A BIOMECHANICAL ANALYSIS OF MAXIMUM VERTICAL JUMPS AND SIT TO STAND

A Thesis in
Kinesiology

by

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Abstract

The dissertation is made up of four reports, prepared in the format of journal manuscripts. Only the final report deals directly with the purpose of this dissertation, to develop a simulation of sit to stand that can be used to examine performance of the sit to stand in the elderly, and exploited to examine the influence of strength training on sit to stand performance in the elderly. The other three studies were used for the development of the simulation model. The first study investigated the effect of squat depth on jump height. Experimental subjects showed no difference in jump height in jumps from the deep and preferred positions. The optimal squat depth for the simulated jumps was the lowest position the model was able to jump from. Because jumping from a deep squat is rarely practiced, it is unlikely that these jumps are optimally coordinated.

The second study examined how arm swing contributes to maximum vertical jump height. There was an increase in jump height with arm swing. The vertical velocity of the arms relative to the shoulder at takeoff was large enough to account for the performance enhancement. The work produced by the shoulder extensors was more than sufficient to explain the increase in jump performance. The work produced by the lower extremity muscles was slightly reduced in jumps with arm swing.

The third study investigated the effect of chair height on sit to stand mechanics. It was found that forward trunk movement prior to seat off decreased as seat height decreased. This is the result of a higher initial hip angle in trials from low seat heights and forward trunk movement moving the hip extensors to a less favorable region of the force-length curve. The reduction in trunk movement prior to seat off increases the load on the knee extensors, which are already heavily burdened as a result of high initial knee angles.

The fourth study examined the effect of strength training individual muscle groups on sit to stand performance. It was found that strength training the hip extensors and the knee extensors produced similar improvements in performance. These improvements were approximately one half of the improvement seen when training all muscles. There was potentially better control of the movement when all muscle groups
were trained; as demonstrated by a reduction in peak joint extension angular velocities and a change in the timing of the maximum horizontal momentum. The results of these simulations suggest it is important to train all of the muscles rather than one specific muscle group for improving sit to stand performance.
## Table of Contents

List of Figures xi  
List of Tables xii  
Acknowledgements xv  

1. Introduction  
1.1. General Background 2  
1.2. Purpose of the Study 3  
1.3. Overview of the Study 3  
1.4. Specific Aims 6  
1.5. Dissertation Structure 6  
1.6. References 7  

2. Literature Review 9  
2.1. Relevant studies of jumping 10  
  2.1.1. The effect of initial position on vertical jump performance 10  
  2.1.2. The effect of arm swing on vertical jump performance 14  
2.2. Relevant studies of sit to stand 20  
  2.2.1. The effect of age on sit to stand performance 20  
  2.2.2. The effect of chair height on sit to stand performance 24  
  2.2.3. The effect of muscular strength on sit to stand performance 27  
2.3. Effects of aging on skeletal muscle 30  
  2.3.1. Decreased force output 30  
  2.3.2. Increased twitch times 32  
  2.3.3. Decreased maximum shortening velocity 34  
  2.3.4. Effects of aging on tendon 36  
  2.3.5. Effects of strength training on aging muscle 37
2.4. Direct dynamics simulations of jumping and sit to stand  
2.4.1. Direct dynamics simulations of jumping  
2.4.2. Direct dynamics simulations of sit to stand  
2.5. Summary  
2.6. References  

3. The influence of squat depth on maximal vertical jump performance  
3.1. Abstract  
3.2. Introduction  
3.3. Methods  
  3.3.1. Experimental analysis  
  3.3.2. Model analysis  
3.4. Results  
  3.4.1. Experimental analysis  
  3.4.2. Model analysis  
3.5. Discussion  
3.6. References  

4. The contribution of arm swing to maximum vertical jumps  
4.1 Abstract  
4.2 Introduction  
4.2 Methods  
  4.2.1 Experimental analysis  
  4.2.2 Model analysis  
  4.2.3 Induced Accelerations Analysis  
4.3 Results  
  4.3.1 Experimental analysis  
  4.3.2 Induced acceleration analysis  

vi
List of Figures

**Figure 3.1.**
Vertical ground reaction force for jumps from the deep position (top) and preferred position (bottom).

**Figure 3.2.**
Stick figure comparing the model to an experimental subject throughout a jump at 0.05 s intervals. Top – subject. Bottom – mode.

**Figure 3.3.**
Neural excitation for the biarticular muscles during both simulated jumps.

**Figure 4.1.**
Acceleration induced at the ankle, knee, and hip joints by the moment associated with arm swing during a jump for a typical subject.

**Figure 4.2.**
Stick figures of a typical experimental subject performing a jump with arm swing (lower sequence), and simulated jump (upper sequence).

**Figure 5.1.**
Joint angle definitions.

**Figure 5.2.**
Vertical ground reaction force for a typical subject (top) from Burgess (2003), and the model (bottom) performing the sit to stand from a seat height of 42 cm. Vertical lines indicate when upright stance was achieved.

**Figure 6.1.**
Stick figure of the model standing without any strength increases.

**Figure A.1.**
The active state of a muscle plotted against time for a trise of 0.05 and a tfall of .020.

**Figure A.2.**
Normalized force-length relationship of a muscle

**Figure A.3.**
A normalized force-velocity relationship for a muscle.

**Figure A.4.**
A normalized strain versus muscle force plot for a tendon.
Figure A.5.
The flow of variables in the muscle model.

Figure B.1.
Example of a two-link system connected to the ground by a pin joint with another point joint connecting the segments

Figure B.2.
Flow chart summarizing the direct dynamics simulation.

Figure C.1.
Muscle length plotted against ankle angle for muscles crossing the ankle joint.

Figure C.2.
Muscle length plotted against knee angle for muscles crossing the knee joint.

Figure C.3.
Muscle length plotted against hip angle for muscles crossing the hip joint.

Figure C.4.
Muscle length plotted against shoulder angle for muscles crossing the shoulder joint.

Figure C.5.
Moment arm plotted against ankle angle for muscles crossing the ankle joint. Note: Soleus and Gastrocnemius have the same moment arm

Figure C.6.
Moment arm plotted against knee angle for muscles crossing the knee joint.

Figure C.7.
Moment arm plotted against hip angle for muscles crossing the hip joint.

Figure C.8.
Moment arm plotted against shoulder angle for muscles crossing the shoulder joint.

Figure C.9.
Moment versus angle curves for the young model plantarflexors.

Figure C.10.
Moment versus angle curves for the young model knee extensors.

Figure C.11.
Moment versus angle curves for the young model knee flexors.
Figure C.12.  
Moment versus angle curves for the young model hip extensors.

Figure C.13.  
Moment versus angle curves for the young model shoulder extensors.

Figure C.14.  
Moment versus angle curves for the young model shoulder flexors.

Figure C.15.  
Moment versus angle curves for the old model plantarflexors.

Figure C.16.  
Moment versus angle curves for the old model dorsiflexors.

Figure C.17.  
Moment versus angle curves for the old model knee extensors.

Figure C.18.  
Moment versus angle curves for the old model knee flexors.

Figure C.19.  
Moment versus angle curves for the old model hip extensors.

Figure C.20.  
Moment versus angle curves for the old model hip flexors.

Figure E.1.  
Muscle force plotted against mass position for the maximal activation movement.

Figure E.2.  
Muscle force plotted against mass position for the movement minimizing the rate change of muscle force.

Figure E.3.  
Muscle force plotted against mass position for the movement minimizing the time of the movement.
### List of Tables

**Table 3.1.**  
A summary of the muscle model parameters.  

**Table 3.2.**  
Means and standard deviations of the initial squat positions in the jumps for each of the conditions.  

**Table 3.3.**  
Mean and standard deviations of jump heights and jump times for each of the conditions.  

**Table 3.4.**  
Normalized maximum joint moments and their times for each of the conditions.  

**Table 3.5.**  
Jump height and time for the simulated jumps.  

**Table 4.1.**  
A summary of the muscle model parameters.  

**Table 4.2.**  
Means (± standard deviations) of ground contact time and jump height for the jumps with and without arm swing.  

**Table 4.3.**  
Means (± standard deviations) of the initial and final and joint angles for the jumps with and without arm swing.  

**Table 4.4.**  
Means (± standard deviations) of the maximum joint angular velocities during the jumps with and without arm swing.  

**Table 4.5.**  
Means (± standard deviations) of the normalized maximum joint moments, and the times at which they occurred during the jumps with and without arm swing. Times are presented as time before takeoff.  

**Table 4.6.**  
Jump height and ground contact time for the simulated jump with and without arm swing.
Table 4.7.  
Maximum muscle shortening velocities for the simulated jumps with and without arm swing.

Table 4.8.  
Maximum muscle force for the simulated jumps with and without arm swing.

Table 4.9.  
Work produced by the muscles for the simulated jumps with and without arm swing.

Table 5.1.  
A summary of the muscle model parameters.

Table 5.2.  
Mean characteristics of the ground reaction force curves from Burgess (2003) and from the model from a seat height of 42 cm.  Forces are normalized with respect to body weight.

Table 5.3.  
Time in contact with the seat, and time to achieve upright stance for the simulations from the three seat heights.

Table 5.4.  
Hip and trunk angles at seat off for the simulations from the three seat heights.

Table 5.5.  
Maximum hip flexion velocity and its timing for the simulations from the three seat heights.

Table 5.6.  
Maximum hip and knee extension velocities for the simulations from the three seat heights.

Table 5.7.  
The maximum joint moments for the simulations from the three seat heights presented in Newton meters and as a percentage of maximum moment available at given joint angles and velocities.

Table 6.1.  
Time to rise for the simulation model compared with means (±standard deviations) reported in the literature for maximum speed sit to stands in the elderly.
Table 6.2.  114
Time to rise and time to seat off for the model of sit to stand performance before strength training and as consequence of strength training.

Table 6.3.  115
Trunk flexion angular velocity at seat off during the sit to stand simulations, before strength training and as consequence of strength training.

Table 6.4.  115
Maximum hip and knee extension angular velocities during the sit to stand simulations, before strength training and as consequence of strength training.

Table 6.5.  116
Maximum center of mass momentum and timing during the sit to stand simulations, before strength training and as consequence of strength training.

Table 6.6.  117
Maximum muscle active states during the sit to stand simulations, before strength training and as consequence of strength training.

Table C.1.  155
Body segment inertial parameters used in chapters 3 and 4 to represent a young subject in the simulations of jumping.

Table C.2.  155
Body segment inertial parameters used in chapters 5 and 6 to represent an older subject in the simulations of sit to stand.

Table C.3.  157
Parameters for determining muscle length and moment arm.

Table E.1.  186
Muscle model parameters.

Table E.2.  188
Maximum force enhancement as a consequence of tendon elasticity in each of the movements examined.
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CHAPTER 1
INTRODUCTION

1.1. General Background
1.2. Purpose of Study
1.3. Overview of Study
1.4. Specific Aims
1.5. Dissertation Structure
1.6. References
1.1. General Background

With an increasing proportion of the America population reaching old age, research improving functional ability in activities of daily living has become an important research focus. It has been shown that rising from a seated position is a challenging activity of daily living (Rodosky et al., 1989). Dawson et al. (1987) reported that more than two million people in the United States over the age of 64 years had difficulty in performing sit to stand. In fact many elderly are unable to rise from a chair without assistance (Alexander et al., 2000). Rising from a chair is clearly a precursor to gait and therefore is a significant determinant of independence.

There are a number of reasons why rising from a chair is a difficult task. There is clear evidence of changes in physical capabilities with age (e.g. muscular strength - Frontera et al., 1991; joint mobility - Einkauf et al., 1987), as well as changes in the way in which movements are coordinated (e.g. Chao, 1986). Standing up causes a balance perturbation, which increases the risk of falling. Because falls can be so devastating, fear of falling is an important component of the difficulty an elderly individual has in standing from a chair (Kressig et al., 2001). Strength loss in the muscles crossing the ankle joint may worsen balance problems further complicating sit to stand (Winter et al., 2001). Hughes et al. (1996) reported that in the functionally impaired elderly the peak moment at the knee joint was 97% of maximum isometric knee moment during sit to stand when chair height was low. Riley et al. (1997) examined failed sit to stand for 13 elderly subjects. Their analysis suggested that the magnitude and timing of the momentum generation and knee extensor moments were the primary reasons for failure. Comparing strength levels and the kinematics and kinetics of young and elderly populations Gross et al. (1998) concluded that strengthening the hip musculature could be important to maintain STS ability in the elderly.

It is known that there is a progressive loss of strength associated with aging (Young et al., 1985). Fortunately, strength training has been shown to cause significant strength gains in the elderly (Fiatarone et. al., 1990). This project will aim via a simulation model of elderly subjects performing the sit to stand to provide
recommendation of the appropriate strength training an elderly subject may perform to maintain this important skill. Because getting people to participate and adhere to a strength training program can be difficult, it is hoped that it will be possible to recommend minimal interventions required for the elderly to maintain this important skill.

1.2 Purpose of Study

The purpose of this study was to develop a simulation of sit to stand that can be used to examine performance of the sit to stand in the elderly, and exploited to examine the influence of strength training on sit to stand performance in the elderly.

1.3 Overview of Study

The dissertation is made up of four reports, prepared in the format of journal manuscripts. Only the final report deals directly with the purpose of this dissertation. The other three studies were used for the development of the simulation model. This section reviews each of these reports and provides the justification for the first three reports inclusion in the dissertation. The first two studies are examinations of a young healthy population performing maximum vertical jumps. These studies include both experimental and simulation work. The last two studies are of a healthy elderly population performing sit to stand. These two studies are entirely composed of simulation work and are compared with other investigators’ experimental work for validation.

The first study examined the effect of initial position on jumping ability. Squatting as deeply as possible before performing a jump allows for increased time to apply and develop force. It is therefore logical to deduce that jump height may be increased by squatting deeply, however when observing athletes perform maximum jumps, they do not do this. This study performed an investigation of the effect of squat depth on jump height using experimental methods and a simulation model. It was hypothesized that the subjects would jump highest from their self selected squat
depth and that the simulation would jump highest from a very deep squat. The difference being a result of the experimental subjects not being optimally coordinated when jumping from a deep squat.

The first study was selected as an initial test of the simulation model. The activity of jumping was chosen because it is uses the same joints, and muscles operating through a similar range of motion as the sit to stand. Maximum vertical jumping compared with the sit to stand and has a very well defined performance criterion. In maximum vertical jumping the goal is to raise the center of mass of the body as high as possible. The simulation model maximizes gravitational potential energy and vertical kinetic energy at the instant of takeoff to accomplish this. In sit to stand the performance goal is not clear and therefore is a poor initial test of the simulation model.

The second study examined the effect of arm swing on vertical jumping ability. In studies of jumping subjects are often asked to perform the jumps without swinging the arms. This simplifies analysis and data collection. However, evidence exists that arm swing increases jump height. The mechanisms by which this is accomplished are not clearly understood. Several mechanisms have been proposed as to the source of the improvement in jump height, and this study will investigate the contribution of each of these mechanisms. It was hypothesized that each of the proposed mechanisms contribute, but that the largest contribution to jump height would be from the kinetic energy of the arms at takeoff.

The second study was performed because one possibility for improving sit to stand ability is to swing the arms while performing the task. Maximum vertical jumping was used as a surrogate for sit to stand as the performance criterion is well known for maximum jumping; whereas it is not known for sit to stand. If the study of jumping suggests this is a useful strategy in sit to stand, trials with elderly subjects would be warranted.

