectively. Thus, a large reduction in the total number of the muscle fibers occurred after 60 years of age. With fewer muscle fibers, the capacity of the muscle to produce force decreases. Those changes result in a decline of peak muscle power and torque (Thom et al., 2005).

Aging also causes changes in the distributions of muscle fibers, with a preferential decrease in the number of type IIA and IIB fibers with age (Larsson et al., 1978; Ishihara & Araki, 1988). Sato et al. (1984) found that the numbers of type I and type II fibers in the pectoralis minor decreased significantly from 94 and 62 to 74 and 44, respectively, between women in their 30's and 70's. Larsson et al. (1978) studied the effects of age on the distributions of muscle fibers in five different age groups: people aged in their 20's, 30's 40's, 50's and 60's. The groups in their 20's, 30's 40's, 50's and 60's showed type I fiber distributions of 40, 35, 50, 50, and 55%, respectively. The proportion of type I, slow twitch fibers, increased with age. On the other hand, the groups in their 20's, 30's 40's, 50's, and 60's showed 60, 65, 50, 50, and 45% of type IIA and IIB fibers. Aging caused a reduction in the proportion of type IIA and IIB, fast twitch fibers. The aging-related changes in muscle fiber distributions, as well as in the size and numbers of muscle fibers, can lower the capacity of muscle fibers or whole muscle to produce muscle force (Trappe et al., 2003; Thom et al., 2005), contributing to a reduction in peak muscle torque and power (Thom et al., 2005).

Other muscle properties, besides maximal force, could potentially be affected by aging. These properties include the active and passive length-tension relationships, the active and passive force-velocity relationship, the rates of muscle activation and deactivation, and series elasticity (Trappe et al., 2003; Alnaqeeb et al., 1984; Kent-Braun et al., 2000; Thelen, 2003). Muscle could be referred to as an aggregation of muscle fibers in parallel (Zajac, 1989). Both of the ends of muscle are connected to tendons, and muscle forces pass through the tendons and are received by segments of the body (Zajac, 1989). Tendon has a series elastic property and can generate passive force when stretched (Zajac, 1989). Active muscle force is produced by contracting muscle fibers, whereas passive muscle force is generated through stretch of muscle fibers. The capacity of muscle to generate active force depends on both its length and its shortening velocity. Active muscle force also changes at a finite rate in response to changes in neural excitation.

Aging could impact the active and passive length-tension relationships (Alnaqeeb et al., 1984; Kent-Braun et al., 2000; Thelen, 2003). Optimal muscle length (l_o^M) influences the production of active and passive forces (Brooks et al, 2005). l_o^M is the length of a muscle fiber at which actin and myosin filaments overlap each other optimally (Brooks et al, 2005). At l_o^M , maximal active force can be generated. When muscle fiber length is stretched beyond l_o^M , passive force starts to be produced. Although the studies of van Schaik et al. (1994) and Winegard et al. (1997) found no effect of aging on the length-tension relationship, the passive length-tension relationship can be affected by aging (Alnaqeeb et al., 1984; Kent-Braun et al., 2000; Thelen, 2003). The content of connective tissue increases inside muscle fibers of older people and rats (Alnaqeeb et al., 1984;

Kent-Braun et al., 2000; Thelen, 2003). An increase in connective tissue alters the elasticity of a muscle fiber; compliance of muscle fibers decreases and stiffness of muscle fibers increases (Alnaqeeb et al., 1984). Muscle fibers in older mice can generate passive forces at a faster rate than those in young mice when stretched (Alnaqeeb et al., 1984). Thus, aging causes an increase in passive muscle stiffness.

In addition to the length-tension relationships, the force-velocity relationships could potentially change with advancing age (Trappe et al., 2003). The force-velocity relationship is one of the factors that contribute to the amount of active force produced by a muscle (Brooks et al., 2005). The development of active force depends on contraction velocity (Brooks et al., 2005). Muscle active force is inversely proportional to the shortening velocity: the active force produced decreases as the shortening velocity increases, whereas the force produced increases when the muscle is lengthened (Brooks et al., 2005). This property also influences the maximum muscle shortening velocity for a given applied load. Consistent with the decreased proportion of Type II fibers in older age, Thompson and Brown (1999) found 30 month-old mice showed a 50 % reduction in unloaded contraction velocity compared to 24 month-old mice. In addition, Larsson et al. (1979) found that older adults aged over 50 years showed significantly lower maximum knee extension velocity than young adults aged in their 30's. Thus, aging causes changes in the force-velocity relationships (Larsson et al., 1979).

In addition to the above-stated changes in muscle properties, aging could cause decreases in the rates of muscle activation and deactivation (Vandervoort, &

McComas, 1986; Larsson & Salviati, 1989; Hunter et al., 1999). The sarcoplasmic reticulum (SR) influences the rates of muscle activation and deactivation (Heilmann & Pette, 1979; Larsson & Salviati, 1989; Hunter et al., 1999). The capacity for Ca^{2+} uptake by the SR can determine the times of muscle contraction (activation) and relaxation (deactivation). Aging can decrease the capacity of the SR to uptake Ca^{2+} (Larsson & Salviati, 1989; Hunter et al., 1999). Larsson and Salviati (1989) found that Ca^{2+} uptake activity decreased by 20 – 30% in fast twitch fibers of old mice, and old mice showed a 40% increase in the contraction time of an isometric twitch. Vandervoort and McComas (1986) studied the effect of aging on muscle contraction and relaxation times among persons aged in their 20's, 40's, 60's, 70's, and 80's. They found that, as age advanced, muscle contraction and relaxation time increased. Thus, the rates of muscle activation and deactivation decrease with advancing age (Vandervoort & McComas, 1986; Larsson & Salviati, 1989; Hunter et al., 1999).

Aging also could increase the series elasticity of tendon (Porter et al, 1997). Porter et al. (1997) studied the effect of aging on the passive ankle plantarflexion moment. They found that the passive ankle plantarflexion moment in older women was greater by 25 % than that in younger women. Thus, tendon stiffness increases with advancing age, and older adults can generate larger passive muscle forces than young adults.

In conclusion, aging influences the ability of older adults to generate muscle and tendon forces. Older adults show changes in maximal muscle force, in the passive length-tension relationship, in the rates of muscle activation and deactivation, and in series elasticity. The extent to which these changes influence the ability of older adults to recover balance following a backward step from a backward balance loss has not been investigated previously. This question can potentially be answered through the use of a musculoskeletal model in which the aging-related changes in muscle are simulated by adjusting parameter values of the model's musculotendon actuators. Forward dynamic simulations can be used to map the feasible region of balance recovery for young and older adults, allowing the influence of the aging-related loss of lower extremity strength to be quantified. This information might aid in the development of interventions to reduce the incidence of falls and fall-related injuries in older adults.