The third study investigated the effect of chair height on sit to stand ability in the elderly. This task has been fairly extensively studied experimentally, but a simulation study has not yet been performed. It has been shown that decreasing chair height makes it more difficult to stand up (e.g., Arborelius et al., 1992). The source of this increased difficulty, in performing the sit to stand by the elderly, has not been
thoroughly investigated. Increased trunk anterior displacement (Weiner et al., 1993) and increased trunk flexion velocity (Schenkman et al., 1996) while in contact with the chair have been reported accompanying reduced chair height, however there are contradictory reports as to the reason for these changes. They have been attributed to either an effort to improve stability to cope with possible balance disorders (Alexander et al., 1991) or to reduce the load on the knee extensors to cope with strength loss (Schenkman et al., 1996). The simulation model used in this study takes into account the strength loss that accompanies aging, however there is no balance disorder modeled. The model allows for the effects of strength loss to be separated from the effects caused by balance disorders, because any changes that are seen are a result of reduced strength. It is hypothesized that the simulation model will increase forward trunk movement, both increased velocity and displacement, as seat height decreases. These changes will be attributable to an effort to reduce the activity of the knee extensors as the model has no component reflecting any balance disorder.

The third study was conducted as a first test of the simulation performing sit to stand. A validation of a sit to stand model based on elderly subject parameters has not as of yet been performed. The majority of experimental work on sit to stand has been done at a self selected speed, therefore it was desirable to conduct a first validation of this model at a self selected speed.

The fourth study investigated the effects of different strength training programs on sit to stand ability. Developing a simple strength training program to improve sit to stand performance is the overall goal of the dissertation. Testing strength training programs experimentally is very time consuming, however testing them using a simulation model is relatively easy and is an excellent precursor to experimental work. This study will allow several different training programs to be evaluated; a program can then be recommended based on simplicity and functional improvement. It was hypothesized that training the knee extensors would provide the largest reduction in time to complete the movement.
1.4 Specific Aims

The following specific aims have been developed,
1. To develop a muscle model driven direct dynamics model of sit to stand.
2. To examine the validity of this model by comparing its output with experimental data from subjects jumping and performing sit to stand.
3. To use the model to examine the influence of strengthening different muscle groups in the model on the performance of the sit to stand.
4. To provide recommendations as to the most appropriate muscle group(s) to strengthen to help the elderly maintain performance of the sit to stand, or regain this important skill.

1.5 Dissertation Structure

Chapter two reviews the relevant literature for the entire dissertation. Chapters three, four, five and six are reports of experimental and modeling work, and are written in the format of journal manuscripts. Chapter seven is the discussion of the work performed. Due to space restrictions in journal manuscripts more details of the models used in this study are provided in the appendices. Specifically, appendix A presents the muscle model; the rigid body model is presented in appendix B; appendix C contains the model parameters; the optimal controller is discussed in appendix D; and appendix E details an experiment examining the effect of tendon on muscle force output. Relevant references are presented at the end of each chapter. The final section in the dissertation contains all of the references used throughout.
1.6. References


CHAPTER 2

Literature Review

2.1. Relevant studies of jumping
   2.1.1. The effect of initial position on vertical jump performance
   2.1.2. The effect of arm swing on vertical jump performance

2.2. Relevant studies of sit to stand
   2.2.1. The effect of age on sit to stand performance
   2.2.2. The effect of chair height on sit to stand performance
   2.2.3. The effect of muscular strength on sit to stand performance

2.3. Effects of aging on skeletal muscle
   2.3.1. Decreased force output
   2.3.2. Increased twitch times
   2.3.3. Decreased maximum shortening velocity
   2.3.4. Effects of aging on tendon
   2.3.5. Effects of strength training on aging muscle

2.4. Direct dynamics simulations of jumping and sit to stand
   2.4.1. Direct dynamics simulations of jumping
   2.4.2. Direct dynamics simulations of sit to stand

2.5. Summary
This chapter is comprised of five sections. Section 2.1 provides a review of relevant studies of jumping. Section 2.2 reviews studies of sit to stand relating to chair height and muscular strength. The effects of aging on skeletal muscle are reviewed in section 2.3. Section 2.4 covers the use of whole body direct dynamics simulations in the study of jumping and sit to stand. The final section provides a summary.

2.1. Relevant studies of jumping

Many biomechanical aspects of jumping have been studied. Relevant to this dissertation are studies on the effect of initial position and the effect of arm swing on jump performance. Section 2.1.1 covers studies investigating the effect of initial position on jumping performance. Section 2.1.2 reviews studies of the effect of arm swing on jumping performance.

2.1.1. The effect of initial position on vertical jump performance

There have been no published studies that have explicitly examined the effect of initial position on vertical jumping performance experimentally. Bobbert et al. (1996) presented some data on this in a paper examining the effects of countermovements on jump performance. There have been two studies that have performed simulations of jumps from different initial postures (Selbie and Caldwell, 1996; van Soest et al, 1994).

Bobbert et al. (1996) asked subjects to perform countermovement jumps and squat jumps from preferred positions. Squat jumps were also performed from a position matching the bottom of jumps performed with a countermovement, and from a deep squat position. They found that countermovement jumps were significantly higher than all three squat jumps. There were no statistical differences between the squat jumps. The authors theorized that countermovement jumps are higher than squat jumps, because there is increased time for muscle force generation. This same theory would suggest that jumps from deeper squats would be higher, than those from
more shallow squats, because as the depth of the squat increases the time to complete the jump increases as well; but this was not the case for their results.

The effect of initial position was not the focus of their paper, therefore few comparisons were made between squat jump conditions. This makes it difficult to hypothesize why there was no difference in squat jump heights, however the positions that were chosen could be the reason. The preferred depth squat jump was the most upright of the three conditions. The squat jump from the position matching the countermovement was slightly deeper. It is possible this jump was not higher because it was difficult for subjects to match this starting position. The deep squat jump was performed from as deep a squat as possible. It is possible that some muscles were stretched beyond where they can exert forces in this posture (Sale et al., 1982; Gravel et al., 1987).

Selbie and Caldwell (1996) used a simple simulation model to investigate the effect of initial position on countermovement jump height. They simulated jumps from 125 different starting positions ranging from upright to semi-squat. In order to perform the large number of simulations, a simple model was used to keep simulation time reasonable. The model was actuated by moment generators at the ankle, knee and hip joints. These actuators had moment - angular velocity properties, but not moment – joint angle properties. Muscle activation was modeled as an exponential rise to maximal activation from an onset time that was determined by optimization. The optimization algorithm maximized jump height by selecting the timings of actuator activations. Once an actuator reached maximal activation it remained fully active until the end of the jump.

Selbie and Caldwell found that optimal muscle activations were such that the body was in a similar position at the bottom of the countermovement for all of the starting positions. The initial position had only small effects on vertical jump height in most conditions. The center of mass of the model reached between 1.37 and 1.44 meters for 108 of the 125 positions examined. There were much larger variations in the horizontal distance traveled. They then simulated the jumps constraining them so that there was no horizontal component to the jump flight phase. For approximately 75% of the jumps under this new condition there was still very little change in jump performance, however there were some positions where jump performance was
significantly decreased. Unfortunately the range of jump heights is not given for these additional simulations.

The model used in this study was flawed in several ways. The moment a muscle can exert varies throughout the range of motion in two ways: the moment arm of the muscle changes with joint angle and the force it can generate is a function of its length, so therefore varies with joint angle. Selbie and Caldwell’s model does not take these factors into account. Adding a moment angle relationship could have been done without significantly increasing simulation time. There are also three simplifications used in this model to decrease simulation time that may not be appropriate for countermovement jumping. Muscle activation dynamics similar to the one used here have been used elsewhere to study squat jumps (e.g. van Soest et al, 1993), however it is probably not appropriate for simulating jumps with a countermovement. During the propulsive phase of the jump assuming muscles are maximally activated is reasonable (Pandy et al., 1990) however during the countermovement muscles are likely only partially active, otherwise the center of mass would not be lowered. The model also lacked biarticular muscles and series elastic elements, both of which have been shown to be important in jumping (van Soest et al, 1993; Alexander 1995).

Because this model is probably oversimplified and the bottom most position in the jumps was very similar for all jumps irrespective of initial position, this study does not provide much insight into the effect of initial squat depth on jump performance.

Van Soest et al. (1994) claimed that initial position had no effect on jump height. They make this claim based on unpublished data. They hypothesized that there is a single muscle activation pattern that is applied to maximal jumps from all starting postures. They tested this hypothesis using a direct dynamics simulation model of jumping. The gluteal muscles, hamstrings, vasti, rectus femoris, soleus, and gastrocnemius were all modeled. Each muscle consisted of a contractile element, a series elastic element, and a parallel elastic element. The contractile element had force-length and force-velocity properties. The elastic elements had a non-linear stress-strain relationship. Muscle activation was described by a first order differential equation and is related to neural excitation. Neural excitation was initially set at the
level required to produce static equilibrium in the initial squat position. During the jump, neural excitation could be either remain at this value or increase in an exponential fashion to its maximal value. Once maximal the neural excitation remained maximal throughout the jump.

There are three parts to their study. First they determined the optimal muscle activation patterns for five different starting postures. They then applied the optimal activation pattern from one position to jumps from two other starting postures. The authors do not say how the positions were selected. Finally they determined the global optimal muscle activation pattern that could maximize jump height at all positions.

Jumps simulated using activation patterns optimized for each starting posture matched their experimental kinematics fairly well. Jump height from the experimental subjects was 44.7 cm. Jump height from the position that matched the mean position used by the subjects was 39 cm. Although they do not discuss it, the height of the jump increased as the depth of squat increased for their model. In the deepest position their simulated jump height was 42.6 cm and from the most upright posture the jump height was 30.6 cm.

When applying the muscle activation patterns for the position matching the experimental data to jumps from other positions, jump height was decreased. Jump height was reduced by over 50 percent in jumps from the deepest squat. In addition to jump height being reduced there were kinematic changes as well, for example in jumps from the deepest squat there was hyperextension at the hip and very little plantar flexion of the ankle at takeoff.

The global optimal activation pattern produced jumps that were similar to the optimal performances. There was a decrease in jump height ranging from 0.3 cm to 4.8 cm. The kinematics of the jumps were very similar to the optimal jumps. Van Soest and colleagues believe this supports their theory that there is a single muscle activation pattern that is applied to jumps from a range of starting postures.

While the loss of jump height caused by using the global optimal muscle activation is relatively small, it is 3.8 cm from the initial position selected by the subjects. It seems unlikely that the activation pattern selected would be non-optimal for the preferred position.
There is one assumption in this study, which may influence interpretation of their results. Simplifying muscle activation in the manner done in this model may be acceptable for studying many aspects of jumping, however when the area of interest is muscle activation patterns this seems suspect. A global optimal activation scheme may be possible with the simplified muscle activation these use, but not possible for a more complex scheme that is physiologically realistic. They also present no reasons why their simulation jumps higher as squat depth increases even though their unpublished data suggest that people do not.

To date the literature has not thoroughly addressed the question of the effect of squat depth on jump height. Bobbert et al. (1990) reported no difference in jump height from several different postures, however they did not analyze this data farther. Van Soest et al. (1994) claim that squat depth has no effect on jump height, however their simulation results show otherwise. Selbie and Caldwell (1994) showed initial position had only small effects on countermovement jump height, however the position at the bottom of the countermovement was similar for all their simulated jumps.

2.1.2. The effect of arm swing on vertical jump performance

There are three potential mechanisms by which arm swing can improve jump performance. First, swinging the arms raises the center of mass of the body, so takeoff occurs with the body’s center of mass higher. The second mechanism is that the momentum of the arms caused by arm swing contributes to the overall momentum of the body. The third is that arm swing can change the conditions at other joints improving conditions for muscle force production.

Khalid et al. (1989) performed a study to quantify the effect of arm swing on vertical jump performance. Twenty-eight subjects performed jumps under four different conditions, squat jumps with and without arms, and countermovement jumps with and without arms. Jump height was determined from takeoff velocity obtained by integrating vertical ground reaction force recordings. They found a 3.8 cm (11 percent) increase in squat jump performance when using the arms, and a 3.6cm (10 percent) increase in countermovement jump performance. This is about twice as large
as the improvement in performance when they compared squat jumps with
countermovement jumps. They do not report the height of the center of mass at
takeoff. The authors hypothesize that the improvement in performance caused by arm
swing is a result of improved re-utilization of elastic energy, however they do not
provide any support for this. They do not identify a potential site for this energy
storage.

Shetty and Etnyre (1989) investigated the effect of arm swing on squat jump
performance. Eighteen subjects performed squat jumps with and without arm swing.
Jump height was determined from takeoff velocity obtained by integrating vertical
ground reaction force recordings. They found a 5.1 cm (25 percent) increase in squat
jump performance when using the arms. They also did not report the height of the
center of mass at takeoff. They did not attempt to explain the mechanism for the
improvement in performance.

Harman et al. (1990) performed a very similar study to Khalid et al. (1989).
They wanted to investigate the effects of arm swing and countermovements on
vertical jumping performance. They had 18 subjects perform jumps under four
different conditions, squat jumps with and without arms, and countermovement jumps
with and without arms. Jump height was determined from takeoff velocity obtained
by integrating vertical ground reaction force recordings. The displacement of the
center of mass prior to takeoff was also measured. They found a 5.8 cm (21 percent)
increase in squat jump height when using the arms and a 6.2 cm (21 percent) increase
in countermovement jump performance. Countermovements improved performance
by 1.7 cm (6.2 percent) when not using the arms and 2.1 cm (3.0 percent) when using
the arms. When swinging the arms takeoff occurred with the center of mass 4.4 cm
higher in the squat jumps and 4.7 cm higher in the countermovement jumps.

The authors state that the arms decelerate relative to the rest of the body near
takeoff. The authors argue that if the arms return to a zero velocity relative to the rest
of the body, the momentum of the arms is not the cause of the increase in takeoff
velocity. They hypothesize that the upward acceleration of the arms produces a
downward force on the rest of the body. This would reduce the velocity of shortening
of the muscles of the lower extremity and that this occurs when these muscles are in
an advantageous position to produce force. This would then explain how arm swing
improves performance. This argument is flawed, because the arms may not reach zero velocity relative to the rest of the body and in fact increase the total vertical momentum of the center of mass. Because they do not report the velocity of the arms relative to the body, it is unclear how large a contribution to performance is made by the momentum of the arms if at all. The mechanism Harman and colleagues propose would also increase the time of force application, which could improve performance as well (Bobbert et al., 1996).

The authors spend very little or no time discussing two potentially important results from their study. First, the increase in jump height caused by raising the center of mass higher in the body at take off is nearly as large as the increase in height after takeoff. This is a clear advantage of performing jumps with an arm swing. Secondly, there may have been a statistical interaction effect between arm swing and countermovements, however the authors did not perform any statistical analysis on this. There was no interaction found by Khalid et al. (1989) and would certainly be an interesting finding if it were shown here.

Lees and Barton (1996) developed a procedure to determine the effects of the motion of free limbs to vertical velocity of the center of mass of sporting movements. Specifically they were interested in arm movement in jumping and running and the movement of the free leg in running. They proposed that the contribution of a limb to the total linear momentum at takeoff should be measured by examining the relative linear momentum between the limb and the trunk. The momentum a limb can contribute is the change in the relative momentum that occurs during the movement. They present a simple example of a two mass system to illustrate this analysis procedure.

They applied this procedure to a one arm swing squat jump performed by an athlete. The jump was filmed and the velocity of the segments was determined from differentiation of position data. The masses of the segments were determined using the data of Dempster (1955). The total momentum of the body at takeoff was found to be 244 Ns. This would give a jump height of 40 cm. The contribution of the arms to this was found to be 30.9 Ns. If the jump was performed without this momentum jump height would have only been 30.6 cm.
This study does suggest that the momentum of the arms is likely a large source of the improvement in performance associated with arm swing. However, these results have to be looked at with some skepticism. The effect of the arms on jump height in this study is larger than the improvement in performance seen in other studies reviewed in this section. Only one subject was used in this study and applying Dempster’s data to athletes can cause large error, so the momentum of the arms could have substantial error (Sprigings et al., 1987). The momentum of the arms could have been caused by moments at other joints, as well as the joints of the upper limb.

Feltner et al. (1999) investigated the effects of arm swing on countermovement jumps in athletes. He had 25 (14 males, 11 females) volleyball players from Pepperdine University perform countermovement jumps with and without arm swing. Force plate and motion analysis data were recorded. Jump height was determined from takeoff velocity obtained by integrating vertical ground reaction force recordings. Jumps performed with arm swing were on average 8.2 cm (27 percent) higher than those without. The center of mass was 6.1 cm higher at takeoff in jumps performed with arm swing.

The jump was then broken into phases for analysis. Period A was defined as the time from movement initiation to peak negative vertical velocity of the center of mass. Period B was defined as the time from peak negative vertical velocity to the lowest position of the center of mass during the countermovement. Period C lasted until maximum positive vertical velocity. Period D went from the time of maximum positive vertical velocity until takeoff. Most of the analysis concentrated on period C.

At the start of period C the no arm swing jumps showed increased knee and hip flexion compared with arm swing jumps. The authors did not speculate on the effect this would have on performance. The key finding of the paper is that the rate of hip extension was slower during period C in jumps performed with arm swing. The authors attribute this to a downward reaction force at the shoulder cause by upward acceleration of the arms. They speculate that this would put the hip extensors in physiological conditions that favored generation of large moment. This could be decreased shortening velocity or increased time to develop force.
While it could be possible that the reaction force produced at the shoulder could act to slow hip extension, the authors provide no real evidence for this. They also do not measure the momentum of the arms at takeoff.

The role of arm swing has also been studied in the long jump. Ashby and Heegaard (2002) examined the role of arm swing in the standing long jump. They asked three unskilled subjects to perform standing long jumps from a force plate. Three jumps were performed with arm swing, and three jumps were performed with the arms crossed in front of the chest. Motion analysis data was also recorded. Performance was defined as the horizontal displacement of the toe.

Jumps performed with arm swing were 36 cm (21.2 percent) longer than jumps without use of the arms. The resultant takeoff velocity was larger in jumps with arm swing (3.32 m/s versus 2.95 m/s), however the angle of takeoff was decreased from 40.2° to 38.6° when using arm swing. The center of mass was shifted 8 cm forward at takeoff when swinging the arms. The toes were 2 cm farther in front of the center of mass at landing in jumps with arm swing. Both of these differences were statistically significant.

The majority of the improvement in performance was caused by increasing takeoff velocity. The authors believe this increase comes from the combination of the momentum of the arms and increased force generation by the muscles of the lower extremities as a result of lower contraction velocities. They do not provide any evidence as to the relative contributions each of these mechanisms. The authors also discuss another possible mechanism for improvement in takeoff velocity. The arms can be used to control the angular momentum of the rest of the body during flight. In the standing long jump takeoff occurs with the feet behind the center of mass. During flight the body rotates so that landing occurs with the feet in front of the center of mass. If the arms are rotated during flight, the body will rotate to conserve angular momentum. Without this mechanism takeoff must occur with the correct amount of angular momentum to rotate the body and this would likely result in a decrease in takeoff velocity.

In ancient Greece long jumpers jump while holding weights called halteres. Minetti and Ardigo (2002) investigated how adding mass to the arms could improve performance. They did this with a simple simulation model and a four subject
experimental study. The simulation used was a simple four link system with moment
generators with muscle-like properties at the shoulder and knee. Vertical jumps were
simulated with masses ranging from 0 to 20 kg added to the arms. Subjects
performed vertical jumps with halteres ranging in mass from 0 to 17 kg.

Minetti and Ardigo predicted that the change in the location of the center of
mass at takeoff for a long jump to be 8 cm higher and 7 cm more anterior when
jumping with halteres. For a jump of 3 m this would increase jump distance by about
17 cm. The results from their simulation showed a 2 percent increase in takeoff
velocity when jumping with a 6 kg haltere. This was the largest increase seen. The
results from their experimental work showed this to be the optimal mass for their
subjects as well. Their subjects were able to jump 6 percent higher with a 6 kg
haltere.

This work shows the improvement in performance associated with arm swing
can be enhanced by adding mass to the arms. This will increase the changes in the
center of mass location at takeoff as well as improving takeoff velocity. The
improvement in takeoff velocity could be a result of the arms having more mass and
therefore more momentum at takeoff or could be the result of a higher shoulder
moment producing more of a downward force slowing extension of the lower
extremity muscles.

In summary, the effect of arm swing on vertical jump performance has been
fairly well quantified. The center of mass has been shown to be 4.4 to 5.2 cm higher
at takeoff in jumps with arm swing compared with jumps performed without swinging
the arms (Khalid et al., 1989; Shetty and Etnyre, 1989; Harman et al., 1990; Feltner et
al., 1999). The increase in takeoff velocity has been shown to increase post takeoff
jump height by 3.6 to 8.2 cm (Khalid et al., 1989; Shetty and Etnyre, 1989; Harman et
al., 1990; Feltner et al., 1999). The mechanism behind the improvement in takeoff
velocity is unclear. Lees and Barton (1996) have suggested this could be a result of
increased momentum of the arms. Harman et al. (1990) and Feltner (1999) have
theorized that arm swing could act to slow the extension of the legs early in the
movement allowing higher muscle forces to be generated.
2.2. Relevant studies of sit to stand

The mechanics of rising from a chair has received a lot of attention over the past fifteen years. The ability to perform sit to stand is an important factor in independent living. This section of the review will concentrate on studies that have focused on the effect of chair height, age, and muscular strength on performance of sit to stand. Section 2.2.1 covers studies investigating the effect of age on sit to stand performance. Section 2.2.2 reviews studies of the effect of chair height on sit to stand performance. Section 2.2.3 reviews studies of the effect of strength on sit to stand performance.

2.2.1. The effect of age on sit to stand performance

Ikeda et al. (1991) investigated the effect of age on the kinematics and dynamics of a constrained chair rise. They asked nine (6 males, 3 females) healthy elderly subjects to perform sit to stand in pace with a metronome set at 52 beats per minute, therefore subjects completed the task in approximately 1.15 seconds. The subjects ranged in age from 61 to 74 years. The subjects had no diagnosed musculoskeletal disorders and were independent in home and community activities. Data were compared with the results from a previous study (Schenkman et al., 1990). This study used the same protocol with nine healthy young women. Kinematic data were recorded and force measurements were taken from each foot. Performance was broken into three phases for analysis. The first phase began with movement initiation and ended with buttocks lift off. The second phase lasted until maximum dorsiflexion and the third phase went until hip extension velocity became zero.

The results from the elderly were very similar for phases one and three to what had been measured in the young. The two groups were most different during the second phase. This phase was shorter in duration for the elderly. Graphically the kinematics during phase two were very consistent in the young subjects, but showed greater variability in the elderly subjects, however the authors did not quantify this. The largest difference was in head position. In the young subjects the head rotated...
relative to the trunk to face forward. In the elderly the head stayed close to fixed relative to the trunk, therefore the elderly subjects were looking down during phase two of the stand.

It is not surprising that there are few differences between these groups, because the time to complete the task was constrained and both groups had to go through similar ranges of motion. The elderly group used in this study was healthy, and younger than most people who have trouble rising from a chair (Alexander et al., 1991). There still may have been more differences if the movement was performed at a self-selected pace. The differences in head position could be important, as this could affect balance, however it is possible that these differences were due to the fact that the studies were conducted at different times and instructions to the subjects could have varied slightly.

Alexander et al. (1991) investigated sit to stand in the young, and elderly with different functional abilities. The elderly subjects were broken into two groups based on ability to stand without the use of the arms. The elderly able group had an age range of 63-86 with a mean age of 72.1 years. The unable group had an age range of 75-92 with a mean age of 84.4 years. Subjects were asked to stand with and without using their arms or the armrests. Trials were videotaped for analysis. Force on the armrests was measured in the arm use trials. The movement was broken into two phases. The first phase started with the back moving off of a support and ended with the head reaching the most anterior position. The second phase ended with fully upright posture.

There were subtle differences between the old able and the young in trials without arm use. The older adults took longer to complete the task. Most of this increase in time was in phase one of the movement. There was more rotation of the trunk, thigh and shank during phase one in the elderly subjects.

There was very little difference between the old able and the young groups in the trials using the arms, however there were large differences for the old unable group. In the old unable group time to complete the task was twice as long as the old able. There was significantly more trunk flexion in phase one in the old unable group. There was also increased hand force relative to body weight.
Because the elderly subjects were broken into groups by functional ability and not by age, it is difficult to draw any conclusions about the effects of age between the two elderly groups. It is not surprising that the unable old group showed large differences compared to the old able. If these subjects cannot stand up without using their arms, then even when using the arms to stand up, it is likely a very near maximal activity. It is an interesting result that in the no arms trials there were differences between the young and the old able. These subjects reported no difficulty standing from a chair yet they have changed their strategy presumably to make this task easier in some way.

Schultz et al. (1992) wrote a companion paper to Alexander et al. (1991). They performed several investigations of the kinetics of Alexander and colleagues’ data. They investigated how segment motions affected the whole body center of mass and the location of the center of pressure at liftoff from the seat. They also computed the strength requirements for standing at liftoff from the seat and compared this with strength values reported in the literature. They do all of their analysis by assuming quasistatic conditions at liftoff. Inertial effects are ignored and the problem is solved as a static problem at the instant of liftoff.

For the 50th percentile male the center of mass was located 33 cm behind the heels at the start of the sit to stand movement. It needs to be moved over the feet after seat off to prevent a backwards fall. They found that flexion of the trunk and of the shanks have the largest effect on moving the center of mass towards the feet. The mechanism used by subjects is probably not flexion of the shanks because the feet are in a fixed location.

The elderly subjects had their center of pressure more anterior at seat off than did the young subjects. In the trials without arm use the mean center of pressure location was 1.5 cm posterior to the ankles in the young subjects and 1.9 cm anterior to the ankles in the able old subjects. In the trials with arm use the center of pressure locations at liftoff were 4.4 cm, 7.1 cm, and 10.7 cm anterior to the ankles for the young, able old and unable old subjects respectively. It is possible size of subjects may have had an influence on results, but the authors do not discuss this.

The highest moment requirements were found at the knee. In the old able subjects the mean requirement was $112 \pm 32$ Nm. This is much lower than strength
values they report for most elderly subjects (approximate age range 60-80). They report only studies of the frail elderly have shown values lower than this. The authors argue that this indicated the changes seen in movements in the healthy elderly are not a result of decreased strength. They argue that the anterior shift of the center of mass is to prevent backward falls not to decrease strength requirements for the task.

The study makes two assumptions to simplify analysis that could contribute to underestimating the strength requirements. First, they assume that the highest strength requirements are at liftoff. It is possible that higher moments are required at a different time in the task. Secondly, they ignore the moments caused by the motion of the segments. The study does provide an estimate of the strength requirements needed to perform sit to stand. It is not surprising that the strength requirements are lower than the capabilities of healthy elderly, as Schultz and colleagues report this group had little difficulty in performing sit to stand. There is not sufficient evidence to conclude that the changes in the sit to stand movement in the elderly are a result of an effort to prevent backwards falls. Backward falls into the chair are less likely to result in injury than forward falls and therefore it seems unlikely that movement would be coordinated to prevent them. It may be possible that the anterior shift of the center of mass at liftoff is designed to simply reduce effort.

Mourey et al. (2000) compared elderly and young subjects performing sit to stand using the concept of dynamic equilibrium area. The dynamic equilibrium area is the theoretical area where a person is stable. It is determined by the area of the base of support and the velocity of the center of mass. If a person is in the rear of the dynamic equilibrium area they are at risk for falling backwards. Forward falls are caused by moving beyond the front limit of the dynamic equilibrium area. Subjects performed sit to stand and kinematic data were recorded. Location within the dynamic equilibrium area was determined at the instant of liftoff.

As seen in other studies, the time to complete the task was greater in the elderly, and the velocity of the center of mass at liftoff was smaller. The elderly subjects were on average farther backwards in the dynamic equilibrium area at liftoff than the young subjects. Several elderly subjects were near the limit that could cause a backwards fall. The location of the center of mass at liftoff was not presented.
It is unclear if these elderly subjects used the strategy seen in Alexander et al. (1991), where the center of mass was moved farther anteriorly before liftoff. Alexander and colleagues proposed that this was to prevent backwards falling, however the data from Mourey and colleagues suggest that subjects are trying to prevent forward falls. If this is the case, then any additional forward displacement of the center of mass at liftoff is possibly to reduce effort not to maintain balance.

Sit to stand movements in the healthy elderly are similar to what is seen in young people particularly when the time to complete the task is constrained as was the case in Ikeda et al. (1991). If the time is not constrained and hands are not used to assist the movement, movement times are slightly longer in the healthy elderly. There is also increased forward displacement of the center of mass at liftoff and decreased velocity. The decreased forward velocity is likely done to prevent forward falls. The anterior shift of the center of mass could be to prevent a backward fall or to decrease the effort needed to complete the task.

### 2.2.2. The effect of chair height on sit to stand performance

Arborelius et al. (1992) studied the effect of chair height and arm use on muscular activity during sit to stand. Young healthy subjects performed sit to stand from three different chair heights without the use of their arms. The chair heights were set at knee height, knee height plus 1/3 thigh length, and knee height plus 2/3 thigh length. A fourth condition allowed arm use from the lowest chair height. Standing was performed with the feet on a force plate. Motion analysis data and EMG data were recorded. Subjects were asked to give ratings of difficulty of each task based on the Borg scale.

Subjective data revealed that standing from the higher chair is easier than from low heights. This is supported by EMG and joint moment data. There was a significant decrease in EMG activity as chair height increased. There was also a 60 percent decrease in the maximum knee moment in stands from the high chair compared with those performed from the low chair. There was very little change in hip or ankle moments. The maximum joint moments when rising from the lowest
chair height while using the arms is similar to what was seen in stands from the high chair without arm use.

The knee extensors are loaded more than any other muscle during sit to stand. Suggesting that for ease of movement reductions in the knee moment are therefore very important. Arborelius and colleagues do not report any kinematic data. It would be interesting to see what kinematic changes accompanied the reduced effort with increased chair height.

Weiner et al. (1993) investigated the effect of chair height on sit to stand performance in the elderly. The mean age of the 22 subjects was 72 years. After measuring many different chair heights to determine a range that an elderly person may encounter, the authors decided to study sit to stand from chairs ranging from 17 to 22 inches (43.2 to 55.9 cm) in height. Kinematic measurements were recorded. Subjects were asked to give a rating of perceived difficulty after each task.

Two subjects were unable to rise from any chair heights. Nine were able to complete all of the tasks. Eleven subjects could complete some, but not all of the tasks. As chair height increased the subjective rating of difficulty decreased and the number of subjects that could complete the task increased. There was significantly more hip flexion in rises from the low chair heights.

Based on the findings in this paper the authors propose raising chair heights in the homes of elderly individuals and in places frequented by elderly people. While this would likely be helpful, it cannot be applied everywhere an elderly person might go. Perhaps the most interesting finding of the paper is that sit to stand tasks performed from high chairs are done with a more upright trunk, which is more characteristic of sit to stand in the young (Alexander et al., 1991).

Hughes et al. (1994) investigated how chair rise strategies change with changes in chair height. They asked a group of 22 elderly subjects to perform a series of sit to stand tasks from chair heights ranging from 43.2 to 55.9 cm. The mean age of the subjects was 72 years, but the range of ages was 64 to 105 years. Kinematic data were measured and center of mass kinematics were analyzed.

Three strategies were defined. The first was characterized by high velocities at liftoff; the second by large movement of the center of mass towards the base of support before liftoff; and the last was a hybrid of the two. More than half of the
subjects used the same strategy at all chair heights. There were no trends with increasing chair height suggesting a change in strategy; this is contradictory to Weiner et al. (1993).

The large age range of subjects in this study complicates the analysis. They do not report any age effects, but given the large range there are almost certainly effects of age. If a smaller age range was used variability in performance would likely be reduced and differences would be seen more easily as chair height was varied.

Shenckman et al. (1996) investigated changes in segment and joint angular velocities occurring with changes in seat height. Two groups of subjects participated in this study. The young group had a mean age of 28.9 years. A second group was made up of healthy elderly subjects; the mean age of this group was 67.3 years. Subjects performed sit to stand from seat heights varying from 65 to 115 percent of knee height. All trials were performed in pace with a metronome set at 52 beats per minute. Kinematic data were recorded.

As height of the seat decreased there were increases in angular velocity of the trunk, hip and knee. The authors mix segment and joint velocities. However there was also an increase in the angular excursion of the knee. If the knee velocity is normalized with respect to excursion, there was no effect for seat height. The authors believe the increase in trunk velocity was to provide more momentum to allow completion of the more difficult task. There were no effects of age seen in the study.

The constraint of performing the task in pace with a metronome seems a strange choice. As seat height decreases Hughes et al. (1994) have shown that time to complete the task increases in subjects using the stabilization strategy. It is not surprising that subjects increased the momentum of their trunk to perform the task, because there was probably not sufficient time to use a strategy that would move the center of mass farther forward prior to liftoff. The time constraint also could have potentially hidden any age effects.

Munro et al. (1998) studied sit to stand from different seat heights and while using an ejector mechanism. Twelve elderly women with rheumatoid arthritis participated in the study. The mean age of the subjects was 65.5 years. Standing maneuvers were performed from a 45 cm seat with and without an ejector mechanism.
and from a 54 cm seat. Trials were performed with the feet on a force plate. Subjects were allowed to use the arms to push off. Kinematic data were recorded.

There were no significant differences in time to stand between the conditions, however the time spent on the seat was longer from the high seat and when using the ejector mechanism. There was a significantly greater angular velocity of the trunk at liftoff when rising unassisted from the low seat.

These data support the findings of Shenckman et al. (1996). The increase in angular velocity at liftoff suggests that momentum was used to rise from the more difficult position. Because the time spent on the seat was less in stands from the low chair, it is unclear whether the center of mass is shifted forwards to aid standing as suggested by Weiner et al. (1993).

Lowering seat height increases the difficulty in performing a sit to stand maneuver. With the exception of Hughes et al. (1994), most authors agree that in elderly populations there are changes in the movement to adjust for this increase in difficulty. There is disagreement as to whether this adjustment is to increase the momentum of the trunk at takeoff or to move the center of mass farther forward over the base of support.

### 2.2.3. The effect of muscular strength on sit to stand performance

Hughes et al. (1996) investigated the role of knee strength as a limiting factor in the sit to stand task. A group of functionally impaired elderly and a young control group performed sit to stands from progressively lower seat heights. Subjects were not allowed to use their arms. The minimum height that the elderly could stand from was recorded. Kinematics and kinetics were measured during each trial and inverse dynamics were performed with both quasistatic and full inverse dynamics methods. Maximum isometric knee strength at 60 degrees of flexion was also measured.

The quasistatic and dynamic methods for inverse dynamics showed very small differences, therefore only the values from the quasistatic analysis are reported. The mean isometric moment produced by the elderly was significantly lower than that of the control group. Hughes and colleagues found these elderly subjects could produce 103 Nm of knee extension moment on average. The lowest chair height any of the
elderly subjects could stand from was 43 cm. All young subjects were able to stand from a 33 cm chair. There were no significant differences in the required knee moment to complete each task. The young subjects used on average 39 percent of available knee strength to rise from the lowest chair. On the lowest successful trial the elderly subjects used 97 percent of their available strength.

The values for knee extension strength found here are smaller than most of the values reported by Schultz et al. (1992) in a survey of the literature. In that paper Shultz and colleagues suggest that strength decreases do not explain kinematic changes in sit to stand in the healthy elderly. Hughes and colleagues do not report kinematic changes, however the elderly subjects may have been using more trunk flexion at liftoff. Although the difference was not shown to be significant the elderly had reduced required knee moments. This is likely a result of increased trunk flexion.

Shenkman et al. (1996) investigated the effect of balance and strength on sit to stand performance in functionally impaired elderly. Strength was defined by a composite index that took knee extension/flexion and ankle plantar/dorsiflexion maximum isometric moments into account. Balance was measured in several ways: a functional reach task, center of pressure path length during quiet standing, and area of center of pressure ellipse during quiet standing. Performance was measured by lowest chair height the subjects were able to complete a successful rise from, and time to rise. Several kinematic variables were also measured: maximum hip flexion velocity and center of mass velocity in the vertical and horizontal directions.

Significant correlations were found between strength and lowest chair height and maximum vertical velocity of the center of mass. Functional reach showed correlations with lowest chair height, time to rise, and maximum hip flexion velocity, however there was a large correlation between strength and functional reach. Most of these correlations were small. The only correlation with an absolute r-value larger than 0.4 was between strength and lowest chair height (r-value = -0.639). There were no correlations between the static measures of balance and performance.

Strength appears to be an important determinant of sit to stand performance. The correlations with strength may have been larger if knee extension strength were used rather than the composite strength index. Hughes et al. (1996) showed the knee extensors are loaded more than other muscles in sit to stand. While balance is needed
to successfully rise from a chair, static balance measures show no correlations with performance and the correlations shown with functional reach are probably largely a result of the strong relationship between functional reach and strength. It would have been interesting to examine the location of the center of mass at liftoff. Schultz et al. (1992) have shown the elderly stand with their center of mass moved more anteriorly at liftoff compared with young people. They hypothesized this was a result of loss of balance not strength, because the strength requirements for their task were less than what is measured in healthy elderly. This may not have been the case for the subjects in Shenckman et al. (1996) because they were functionally impaired.

Gross et al. (1998) examined the relationship between strength and sit to stand speed in healthy elderly and young women. One repetition maximums were obtained for hip flexion and extension, knee flexion and extension, and leg press. Subjects were asked to rise from a 18 cm chair without using their arms. They were asked to do this under two conditions: at normal speed, and as fast as possible.

The young subjects were significantly stronger than the elderly subjects even when strength was normalized to body weight. The hip extensors were the muscle group with the largest strength deficit (57.5 percent). The knee extensors showed a 40.3 percent decrease. Young subjects performed the movement significantly faster than elderly subjects when asked to go as fast as possible. There was not a significant difference at the normal speed, although there was a trend that suggests the young subjects stood faster. Only one correlation was found between any of the strength measures and movement times. There was a small, but significant, \( r = 0.435 \), correlation between movement time and normalized hip extensor strength in the elderly subjects during the fast trials.

Two results from this study suggest that hip extensor strength is an important limiting factor in sit to stand performance in the elderly. First this is the muscle group where the largest strength loss was seen. Secondly this was the only strength measure that was correlated with movement time.

Corrigan and Bohannon (2001) investigated the relationship between knee extension strength and sit to stand performance in elderly women. The mean age of the subjects was 75.3 years. They measured sit to stand time from a 16 inch (40.6 cm)
bench. They also asked for ratings of perceived exertion. Knee extension force was measured with a hand held dynamometer.

Moderate correlations were found between strength and ratings of perceived exertion. These correlations were larger when strength was normalized with respect to body weight. There were also moderate correlations between strength and time to rise. Correlations were larger when a nonlinear fit was applied to knee extension force plotted against time to rise. The correlations found were similar to Shenkman et al. (1996). Plots of knee extension force against time to rise show this relationship is clearly non-linear. It appears their data may be more appropriately fit by two separate lines. The intersection of these lines could indicate a strength threshold for normal sit to stand performance.

Strength has been shown to be an important determinant of sit to stand ability. Hughes et al. (1996) showed that elderly subjects used 97 percent of their available knee strength to rise from the lowest chair height they were able to stand from. Shenkman et al. (1996) and Corrigan and Bohannon (2001) showed strong correlations between strength, particularly knee extensor strength, and sit to stand performance, while Gross et al. (1998) showed that hip extensor strength may be a limiting factor in sit to stand performance in elderly subjects.

2.3. Effects of aging on skeletal muscle

There are many changes in skeletal muscle that accompany aging. This section will review these changes as well as the effects of aging on tendon and the effect of strength training on aging muscle.

2.3.1. Decreased force output

The most obvious change in skeletal muscle accompanying aging is decreased ability to generate force. Loss of muscular strength is responsible for limiting movement in many elderly people (Hughes et al., 1996). This loss of strength could
be the result of loss of muscle mass, a reduction in specific tension of the muscle, or both of these factors.

Young et al. (1985) examined the size and strength of the quadriceps muscles of young and old men. The elderly subjects had a mean age of 75 years. They were of similar body mass and had similar lower leg lengths to the young group. Isometric strength was measured with the knee at 90 degrees of flexion. Ultrasound recordings were taken from the mid-thigh to measure muscle cross sectional area. The elderly subjects produced on average 39 percent less force than the young subjects, however there was only a 25 percent reduction in muscle size. This would suggest a 19 percent decrease in the specific tension of this muscle group. Young and colleagues did not account for intramuscular fat or connective tissue in their cross sectional area calculations. It is also possible that there are other changes that would alter the force-length properties of the muscles, which could complicate the results, for example changes in tendon properties that could shift the force-length curve of the muscles.

Klitgaard et al. (1990) compared muscle strength, muscle size, and the fiber types of elderly subjects with different training backgrounds with young sedentary control subjects. There were four elderly subject groups: a sedentary group, a running-trained group, a swim-trained group, and a weight trained group. All of the subjects in the training groups had been training for at least 12 years. Elbow flexion and knee extension strength were both studied the results for which were similar, therefore only knee extension strength will be discussed. Maximum isometric strength and maximum isokinetic strength at a range of angular velocities were measured. Computerized tomography (CT) scans were taken of the mid-thigh, and a muscle biopsy was performed near the area scanned.

The running-trained and the swim-trained groups were very similar to the sedentary group. They produced 44 percent lower maximum isometric moments than the young controls. They also produced a 20-26 percent lower percentage of maximum moment at any given joint angular velocity than did the young controls. They had 24 percent lower cross sectional area, and a 20 percent lower specific tension. The mean area of the muscle fibers was 24 percent less than the control group. The type II fibers (fast twitch) were affected more than the type I fibers (slow
The strength trained elderly group was similar to the young control group in all measurements.

The running-trained, the swim-trained and the sedentary groups showed similar loses of strength, muscle cross sectional area, and specific tension to the subjects in Young et al. (1985). The muscle biopsy results suggest that the loss of muscle mass is due to decreased fiber size, not loss of fibers. The loss of fiber size being larger with type II fibers could explain the changes in the force-velocity properties, as type II fibers are producing most of the muscle force at high velocities. Strength training appears to be highly effective at stopping the changes in skeletal muscle that occur with aging.

Frontera et al. (1991) compared the muscle strength and muscle mass of elderly subjects and middle-aged subjects. Maximum isokinetic knee extension was measured at 60 and 240 degrees per second. Muscle mass was estimated using urinary creatinine excretion. The elderly subjects produced about 20 percent less force than the middle-aged subjects. There was a 15 percent decrease when expressed relative to muscle mass. The strength decrement was slightly larger when measured at 240 degrees per second. These strength losses are smaller than when compared with a young group (Young et al, 1985) suggesting strength loss begins at an early age.

The loss of strength occurring with aging is a function of both loss of muscle mass, and a decrease in specific tension. The work of Klitgaard et al. (1990) shows that the muscle mass loss is probably a result of a decrease in fiber size rather than a loss of fibers, however this is contradicted by work of Lexell et al. (1988) reviewed in section 2.3.3. Klitgaard and colleagues also demonstrated that strength training can prevent these changes. The work of Frontera et al. (1991) shows that the loss of muscle strength begins before middle-age.

### 2.3.2. Increased twitch times

A more subtle change that occurs in aging muscle is an increase in twitch time. Davies and White (1983) electrically stimulated the triceps surae of elderly subjects and measured twitch times. Results are compared with young subjects from Davies
and White (1982). They found that elderly subjects could tolerate maximal twitch stimulation as well as maximal tetanic stimulation at 10 to 20 Hz and that the results were repeatable. When comparing their results with what has been measured in young people, the elderly subjects took 31 percent longer to reach maximal tension and had a 27 percent longer half relaxation time. Nothing in this study explains these results, however the authors argue that the loss of type II fibers may be important.

Vandervoort and McComas (1986) measured maximal isometric strength, muscle cross sectional area and twitch times in the plantar and dorsiflexors of subjects ranging from 20 to 100 years of age. Their results were similar to Young et al. (1985) for losses of muscle size and strength in subjects in their seventies, however these decreases were observed in subjects after their 8th decade. There was a 50 percent loss of strength in the subjects between 80 and 100 when compared with the young subjects. The results for contraction time and half relaxation time were similar to those reported by Davies and White (1983).

Keh-Evens et al. (1992) measured maximum twitch tensions in the lateral gastrocnemius of aerobically trained elderly subjects. Subjects ran at least three times per week for 30 minutes and had been training for at least a year. Muscle biopsy samples were also taken. They found there was no correlation between time to peak tension and percentage area of type II fibers, similarly there was no correlation for half relaxation time.

Hunter et al. (1999) investigated whether slowed calcium uptake into the sarcoplasmic reticulum is responsible for the increases in relaxation time in aging muscle. Four groups were examined: a young control, an elderly control, a young trained group and an elderly trained group. The training groups participated in a 12-week high intensity strength training program. Electrically evoked relaxation time was measured and the rate of calcium uptake was determined from muscle biopsies.

The elderly control had a longer relaxation time and slower calcium uptake than the young controls. Following training there were no changes in either of these variables for the young subjects. The elderly training group showed an increased calcium uptake rate, however there were no changes in relaxation time. It is therefore unlikely that calcium uptake is the limiting factor affecting relaxation time in aging muscle.
It has been clearly shown that there is both an increase in the time to reach peak tension and relaxation time in aging muscle. The mechanisms however are not clearly understood. Keh-Evens et al. (1992) showed no correlation between time to peak twitch tension and percentage area of type II fibers. Hunter et al. (1999) demonstrated that decreased rate of calcium uptake is unlikely the cause of the increase in relaxation time.

2.3.3. Decreased maximum shortening velocity

Another change that occurs in aging muscle is a decreased maximum shortening velocity. The loss could be due to a loss of, or a decrease in size of, type II fibers. Klitgaard et al. (1990), reviewed in section 2.3.1, showed elderly subjects produced a 20-26 percent lower percentage of maximum moment at any given joint angular velocity than did young controls and that there is a decrease in cross sectional area of muscle fibers, with type II fibers being effected more than type I. It is also possible that muscle fiber shortening velocity decreases with age (Larsson et al., 1997).

Larsson et al. (1979) investigated muscles strength and speed of movement in subjects ranging in age from 10 to 70 years. Maximum isometric knee extension was measured at 30, 60 and 90 degrees of flexion. Maximum isokinetic strength was measured at 30, 60, 90 and 120 degrees per second. Muscle biopsies were taken in subjects between 20 and 65 years old. Maximum knee extension velocity was measured in all but the oldest group.

Maximum isometric strength increased until the mid twenties then remained fairly constant until the forties. It decreases from the forties onward reaching a 38 percent decrease by the seventies. Maximum knee extension velocity showed a similar pattern however, the decrease across their age range was only seven percent. The decrease in isokinetic strength was similar to the decrease in isometric strength. Muscle biopsy results showed a decrease in size and number of type II fibers with aging.

Lexell et al. (1983) measured the size, number and proportion of type I fibers in vastus lateralis from six young and six old cadavers. The specimens were taken from previously healthy males who died of sudden accidents. The number of muscle fibers
was determined by examining the muscle as a group of 1 mm² sections. The fibers in every 48th section were counted and the average number was multiplied by the cross sectional area. The number of fibers per mm² was taken as an indirect measure of fiber size.

There was a 13 percent decrease in muscle cross sectional area in the old cadavers compared with young cadavers. There was a 24 percent decrease in the number of fibers. Both of these differences were significant, however there was not a significant change in fiber size. There was also not a change in the percentage of type I fibers.

Lexell et al. (1988) performed a similar study with 43 cadavers from a range of different ages, however in this study fiber size was directly measured in 125 fibers of each type. There was a 40 percent decrease in muscles cross sectional area in the oldest cadavers. There was a 39 percent decrease in the number of fibers. There was no change in the percentage of type I fibers. There was no change in the size of type I fibers, however there was a decrease in the size of type II fibers.

Larsson et al. (1997) investigated the contractile properties of isolated muscle fibers from young (n=4), sedentary elderly (n=2), and active elderly subjects (n=2). Maximum unloaded contraction velocity, maximum isometric tension and muscle fiber type were measured from samples taken by muscle biopsy. When comparing fibers of the same type, maximum unloaded contraction velocity and maximum isometric tension were generally lower in elderly sedentary subjects compared with the young subjects. The active elderly subjects were in the middle. This pattern was not followed in type IIb fibers. Although the sample size of this study was small, it shows that in addition to decreased size there are likely changes in aging muscle fibers which will result in decreased maximum force and shortening velocity.

Data from Larsson et al. (1979) and Klitgaard et al. (1990) have shown that there is a decrease in maximum shortening velocity in aging muscle. The work of Lexell and colleagues (1983 and 1988) show that this is probably a result of loss in size of type II fibers rather than a change in fiber type proportions. Larsson et al. (1997) shows this could also be the result of a decrease in shortening velocity of both muscle fiber types.
2.3.4. Effects of aging on tendon

Hubbard and Soutas-Little (1984) performed mechanical testing on the palmaris longus and hallucis longus tendons from 14 cadavers ranging in age from 16 to 88 years. No differences were found in the tendon’s Youngs modulus and the effects on hysteresis and rate of relaxation in cyclic tests were very small. No measurements of tendon dimensions were reported.

Gillis et al. (1995) investigated the properties of the superficial flexor digital tendon in horses of ages ranging from 2 to 23 years. Tendon cross sectional area was determined in vivo using ultrasound measurements, and following euthanasia Youngs modulus was measured during mechanical testing. There was no correlation between tendon cross sectional area and age. There was found to be a significant increase in Youngs modulus as the horses reached maturity. There was a small decrease in tendon modulus between mature and aged horses, although this difference was not significant.

Valour and Pousson (2003) investigated the effect of aging on the series elastic component of the human elbow flexors in vivo. Subjects were instructed to produce a maximum isometric contraction and the resistance was suddenly released. The behavior immediately following the release is a result of the series elastic component only and its compliance can therefore be estimated. Thirteen young subjects, mean age 21.5 years, and 15 elderly subjects, mean age 67.4 years, participated in the study.

There was an over 30 percent decrease in strength with age. Because of this the authors chose to express compliance in relative terms. The young subjects were shown to have a significantly more compliant series elastic components. If the effect was measured in absolute terms, there would be a larger difference between the two groups. There is no way to separate out what effect is caused by changes in the tendon elasticity from potential changes in the muscle cross bridge elasticity.

There appears to be very little change in tendon dimensions as a result of normal aging. Measurements of in vitro tendon mechanical properties also have shown little change. Valour and Pousson (2003) demonstrated that in vivo there is an increase in series elastic stiffness accompanying aging. Even if there is no change in
tendon material properties there will be a change in tendon function because there is a decrease in the maximal forces applied to the tendons, therefore aged tendons experience a smaller strain under maximum isometric force.

2.3.5. Effects of strength training on aging muscle

There have been a number of studies that have examined the effects of strength training in the elderly. This review will concentrate on two classic papers in the area (Frontera et al., 1988; Fiatarone et al., 1990). There also have been several studies that have specifically addressed the effect of strength training on sit to stand performance; several of these are also reviewed here.

Frontera et al. (1988) examined the effects of strength training on healthy sedentary men between the ages of 60 and 72. Subjects performed leg flexion and extension exercises three times a week for 12 weeks. Three sets of eight repetitions of each exercise were performed at 80 percent of each subject’s one repetition maximum. One repetition maximum was tested every week to adjust the training load. At baseline, 6 weeks and 12 weeks isokinetic strength measurements were made at a range of velocities and CT scans and muscle biopsy samples were taken.

Very large gains were reported in the one repetition maximums. Over 100 percent increase was seen in the knee extensors and over 200 percent increase in the knee flexors. The increase was about 10 times smaller in the isokinetic measurements with the largest gains for the low velocity test conditions. Muscle cross sectional areas increased by about 10 percent, with similar gains in size of the type I and type II fibers.

The strength improvements seen in this study were large. Although there was significant muscle hypertrophy, the performance improvement was also very specific. Additional study is needed to determine if these strength improvements can have functional value.

Fiatarone et al. (1990) examined the effects of strength training on the frail elderly. A group of nine subjects with a mean age of 90 years completed an eight week strength training program under close medical supervision. Subjects trained three times a week performing leg extensions. Each training session consisted of
three sets of eight repetitions. During the first week repetitions were performed at 50 percent of maximum. After the first week repetitions were performed at 80 percent of maximum. Before and after the study subjects were given functional tests for their ability to walk and rise from a chair. Seven of the subjects had CT scans before and after the training period.

Strength gains of 180 percent were seen as a consequence of the training. The subjects who had the CT scans showed an increase in muscle cross sectional area of over 10 percent. There was a 48 percent improvement in tandem gait speed. Two subjects no longer needed canes to walk, and one subject who had been unable to stand from a chair without the use of arms was able to do so. After the study all subjects resumed a sedentary lifestyle. Large losses in strength (32 percent on average) were seen after only four weeks of detraining.

Fiatarone and colleagues have shown that significant strength gains are possible in the very old. They have also demonstrated that these strength gains can have benefits for activities of daily living, however the training must be ongoing, because detraining effects are large. The type of training that would cause the largest functional improvement has not been determined.

Alexander et al. (2001) investigated how strength training affects sit to stand performance under a variety of conditions. A training group completed a 12 week strength training program consisting of hip abduction, knee extension/flexion, stair climbing, squatting, ankle dorsi/plantarflexion, and weighted chair rise. A control group completed a 12 week seated flexibility training program. The mean age of the training group was 82 years, while in the control group it was 84 years. Before and after the study subjects were tested in seven sit to stand tests. One test was sit to stand conducted as fast as possible. A second test had subjects perform sit to stand with their feet on a narrow beam. The other tests manipulated seat height and arm use. Seat heights used were 140 percent, 100 percent, and 60 percent of knee height.

There were no significant differences between the groups at baseline. When using the hands most subjects were able to rise from all three chair heights tested. Seventy-three percent were able to rise from the lowest height when using the arms. Only 57 percent of the subjects could stand from the middle chair height without arm use and only 27 percent could complete the task with the feet on the narrow support.
There were no data reported on strength increases following training. A few subjects became able to complete tasks that they were previously unable to perform. There was no change in rise time following training in all of the tasks except the hands use middle height condition. The training subjects generally had a more posterior location of the center of mass at liftoff than did the control group. There was an increase in velocity at liftoff in the training group as well.

Overall the effects of training were small. Only a few subjects became able to complete tasks that they were previously unable to perform. The training group had changes in the center of mass location and velocity at liftoff that more closely resembled young people, but the effects were small. Because there is no detail given about the intensity of training or strength gains that were seen, it is difficult to determine why the effect was not larger.

Schlicht et al. (2001) investigated the effect of strength training on a number of functional outcomes including sit to stand performance in healthy older adults. A training group of 11 subjects was compared to a control group. The mean age of the subjects was 72 years. Subjects were trained three days a week for 8 weeks. Six different lower body exercises were performed at each training session. During the first two weeks intensity was low to provide an acclimation period. Over the next six weeks subjects performed two sets of ten repetitions at approximately 75 percent of maximum. Subjects were tested pre- and post-training in maximal walking speed, time to complete five sit to stand trials, and one-legged blind balance time.

Increases in one repetition maximums ranged between 20 and 48 percent depending on the exercise, from week two until the end of the study. Maximal walking speed increased in both the training group and the controls. There was a significant increase in walking speed of the training group over the controls following the study. Sit to stand time decreased in both the training group and the controls; there was no significant difference between the two groups. There were no changes in balance time.

Increases in strength seen in this study were smaller than in other studies (Frontera et al, 1988). This is likely due to higher initial strength levels. These subjects were healthy and were given initial strength tests after two weeks of training. Training appeared to have an effect on maximal walking speed, but there was not an
effect on sit to stand time or one legged balance time. One possible reason for no
difference being found in the sit to stand task was the large improvement that came
with familiarization with this task. Results might have been different if time to
complete one rise was analyzed. It would be interesting to see if there were changes
in the kinematics of the sit to stand following training.

Sullivan et al. (2001) investigated the effects of strength training on elderly
subjects recovering from illness or surgery. The mean age of the subjects was 83
years. At study entry the subjects had very low physical performance. Subjects were
trained three times a week for 10 weeks performing leg presses. After two weeks to
get accustomed to the equipment, subjects performed three sets of eight repetitions at
80 percent of maximum. Subjects were tested in walking speed and sit to stand ability
pre- and post-training. It was determined that a subject’s performance on the test
improved if one of two criteria were met. The first was progressing from unable to
able; the second was improving speed of the test by at least 25 percent.

Subjects were able to complete the training safely and with very few minor
problems. Strength improved by 74 percent; 15 out of 19 subjects improved their
performance in sit to stand; and 10 subjects improved their maximum walking speed.
Strength improvement expressed relative to body weight showed a low correlation
with improvement in gait speed.

This study did not have a control group therefore all of the gains cannot be
attributed to strength training. Some of the improvement is certainly a result of the
conventional physical therapy they were receiving and natural recovery from their
illness or surgery. Schlicht et al. (2001) has shown that there can be a large
improvement after becoming accustomed to the testing, however this is unlikely to
allow subjects to progress from unable to able to perform certain tasks.

Recently Schot et al. (2003) investigated center of mass kinematics of sit to
stand performance following strength training. They focused on the velocity of the
center of mass and specifically at the transition from horizontal motion to vertical
motion. Subjects performed whole body strength training three times a week for eight
weeks. After familiarization with the exercises, subjects performed 7-10 repetitions at
80 percent of maximum. Kinematics of the sit to stand trials were recorded pre- and
post-intervention. Subjects were allowed to push off their knees with their hands. All subjects were healthy with a mean age of 73.2 years.

There were very large strength gains seen in the study. The score for the whole body improved an average of 161 percent. Values for the lower body were even larger, averaging 196 percent. There was a statistically significant increase in all velocities measured. There was a decrease in movement time after transition from horizontal motion to vertical motion, however there was not a significant difference in total movement time. The authors do not report center of mass positions. They believe that there is more anterior movement of the center of mass because the horizontal velocity is significantly increased and movement time is not.

Strength gains seen were very large for an eight week program with healthy elderly subjects. The conclusion that the center of mass moves more anteriorly is not justified. This would be an interesting finding as it would contradict Alexander et al. (2001), but it seems unlikely. Increasing velocity while simultaneously increasing forward movement of the center of mass would greatly increase the likelihood of forward falls.

Studies have shown that significant strength gains are possible even in the frail elderly. These strength gains are somewhat movement specific and there is a large detraining effect. The amount to which these strength gains can affect functional performance varies and is probably largely dependent on initial strength level. There is evidence though that strength training can improve sit to stand ability in those unable to perform difficult rising tasks, and can make sit to stand movements of the healthy elderly more like that of young people.

2.4. Direct dynamics simulations of jumping and sit to stand

Direct dynamics simulations have been used to study a wide variety of human movements. Running, walking, cycling, and many other movements have been studied. This section of the review will focus on studies of jumping and sit to stand.
2.4.1. **Direct dynamics simulations of jumping**

There are three groups of researchers who predominantly have conducted the majority of simulations of jumping. McNeil Alexander has performed several studies on jumping using simple models. Researchers at the Vrije University in Amsterdam have studied a number of sporting movements including jumping and often use simulation models in their work. These simulations are typically much more complex than those performed by Alexander. Maarten Bobbert has done most of this work. Marcus Pandy has done simulations of a number of human movements including several studies on jumping. Most of Pandy’s simulations are very complex and require large amounts of CPU (Central Processing Unit) time.

There are a number of advantages in performing simple simulations such as those performed by Alexander. Because of the relative simplicity there is far less chance of making errors and simulation time is reduced. Developing a simple model also forces the researcher to focus on the important elements of the system.

Alexander (1990) used a very simple model to examine why high jumpers have significantly lower approach velocities than long jumpers. The model had only one muscle, a knee extensor. The muscle was a moment generator with force-velocity properties. The optimal values for the angle of the take-off leg and run-up velocity were determined for both the long jump and high jump. For long jumps the model chose a high run-up velocity and for high jumps a slower run-up velocity was chosen. High run-up velocities in the high jump decreased the time the foot was in contact with the ground and therefore the impulse that was applied to the body.

Later Alexander (1995) used a slightly more complicated model to investigate jumping in humans, bushbabies, and insects. This muscle model had linear actuators with moment arms as opposed to moment generators in Alexander (1990), however only the knee joint had muscles. A number of aspects of human jumping were examined. The effect of the maximum shortening velocity of muscle, the compliance of tendon, number of leg segments, leg length, and mass distribution of the legs were all examined. Squat jumps were also compared with countermovement jumps.

Every parameter examined had an effect on jump height. Increasing the maximum velocity of shortening and increasing leg length produced the largest increases in performance. Countermovement jumps were higher than squat jumps.
when realistic values for tendon compliance were used, however jump heights were the same when an inelastic tendon was used in the simulation.

Bobbert and colleagues have used a more complicated model to examine a number of aspects of human jumping. These models are physiologically more realistic than those of Alexander, however the simulations consequently take much longer. The model used in all of their studies is very similar. The rigid body model consists of four links connected by frictionless hinge joints. The gluteal muscles, hamstrings, vasti, rectus femoris, soleus, and gastrocnemius were all modeled. Each muscle model consists of a contractile element, a series elastic element, and a parallel elastic element. The elastic elements are modeled with a non-linear stress strain relationships. The contractile element has force-length and force-velocity properties as well as activation dynamics. Muscle active state is related to muscle stimulation by a first order differential equation. Muscle stimulation patterns are optimized to produce maximum jump height, however the values for muscle stimulation are tightly restricted. The initial activation levels are determined such that the model is in static equilibrium. After that the muscle can be turned on only once and is then maximally stimulated throughout the movement.

One of the first papers using this model was van Soest et al. (1993). This paper examined the effect of the biarticular gastrocnemius on jump height. Jumps were simulated with a normal gastrocnemius as well as with a gastrocnemius only acting as a plantar flexor. The model with the biarticular gastrocnemius jumped 1 cm (2.6 percent) higher than with the uniarticular model. The biarticular model also had slightly higher efficiency.

This same model was used to examine why countermovement jumps are higher than squat jumps by Bobbert et al. (1996). A portion of this paper was reviewed in section 2.1.1 as it related to effect of initial squat depth on performance, however the main portion of the paper was not covered. The authors simulated jumps with and without a countermovement. Jumps with a countermovement were 3 cm (9.4 percent) higher than those without. They found that the muscles were at a higher active state at the bottom of the countermovement than in the initial position of a squat jump and that it took a considerable amount of time for the force produced by
the muscles in the squat jump to catch up with what is produced during a countermovement jump.

They also found that there was only 13 Joules of additional elastic energy stored in the tendons following a countermovement. This was not sufficient to explain the difference in performance. The authors argue that this energy does not contribute to enhanced performance, but only to enhancing efficiency. They demonstrate this with a simple one joint, one muscle simulation. They simulated three rotations with this model: one with a countermovement, one with a fast countermovement, and one without a countermovement. The simulation with the fast countermovement had more energy stored in the tendon, but showed a decrease in performance as compared with the slow countermovement condition. They state this proves that the stored energy can only contribute to increases in efficiency not improved performance. This would have been better shown by simulating a countermovement jump with an inelastic tendon.

In a recent study Bobbert examined how Achilles tendon compliance affects squat jump performance (Bobbert, 2001). Jumps were simulated with the strain under maximum isometric force ranging from 1 to 20% for the Achilles tendon. Jump height was 8 cm (22 percent) higher when simulated with a tendon that stretches 10 percent under maximum isometric force compared with a tendon that stretches 1 percent under maximum isometric force. The increased recoil of the tendon allowed the muscles to contract at lower velocities and produce higher forces. Jumps with the more compliant tendons had higher angular velocities of the shank and feet and lower angular velocities of the thighs and trunk when compared to simulations with the stiff tendons. This results in a lower percentage of the muscular work used for rotational energy. When tendon strains were increased beyond 10 percent, performance decreased as a result of the large shift in the muscles force-length properties. It would have been interesting to determine the effects of adjusting tendon compliance in other muscles in the model.

The results from this study seem to be contradictory to Bobbert et al. (1996). In that study Bobbert argues that increasing elastic energy storage by performing a countermovement does not enhance jump performance. Bobbert has argued (personal communication) that these studies are not contradictory. He stated that
increasing the elasticity of the tendons would improve performance by equal amounts in both squat and countermovement jumps.

The model used by Marcus Pandy is similar to the one used by Bobbert. He performs these simulations on a very powerful parallel processing computer, which allows him to run more complicated simulations. Pandy simulates two additional actuators, one representing the tibialis anterior and the other representing other plantar flexors. Optimization of muscle stimulation patterns is more complex as well. In Pandy (1990) muscle stimulation is allowed to be either 1 or zero, but unlike Bobbert's model stimulation can turn off and on again. This turns out to be of little importance as in the jump simulations once a muscle model is turned on the optimal solution has it remain on until the end, or very near the end for the hamstrings and soleus.

In a paper investigating the storage and use of elastic energy, Anderson and Pandy (1993) used more complex muscle stimulation patterns. The time history of muscle activation is represented as a series of control nodes with the activation between these nodes determined by linear interpolation. In this paper they found the elastic tissues contribute 35 percent of the total energy delivered to the skeleton during either a countermovement or a squat jump. They found that increasing tendon compliance improved jump height by 3 percent. In jumps with more compliant tendons a larger amount of the energy delivered to the skeleton came from the tendons in countermovement jumps as compared with the squat jumps.

Direct dynamics simulation has been shown to be a very powerful tool for studying jumping. Models of varying levels of complexity have all been useful. More complex models are more physiologically realistic, however take longer to run and have more parameters, which need to be determined.

2.4.2. Direct dynamics simulations of sit to stand

Very little work has been done simulating sit to stand movements. It is a very complicated movement, which has a much less well defined performance criteria than athletic movements. Because the time to complete the task is also much longer than
athletic movements, simulation times are much longer as well. This may be the primary cause for the lack of work in this area.

Pandy et al. (1995) developed a simulation of rising from a chair to determine a suitable performance criteria for sit to stand. The model used was very similar to Anderson and Pandy (1993). A chair was modeled as a spring damper at the hip. Five young healthy subjects performed sit to stand trials to compare with the model. The main performance criterion examined were minimizing the sum of muscle stresses squared, and minimizing the rate change of muscle force.

Neither performance criterion performed very well when comparing subject and model kinematics. Minimizing stress worked fairly well while the model was in contact with the chair, but resulted in high muscle forces and too quick a rise after seat off. Minimizing the rate change of muscle force worked well after seat off, but produced excessive muscle activation while seated. Pandy and colleagues examined a performance criteria that combined these two and found good agreement with EMG and force plate data from a group of experimental subjects. The combination of minimizing muscles stress while in contact with the chair and rate minimizing the rate change of muscle force after seat off seems reasonable. This combination produces movements similar to what subjects perform.

More recently Davoodi and Andrews (1999) developed a model to plan functional electrical stimulation assisted sit to stand. Muscles were modeled as moment generators with static and dynamic nonlinearities, time delays and fatigue effects. They used a genetic algorithm to adjust the gains for proportional integral derivative controller. The model was able to successfully reproduce desired trajectories.

Simulation of sit to stand movements is an undeveloped area. Pandy et al. (1995) demonstrated sit to stand can be simulated by minimizing muscle stress while in contact with the chair and minimizing the rate change of muscle force after seat off. Davoodi and Andrews (1999) have demonstrated that this can be an important area for clinical research. There has yet to be a study of sit to stand problems in the elderly utilizing a direct dynamics simulation.
2.5. Summary

The effects of initial squat depth and arm swing on maximum vertical jumping performance were reviewed. To date the literature has not thoroughly addressed the question of the effect of squat depth on jump height. The effect of arm swing on vertical jump performance has been fairly well quantified. The mechanism behind the improvement in takeoff velocity is unclear. It could be a result of increased momentum of the arms or arm swing could act to slow the extension of the legs allowing higher muscle forces to be generated.

The effect of aging, chair height, and strength on sit to stand movements were reviewed. There have been a number of reported changes in sit to stand movements associated with normal aging including: slightly longer movement times, increased forward displacement of the center of mass at liftoff, and decreased velocity of the center of mass at liftoff in the elderly. Lower chair heights have been shown to increase the difficulty of performing a sit to stand task. Changes in the movement to adjust for this increase in difficulty have been reported such as: increasing the momentum of the trunk at takeoff, or moving the center of mass farther forward over the base of support. Strength, particularly hip and knee extensor strength, has been shown to be an important determinant of sit to stand ability. It has been shown that elderly subjects may use nearly all of their available knee strength to rise from the lowest chair height they were able to stand from.

The effects of aging on skeletal muscle were reviewed. It has been shown that there is a large loss of strength. This is a function of both loss of muscle mass and a decrease in specific tension. It has also been shown that there is both an increase in the time to reach peak tension and relaxation time in aging muscle, however the mechanisms for this are not clearly understood. There is a decrease in maximum shortening velocity in aging muscle as well. This may be a result of loss size of type II fibers or a decrease in shortening velocity both muscle fiber types. There appears to be very little change in tendon dimensions as a result of normal aging. Measurements of in vitro tendon mechanical properties also have shown little change. In vivo measurements of series elastic stiffness show an increase accompanying aging.
Studies have shown that significant strength gains are possible even in the frail elderly. These strength gains are somewhat movement specific and there is a large detraining effect. The amount to which these strength gains can affect functional performance varies and is probably largely dependent on initial strength level.

The use of direct dynamics simulation in the study of jumping and sit to stand was reviewed. Direct dynamics simulation has been shown to be a very powerful tool for studying jumping. Models of varying levels of complexity have all been useful. Simulation of sit to stand movements is an undeveloped area. There has yet to be a study of sit to stand problems in the elderly utilizing a direct dynamics simulation.
2.6. References


Chapter 3

The influence of squat depth on maximal vertical jump performance.

3.1. Abstract
3.2. Introduction
3.3. Methods
   3.3.1. Experimental analysis
   3.3.2. Model analysis
3.4. Results
   3.4.2. Experimental analysis
   3.4.3. Model analysis
3.5. Discussion
3.6. References
3.1. Abstract

Increasing the distance a projectile will travel can be accomplished by increasing the time of force application. In jumping this can be accomplished by squatting as deeply as possible, however athletes do not do this when performing a jump. The effect of squat depth on jump height has not been thoroughly examined. Ten subjects performed jumps from preferred and deep squat positions. A computer model was also developed to simulate jumps from different starting postures. Experimental subjects showed no difference in jump height in jumps from the deep and preferred positions. Simulated jumps produced similar kinematics to experimental subjects. The optimal squat depth for the simulated jumps was the lowest position the model was able to jump from. Because jumping from a deep squat is rarely practiced, it is unlikely that these jumps are optimally coordinated. Differences in experimental vertical ground reaction force patterns also suggest that jumps from a deep squat are not optimally coordinated. Athletes trained to jump from deep squats may be able to increase performance over jumps from preferred squat depths.
3.2. Introduction

Maximum vertical jumping has received considerable biomechanical attention (e.g. Alexander, 1990; Challis, 1998; van Soest et al., 1985), and much is known about optimal performance (Pandy and Zajac, 1991). Jump height is a function of the vertical impulse generated during the ground contact phase of the jump. If an athlete wants to jump higher, greater impulse can be generated by applying greater forces, applying forces for a longer period, or a combination of both approaches. One option for increasing the duration of force application is to jump from a deep squat position. There are no studies, which have explicitly examined the influence of initial squat depth on jump height.

Bobbert et al. (1996) asked subjects to perform squat jumps from a preferred position, a position matching the bottom of jumps performed with a countermovement, and from a deep squat position. They found no difference in jump height between these conditions, suggesting that increased squat depth does not increase jump height nor does it decrease jump height. They performed no further comparisons between these conditions. Selbie and Caldwell (1996) used a simple simulation model to investigate the effect of initial position on countermovement jump height. Initial position had only small effects on jump height, but these were countermovement jumps and the bottom most position in the jumps was very similar for all jumps irrespective of initial position. Van Soest et al. (1994) used a direct dynamics simulation to investigate how muscle model activation patterns influence jumps made from different initial positions. Muscle activation was optimized to find maximum jump heights from each initial posture examined. A global optimal activation pattern for jumps from different initial positions was also determined. This activation pattern was one that would maximize the sum of the jump heights from each position. Jumps using the global optimization pattern were close to the optimal jump height for each position tested. They suggest that jumping from a deeper initial squat might not require significant changes in muscle activation patterns, however when jumps were optimally activated from deep squats, jump height was higher than in the preferred position.
A number of studies have indicated that jump height may not be influenced by initial squat positions, but none of these studies systematically examined the influence of squat depth on jump height. It is the purpose of this study to examine the effect of squat depth on maximum vertical jump performance; this will be achieved using a combined modeling and experimental approach. It is hypothesized that jump height will increase with increasing depth of squat, due to the greater period of time available for force production.

3.3. Methods

In the following sub-sections details are provided of the experimental and then the modeling approaches used for analyzing the jumps.

3.3.1. Experimental analysis

Ten healthy male subjects (age: 23.9 ± 2.7 years; height: 183.3 ± 6.2 cm; body mass: 85.5 ± 17.4 kg) participated in the study. All subjects provided informed consent; all procedures were approved by the institutional review board. Subjects performed six maximum vertical jumps commencing from an initial squat position. Three were performed from the preferred starting position. The subjects were then asked to perform three jumps from a self-selected deeper squat. During all the jumps the subjects kept their hands on their hips, to eliminate arm motion. Subjects warmed-up prior to testing, and rested for at least one minute between jumps.

Kinematic data were obtained using a Pro-Reflex Motion Analysis System (Qualisys, Inc.), sampling at 240 Hz. Markers were placed on the following body landmarks: acromion process, great trochanter, lateral femoral condyle, lateral malleolus, and the base of the fifth metatarsal. Joint angles were defined so that all joint angles were zero in upright stance. Ground reaction force data were synchronously sampled with the motion analysis data using a force platform (N50601, Bertec Corporation, Worthington, Ohio) sampling at 1200 Hz.
Jump height was assumed to be the maximum vertical displacement of the center of mass once contact was lost with the ground. It was computed from the center of mass vertical take-off velocity, which was determined from the impulse obtained by integrating the vertical ground reaction force-time curve with respect to time. Jump time is measured as the time from movement initiation to takeoff.

Resultant joint moments in the sagittal plane were computed for the ankle, knee, and hip joints (Winter, 1990). To determine these moments the segmental inertial parameters were determined for each subject by modeling their segments as series of geometric solids (Yeadon, 1990). The densities of these segments were derived from the cadaver data of Clauser et al. (1969). Moments were defined so that those causing joint extension were positive. The moments for each subject were normalized with respect to the product of the subjects’ body weight and height.

From the time histories of the joint angles, angular velocities, and resultant moments the maximum joint velocities and the maximum and minimum joint moments were measured as well as the timing of these events expressed as time before takeoff. To evaluate the statistical differences a repeat measures analysis of variance (ANOVA) was used. ANOVA was used to compare the jump height, jump time, maximum and minimum joint angles, angular velocities and moments under both conditions. For all statistical comparisons a significance level of 0.05 was used. Homogeneity of variance was confirmed for the data prior to performing the ANOVA using a Bartlett test. Statistical power was assessed to determine if the sample size was sufficient.

### 3.3.2. Model analysis

A direct dynamics simulation model was used to simulate jumps from different starting postures. The jumps were simulated using an optimal control direct dynamics muscle moment driven model. The model had four rigid links (foot, shank, thigh, and a combined head, arms, and trunk), connected by frictionless hinge joints. The foot was connected to the ground by a hinge joint at the metatarsal-phalangeal joint. It had a rotational spring-damper at this joint to represent the floor-heal interaction (Selbie & Caldwell, 1996). The equations of motion were formulated as
mixed differential-algebraic equations (Haug, 1989). The inertial parameters for the models links were the same as those for a typical experimental subject. The equations of motion for these links were integrated forwards in time using the improved Euler method (Ross, 1989), with a 0.0001s time step. Using a fifth order Runge-Kutta to integrate the equations did improve simulation accuracy, and slowed the simulations. (See appendix B for more details on the rigid body model.)

The model was actuated by six muscle models representing the: gastrocnemius, soleus, vastus group, rectus femoris, hamstrings group, and gluteal group. Each muscle was represented by a Hill-type model consisting of a series elastic element and a contractile element (Gallucci & Challis, 2002). The contractile element had force-length, and force-velocity properties, with output scaled via an activation dynamics model. Muscle active state was related to neural excitation by an ordinary differential equation (Pandy, 1992). Length and moment arm for each muscle was determined as a function of joint angle using the polynomial equations presented by Visser et al. (1990). Initial estimates of muscle model parameters for the lower limb were based on van Soest et al. (1993) and Friederich and Brand (1990); they were then adjusted to ensure each muscle produced force over the entire range of motion. Table 1 contains a summary of the model parameters used in this study. (See appendix A for more details on the muscle model.)

A genetic search algorithm (Goldberg, 1989) was used to select sequences of muscle model neural excitations for each of the muscles so that the total potential and kinetic energy was maximized at the instant the foot lost contact with the ground. The optimal control problem was converted into a static optimization problem (Goh & Teo, 1988) by representing the time histories of neural excitation for each of the six muscles as a series of control nodes separated by 0.05s. Neural excitation between these nodes was determined by linear interpolation. Neural excitation was constrained to be between 0 and 1. The initial values for neural excitation were selected so that the model would be in static equilibrium in the initial squat position. This was done by computing the resultant moments at the ankle, knee, and hip joints for this squat position. The muscle forces required to produce these moments were computed in the following fashion. First the contributions of the biarticular muscles to each joints resultant moment were allocated based on the fraction each of these
muscles could contribute to a maximum isometric moment at the joints they crossed. The contributions of the mono-articular muscles were then computed so that the initial squat resultant joint moments were produced. (See appendix D for more details on the optimal controller.)

<table>
<thead>
<tr>
<th>Table 3.1. A summary of the muscle model parameters.</th>
</tr>
</thead>
<tbody>
<tr>
<td>F&lt;sub&gt;max&lt;/sub&gt; (N)</td>
</tr>
<tr>
<td>Soleus</td>
</tr>
<tr>
<td>Gastrocnemius</td>
</tr>
<tr>
<td>Vasti group</td>
</tr>
<tr>
<td>Rectus Femoris</td>
</tr>
<tr>
<td>Hamstrings</td>
</tr>
<tr>
<td>Gluteal group</td>
</tr>
</tbody>
</table>

Note - F<sub>max</sub> – maximum isometric force (values are for both legs), L<sub>f_opt</sub> – optimal fiber length, W – spread of the force-length curve, L<sub>t</sub> – resting length of tendon, C – tendon strain under maximum isometric force. V<sub>max</sub> – maximum unloaded shortening velocity. K – force-velocity curvature constant.

Jumps were simulated from a position that matched the preferred position of a typical subject. This position is referred to as the standard position. The position that would maximize jump performance was also determined. For this problem the genetic search algorithm found the optimal initial segment angles as well as the control nodes. Jump height and the time history of joint angular velocities were analyzed and compared with the subjects for validation of the model.

3.4. Results

As in the methods section the results are broken into sub-sections detailing the experimental and then the modeling approaches.
3.4.1. Experimental analysis

There was a statistical difference between initial positions of the center of mass in the two conditions. There was statistically increased flexion at the knee and hip as the depth of squat increased (table 2). The subjects were very consistent in the depth of squat for each position, there were no statistical differences in squat depth between the trials.

Table 3.2. Means and standard deviations of the initial squat positions in the jumps for each of the conditions.

<table>
<thead>
<tr>
<th></th>
<th>Deep Squat</th>
<th>Preferred</th>
</tr>
</thead>
<tbody>
<tr>
<td>Center of mass height*</td>
<td>0.56 ± 0.07 m</td>
<td>0.68 ± 0.07 m</td>
</tr>
<tr>
<td>Hip angle*</td>
<td>101.5 ± 17.2°</td>
<td>73.4 ± 15.5°</td>
</tr>
<tr>
<td>Knee angle*</td>
<td>93.8 ± 16.0°</td>
<td>74.7 ± 10.3°</td>
</tr>
<tr>
<td>Ankle angle</td>
<td>27.49 ± 7.16°</td>
<td>26.46 ± 8.02°</td>
</tr>
</tbody>
</table>

Note –* - indicates that statistically significant difference exists

Ground contact time increased with squat depth, however there was no statistical difference between jump heights at the preferred or deep postures (table 3).

Table 3.3. Mean and standard deviations of jump heights and jump times for each of the conditions.

<table>
<thead>
<tr>
<th></th>
<th>Deep Squat</th>
<th>Preferred</th>
</tr>
</thead>
<tbody>
<tr>
<td>Jump time*</td>
<td>0.47 ± 0.07 s</td>
<td>0.30 ± 0.07 s</td>
</tr>
<tr>
<td>Jump height</td>
<td>0.27 ± 0.06 m</td>
<td>0.27 ± 0.06 m</td>
</tr>
</tbody>
</table>

Note –* - indicates that statistically significant difference exists

There were no statistical differences in the maximum (table 4) joint moments between jump heights at the preferred or deep postures, however the timing of the maximum joint moments at the knee and ankle occurred earlier in the jumps from the deep squat.
Table 3.4. Normalized maximum joint moments and their times for each of the conditions.

<table>
<thead>
<tr>
<th></th>
<th>Deep Squat</th>
<th>Preferred</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip moment</td>
<td>3.19 ± 1.45</td>
<td>3.28 ± 1.47</td>
</tr>
<tr>
<td>Time of max hip moment*</td>
<td>0.31 ± 0.05 s</td>
<td>0.25 ± 0.07 s</td>
</tr>
<tr>
<td>Knee moment</td>
<td>3.54 ± 1.51</td>
<td>3.50 ± 1.6</td>
</tr>
<tr>
<td>Time of max knee moment*</td>
<td>0.31 ± 0.05 s</td>
<td>0.26 ± 0.04 s</td>
</tr>
<tr>
<td>Ankle moment</td>
<td>3.29 ± 1.54</td>
<td>3.32 ± 1.39</td>
</tr>
<tr>
<td>Time of max ankle moment*</td>
<td>0.30 ± 0.05 s</td>
<td>0.26 ± 0.05 s</td>
</tr>
</tbody>
</table>

*Note* – * indicates that statistically significant difference exists. Times are expressed as time before takeoff. Joint moments are normalized by (100/Body Weight x Height)

A two-peak vertical ground reaction force pattern was seen in eight of the ten subjects in jumps from the deep squat for each of the three trials. This was not seen in any of the jumps from the preferred starting posture. Figure 3.1 shows the vertical ground reaction force for jumps from both positions for a typical subject.

3.4.2. Model analysis

Simulated jumps were similar to those produced by the subjects. Simulated jumps showed a proximal to distal sequence of joint extensions. Figure 3.2 shows a stick figure of the model throughout the jump from the standard squat depth. The simulated jumps for the standard position were of a similar height as the experimental jumps (table 5). The optimal initial position was a much deeper squat than that obtained by any of the experimental subjects (90 degrees of hip flexion, 105 degrees of knee flexion and 40 degrees of dorsiflexion). If any of the model’s joints were flexed two degrees more than this position the model was not strong enough to jump.
Figure 3.1. Vertical ground reaction force for jumps from the deep position (top) and preferred position (bottom).

Table 3.5. Jump height and time for the simulated jumps.

<table>
<thead>
<tr>
<th></th>
<th>Optimal</th>
<th>Standard</th>
</tr>
</thead>
<tbody>
<tr>
<td>Jump Time</td>
<td>0.36 s</td>
<td>0.25 s</td>
</tr>
<tr>
<td>Jump Height</td>
<td>0.26 m</td>
<td>0.19 m</td>
</tr>
</tbody>
</table>

Neural excitation for the uniarticular muscles was similar for jumps for both squat depths, however different neural excitation patterns were required in the biarticular muscles to produce optimal jumps from the two different starting positions. Figure 3 shows the muscle activation patterns for the biarticular muscles during both jumps.
Figure 3.2. Stick figure comparing the model to an experimental subject throughout a jump at 0.05 s intervals. Top – subject. Bottom – model.
Figure 3.3. Neural excitation for the biarticular muscles during both simulated jumps

3.5. Discussion

The experimental subjects jumped the same height from their preferred depth of squat, and from a deeper position. A power analysis showed that over 100,000 subjects would be required to show a statistical difference given the effect size seen. The model, albeit a simplification, had similar properties to the human musculo-skeletal system and jumped from the standard depth with a similar coordination pattern to the subjects. The simulation model jumped higher from a deeper position, but did so using different muscle activation patterns in the biarticular muscles than what was adopted from the higher squat. Doorenbosch et al. (1997) showed that the biarticular muscles are used to control the direction of force application. This is likely the cause for this change in activation.
Jumping from a deep squat is not often practiced, so the subjects presumably were not as well coordinated in the deep squats as they were at their preferred depth. This resulted in a two-peaked vertical ground reaction force in jumps from the deep position. The model is optimally coordinated and used a different coordination pattern for the biarticular muscles in the two jumps. This resulted in the model having a one peak vertical ground reaction force for both jumps. These results indicate that the experimental subjects had the potential to jump higher from a deeper squat.

There are a number of reasons why the subjects may not have typically executed deeper squat jumps. Several subjects commented that jumps from deep squats required much more effort than squats from the preferred position. There is also likely an increased initial strain on the joints during jumps from deep squats (Shelburne and Pandy 2002). In addition to this, in many sports the increased time for the deeper squat jump may be a disadvantage; although for the model this increase in time was not as large as it was for the experimental subjects. Given lack of practice of jumping from a deep squat, it is unlikely that athletes optimally coordinated these jumps.

All models have to be parsimonious representations of the complex systems they seek to emulate, the model used here is similar in complexity to many models of jumping (e.g., Bobbert et al., 1996), and is more complex than others (e.g., Alexander, 1990; Selbie & Caldwell, 1996). Many models of jumping have specified muscle activation sequence by giving an initial activation level, and then specifying a time when the muscle is either turned on to its maximal activation level, or turned off to its minimum activation level. The present model represented the time histories of neural excitation for each muscle as seven control nodes, giving the opportunity for more subtle control of the muscles. In contrast Van Soest et al. (1994) used the simpler activation scheme. Jumping simulations with their model suggested that humans use a similar activation pattern when jumping from a variety of starting positions. The current set of results suggests that their result may have been an artifact of their simple activation scheme.

The model suggests that athletes trained to jump from a deep squat position would likely be able to jump higher than from a preferred position. While jumping from a deep squat is not practical for most sports, it may be applicable to some, for
example high jumping. It may also be advisable for athletes to train to improve flexibility so that the option is available to achieve greater squat depths.
3.6. References


CHAPTER 4
THE CONTRIBUTION OF ARM SWING TO MAXIMUM VERTICAL JUMPS

4.1 Abstract
4.2 Introduction
4.2 Methods
  4.2.1 Experimental analysis
  4.2.2 Model analysis
  4.2.3 Induced Accelerations Analysis
4.3 Results
  4.3.1 Experimental analysis
  4.3.2 Induced acceleration analysis
  4.3.3 Model analysis
4.4 Discussion
4.5 References
4.1. Abstract

This study examined how arm swing contributes to maximum vertical jump height. Twelve male subjects each performed three maximum vertical jumps with arm swing and three without. There was a statistically significant increase in jump height with arm swing. Three possible mechanisms could explain this performance enhancement: the momentum of the arms at takeoff; the moment produced at the shoulder opposing extension of the legs causing the lower extremity musculature to operate on a more advantageous region of the force-velocity curve; or this moment opposing the extension of the legs providing the musculature longer to develop force. The vertical velocity of the arms relative to the shoulder at takeoff was large enough to account for the performance enhancement. Induced acceleration analysis confirmed that the shoulder moment was slowing the extension of the hip. Jumps with arm swing took longer and had lower maximum hip extension velocities than jumps without. Vertical jumps were also simulated using an optimal control direct dynamics muscle moment driven model of jumping. Simulation kinematics and kinetics paralleled the experimental data. The work produced by the lower extremity muscles was slightly reduced in jumps with arm swing. The work produced by the shoulder extensors was more than sufficient to explain the increase in jump performance. Therefore while the muscles of the lower extremity have longer to develop force and are shortening at a lower velocity, neither of these mechanisms increased the total work produced. Performance enhancement due to arm swing was due to the work produced by the shoulder musculature.
4.2 Introduction

Maximum vertical jumps have proven to be a valuable tool for examining human movement. For example examination of these jumps has helped elucidate the role in human movement of elastic energy storage (e.g., Anderson & Pandy, 1993), and the influence of biarticular muscles (e.g., Van Soest et al., 1993). When studying vertical jumping, it is common to ask subjects to perform jumps without arm swing, typically by keeping their hands on their hips. This simplifies analysis and reduces the amount of marker loss when performing motion analysis. Outside of the laboratory jumps are more often performed with an arm swing, and arm swing does increase jump height (e.g., Shetty & Etnyre, 1989), but the mechanism via which performance is enhanced has not been identified.

One mechanism by which a subject can jump higher by using an arm swing is the movement of the arms raises the whole body center of mass at takeoff. For the purpose of the present analysis jump height will be defined as the maximum vertical displacement of the center of mass after takeoff. But even with this definition jump height is greater with an arm swing than without (e.g., Shetty & Etnyre, 1989; Feltner et al., 1999).

Lees et al. (2000) reported that in high jumping the momentum of the arms generates a positive vertical velocity of the center of mass. Such a mechanism could be at play in standing vertical jumps as well. Feltner et al. (1999) proposed a contrasting mechanism, that the moment produced at the shoulder due to an arm swing opposes extension of the legs. Decreased lower limb joint angular velocities would allow the pertinent muscles to operate on a region of their force-velocity curves which permits the generation of greater force. While Feltner and colleagues were able to experimentally show reduced angular velocities at the hip and knee joints during jumps with arm swing, they could provide no direct evidence that the moment produced at the shoulder joint was the cause.

The mechanism proposed by Feltner et al. (1999) could be augmented by another mechanism, that is performance enhancement is due to the moment opposing the extension of the legs permitting increased time to complete the jump. This would
allow for a longer period of muscular force development and therefore an increase in jump height. This would be a similar mechanism to that proposed by Bobbert et al. (1996) for the enhancement of jump performance due to a countermovement.

These earlier studies have suggested three mechanisms for how jump height can be increased by arm swing: production of additional momentum, slowing leg extension which may permit muscles to work on a more favorable region of the force-velocity curve, and provide increased time for the muscles to generate force. It is the purpose of this study to examine how each of these mechanisms contribute to the performance enhancement of a maximum vertical jump due to arm swing. This will be achieved using a combined modeling and experimental approach.

4.3. Methods

In the following sub-sections details are provided for analyzing the jumps. The first sub-section details the experimental analysis. The second sub-section provides details on the model used. The final sub-section describes a procedure for determining the effect of arm movement on other joints in the body using the model and experimental data.

4.3.1. Experimental analysis

Twelve healthy male subjects (age: 24 ± 2.45 years; height: 181 ± 0.075 cm; body mass: 75.1 ± 5.6 kg) participated in the study. All subjects provided informed consent; all procedures were approved by the institutional review board. After appropriate warm-up the subjects performed six maximum vertical jumps commencing from an initial squat position. Arm swing was permitted for the first three jumps, while for the second three the subjects were instructed to keep their hands on their hips. Subjects were instructed to keep their arms straight while performing the arm swing. If the subject performed a countermovement prior to jumping or flexed the arms while jumping the jump was repeated.
Kinematic data were collected using a Pro-Reflex Motion Analysis System, sampling at 240 Hz. Markers were placed on the following body landmarks: styloid process of the ulna, the lateral epicondyle of the humerus, acromion process, great trochanter, lateral femoral condyle, lateral maleolus, and the base of the fifth metatarsal. Joint angles are defined so that all joints are at zero degrees in upright stance with arms at the side. Ground reaction force data were synchronously sampled with the motion analysis data using a Bertec force platform (Bertec N50601).

Jump height was assumed to be the maximum vertical displacement of the center of mass once contact was lost with the ground. It was computed from the center of mass vertical take-off velocity derived from the impulse obtained by integrating the vertical ground reaction force-time curve with respect to time. Resultant joint moments in the sagittal plane were computed for the ankle, knee, hip, and shoulder joints (Winter, 1990). To determine these moments the segmental inertial parameters were determined for each subject by modeling their segments as series of geometric solids (Yeadon, 1990). The densities of these segments were derived from the cadaver data of Clauser et al. (1969).

The initial and final joint angles, the maximum joint angular velocities and the maximum joint moments and their timings were determined from the computed time histories of the joint angles, angular velocities, and resultant moments. Timings of events were expressed as time before takeoff, moments were normalized with respect to subject mass and height, and positive moments designated to cause joint extension. The relative velocity of the arm is calculated as the vertical velocity of the shoulder at takeoff subtracted from the vertical velocity of the center of mass of the arm at takeoff.

The concept of relative momentum has been used to examine the contribution of arm swing previously (Lees and Barton, 1996), but because energy not momentum is linearly proportional to jump height, relative kinetic energy is used here. The relative kinetic energy is calculated by multiplying the square of the relative velocity by half the mass of the arm. This energy has a positive effect on jump height as long as the relative velocity is positive. Therefore, in the event of a negative relative velocity this energy will be expressed as “negative energy”. For determining jump height using an energy analysis all that is important is the energy at takeoff as the
kinetic energy at takeoff is directly related to takeoff velocity. However, it is possible that energy added early in the movement could have an effect on what is going on at other joints and therefore influence jump height. These effects will be examined using an induced acceleration analysis described in section 4.3.3.

To evaluate the statistical differences a repeat measures analysis of variance (ANOVA) was used. ANOVA was used to compare the jump height, jump time, maximum and minimum joint angles, angular velocities and moments with and without arm swing. For all statistical comparisons a significance level of 0.05 was used. A Bartlett test was used to confirm homogeneity of variance prior to performing the ANOVA.

4.3.2. Model analysis

The jumps were simulated using an optimal control direct dynamics muscle moment driven model. The model used was the same as Chapter 3 with the addition of an arm segment and muscles crossing the shoulder joint for the arm swing simulations. The model was actuated by eight muscle models representing the major muscles of the lower limb as well as shoulder flexors and extensors. Each muscle was represented by a Hill-type model consisting of a series elastic element and a contractile element (Gallucci & Challis, 2002). In the model the fraction of maximum muscle force for a particular muscle model state was specified by the muscles active state, which is determined by the neural excitation of the model. The length and moment arm for each muscle was determined as a function of joint angle using the polynomial equations presented by Visser et al. (1990). Initial estimates of muscle model parameters for the lower limb were based on van Soest et al. (1993) and Friederich and Brand (1990). Details on muscle model parameters can be found in Chapter 3. The shoulder muscle model parameters were estimated so that the moment-angle curves matched experimental data (Kulig et al., 1984). Table 4.1 contains a summary of the model parameters used in this study.
Table 4.1. A summary of the muscle model parameters.

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>$F_{\text{max}}$ (N)</th>
<th>$L_{\text{f}_{\text{opt}}}$ (m)</th>
<th>W</th>
<th>$L_{t_{r}}$ (m)</th>
<th>C</th>
<th>$V_{\text{max}}$ ($L_{f_{\text{opt}}}/s$)</th>
<th>K</th>
</tr>
</thead>
<tbody>
<tr>
<td>Soleus</td>
<td>13500</td>
<td>0.076</td>
<td>0.56</td>
<td>0.226</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>4500</td>
<td>0.050</td>
<td>0.56</td>
<td>0.350</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Vasti group</td>
<td>15000</td>
<td>0.158</td>
<td>0.56</td>
<td>0.11</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>3000</td>
<td>0.104</td>
<td>0.56</td>
<td>0.371</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Hamstring group</td>
<td>6750</td>
<td>0.245</td>
<td>0.56</td>
<td>0.154</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Gluteal group</td>
<td>10500</td>
<td>0.171</td>
<td>0.56</td>
<td>0.128</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Shoulders flexors</td>
<td>9000</td>
<td>0.20</td>
<td>0.56</td>
<td>0.08</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
<tr>
<td>Shoulders extensors</td>
<td>9000</td>
<td>0.20</td>
<td>0.56</td>
<td>0.08</td>
<td>0.04</td>
<td>5.2</td>
<td>2.44</td>
</tr>
</tbody>
</table>

**Note** - $F_{\text{max}}$ – maximum isometric force (values are for both legs), $L_{\text{f}_{\text{opt}}}$ – optimal fiber length, W – parameter setting spread of the force-length curve, $L_{t_{r}}$ – resting length of tendon, C – tendon strain under maximum isometric force, $V_{\text{max}}$ – maximum unloaded shortening velocity, K – force-velocity curvature constant.

A genetic search algorithm (Goldberg, 1989) was used to select sequences of muscle model neural excitations for each of the muscles so that the total potential and kinetic energy was maximized at the instant the foot lost contact with the ground. The optimal control problem was converted into a static optimization problem (Goh & Teo, 1988) by representing the time histories of neural excitation for each of the six muscles as a series of control nodes separated by 0.05s. Muscle activation between these nodes was determined by linear interpolation. The initial values for neural excitation were selected so that the model would be in static equilibrium in the initial squat position. This was done by computing the resultant moments at the ankle, knee, and hip joints for this squat position. The muscle forces required to produce these moments were computed in the following fashion. First the contributions of the biarticular muscles to each joints resultant moment were allocated based on the fraction each of these muscles could contribute to a maximum isometric moment at the joints they crossed. The contributions of the mono-articular muscles were then computed so that the initial squat resultant joint moments were produced.
Jumps were simulated without arm swing from an initial position that matched a typical subject. Jumps with arm swing were simulated from the same position with arms starting hanging vertically. For the simulated jumps the work done by each of the muscles in model was computed. This model has been validated in previous work for jumps without arm swing (Chapter 3). The simulated jumps with arm swing were validated by comparing the kinematics of a jump with arm swing between the model and the experimental subjects.

4.3.3. **Induced Accelerations Analysis**

During a jump a moment produced at the shoulder joint will contribute to the joint angular accelerations at the hip, knee and ankle. These accelerations are referred to as induced accelerations (Kepple et al., 1997). To compute these accelerations the experimental data were used to drive the simulation model. The jump was divided into intervals of 0.0042 s, at each instant throughout the jump, the model was given current, experimentally determined, joint angles and angular velocities. The weight of the segments and the shoulder moment were input and all joint angular accelerations computed for each instant. This was repeated with only the weight of the segments input and the difference between the two simulations provided the shoulder muscle moment induced accelerations. These induced accelerations provide information about the influence of arm swing on the kinematics of the other joints in the body.

4.4. **Results**

The results are broken into sub-sections detailing the experimental, induced acceleration, and then the modeling analyses.
4.4.1. Experimental analysis

Jumps with arm swing were significantly higher than jumps with arms on the hips (table 4.2). Jumps with arm swing also took significantly longer than jumps without.

Table 4.2. Means (± standard deviations) of ground contact time and jump height for the jumps with and without arm swing.

<table>
<thead>
<tr>
<th></th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ground contact time*</td>
<td>0.35 ± 0.11 s</td>
<td>0.31 ± 0.07 s</td>
</tr>
<tr>
<td>Jump height*</td>
<td>0.32 ± 0.07 m</td>
<td>0.27 ± 0.06 m</td>
</tr>
</tbody>
</table>

Note – * indicates a statistically significant difference between jump conditions.

Jumps with arm swing showed significant changes in the initial and final knee angle compared to jumps without arm swing (tables 4.3 and 4.4). Initially there is more knee flexion in jumps without arm swing, and at takeoff the knee is more extended in jumps without arm swing. There was a trend for a more extended hip at takeoff in jumps without arm swing, however this was not significant.

Table 4.3. Means (± standard deviations) of the initial and final and joint angles for the jumps with and without arm swing.

<table>
<thead>
<tr>
<th></th>
<th>Initial Angles</th>
<th>Joint Angles</th>
<th>Final Joint Angles</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Arm Swing</td>
<td>No Arm Swing</td>
<td>Arm Swing</td>
</tr>
<tr>
<td>Ankle angle</td>
<td>-25.8 ± 8.25°</td>
<td>-26.2 ± 7.73°</td>
<td>29.9 ± 9.97°</td>
</tr>
<tr>
<td>Knee angle</td>
<td>*68.0 ± 11.5°</td>
<td>74.3 ± 9.74°</td>
<td>*3.02 ± 11.1°</td>
</tr>
<tr>
<td>Hip angle</td>
<td>76.9 ± 14.9°</td>
<td>75.2 ± 15.9°</td>
<td>13.9 ± 13.7°</td>
</tr>
<tr>
<td>Shoulder Angle</td>
<td>29.2 ± 12.0°</td>
<td>-</td>
<td>133.5 ± 33.2°</td>
</tr>
</tbody>
</table>

Note – * indicates a statistically significant difference between jump conditions.

Jumps with arm swing had significantly lower maximum hip extension angular velocities than jumps without arm swing (table 4.4). There were no differences in ankle or knee angular velocities.
Table 4.4. Means (± standard deviations) of the maximum joint angular velocities during the jumps with and without arm swing.

<table>
<thead>
<tr>
<th></th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle angular velocity</td>
<td>543 ± 84°/s</td>
<td>564 ± 46°/s</td>
</tr>
<tr>
<td>Knee angular velocity</td>
<td>627 ± 105°/s</td>
<td>619 ± 70°/s</td>
</tr>
<tr>
<td>Hip angular velocity *</td>
<td>394 ± 84°/s</td>
<td>483 ± 82°/s</td>
</tr>
</tbody>
</table>

Note – * indicates a statistically significant difference between jump conditions.

The maximum joint moments at the hip, knee and ankle were all significantly higher in jumps with arm swing compared with jumps without arm swing. The timings of the maximum knee and ankle moments were significantly earlier in jumps with arm swing (table 4.5).

Table 4.5. Means (± standard deviations) of the normalized maximum joint moments, and the times at which they occurred during the jumps with and without arm swing. Times are presented as time before takeoff.

<table>
<thead>
<tr>
<th></th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum ankle moment*</td>
<td>0.043 ± 0.018</td>
<td>0.034 ± 0.015</td>
</tr>
<tr>
<td>Time of maximum ankle moment*</td>
<td>0.33 ± 0.06 s</td>
<td>0.25 ± 0.07 s</td>
</tr>
<tr>
<td>Maximum knee moment*</td>
<td>0.044 ± 0.015</td>
<td>0.036 ± 0.036</td>
</tr>
<tr>
<td>Time of maximum knee moment*</td>
<td>0.33 ± 0.05 s</td>
<td>0.26 ± 0.05 s</td>
</tr>
<tr>
<td>Maximum hip moment*</td>
<td>0.048 ± 0.021</td>
<td>0.036 ± 0.015</td>
</tr>
<tr>
<td>Time of maximum hip moment</td>
<td>0.21 ± 0.18 s</td>
<td>0.24 ± 0.09 s</td>
</tr>
</tbody>
</table>

Note – * indicates a statistically significant difference between conditions.

The arms had a positive vertical velocity relative to the shoulder at takeoff. This motion contributes to the total vertical kinetic energy and therefore jump height. The vertical velocity of the arms relative to the body at takeoff was 2.68 ± 0.90 m/s. The mean kinetic energy produced by this was 41.9 ± 23.9 J for the subjects. This is sufficient to account for a 4.7 cm increase in jump height.
4.4.2. Induced acceleration analysis

The pattern of induced acceleration for the experimental data was similar for all of the subjects; the results for a typical subject are shown in figure 4.1. For the first half of the jump there was very little movement and therefore small amounts of induced acceleration. Approximately two thirds of the way through the jump the shoulder moment induces an acceleration slowing hip extension, but aiding knee extension and ankle plantar flexion. At the end of the jump the shoulder induced acceleration is aiding extension of the hip and knee, and plantar flexion of the ankle.

Figure 4.1. Acceleration induced at the ankle, knee, and hip joints by the moment associated with arm swing during a jump for a typical subject.
4.4.3. Model analysis

The model results paralleled the experimental data as the simulated jumps with arm swing produced higher jumps than simulated jumps without arm swing, and took longer to produce (table 4.6). The magnitude of this improvement was similar to that seen in the experimental subjects. The kinematics of the jumps were reproduced well by the model, figure 2 shows a stick figure of the model and a typical experimental subject jumping with arm swing.

Table 4.6. Jump height and ground contact time for the simulated jump with and without arm swing.

<table>
<thead>
<tr>
<th></th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ground Contact Time</td>
<td>0.33 s</td>
<td>0.27 s</td>
</tr>
<tr>
<td>Jump Height</td>
<td>0.28 m</td>
<td>0.20 m</td>
</tr>
</tbody>
</table>

Investigation of model output confirmed that the jumps with arm swing have reduced muscle shortening velocities for the lower extremity muscles. Table 4.7 presents the maximum shortening velocities for both simulated jumps. The maximum force produced by the muscles in the two simulations was similar for the ankle musculature and the gluteal group, but was higher in the knee extensors and hamstrings in jumps with arm swing (table 4.8).

Table 4.7. Maximum muscle shortening velocities for the simulated jumps with and without arm swing.

<table>
<thead>
<tr>
<th></th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gastrocnemius</td>
<td>0.25 m/s</td>
<td>0.47 m/s</td>
</tr>
<tr>
<td>Soleus</td>
<td>0.51 m/s</td>
<td>0.79 m/s</td>
</tr>
<tr>
<td>Vasti group</td>
<td>0.65 m/s</td>
<td>0.87 m/s</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>0.32 m/s</td>
<td>0.48 m/s</td>
</tr>
<tr>
<td>Hamstrings group</td>
<td>0.52 m/s</td>
<td>0.54 m/s</td>
</tr>
<tr>
<td>Gluteal group</td>
<td>0.96 m/s</td>
<td>1.08 m/s</td>
</tr>
</tbody>
</table>
Table 4.8. Maximum muscle force for the simulated jumps with and without arm swing.

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gastrocnemius</td>
<td>1621 N</td>
<td>1876 N</td>
</tr>
<tr>
<td>Soleus</td>
<td>5456 N</td>
<td>6045 N</td>
</tr>
<tr>
<td>Vasti group</td>
<td>10054 N</td>
<td>7027 N</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>2508 N</td>
<td>1814 N</td>
</tr>
<tr>
<td>Hamstrings group</td>
<td>2241 N</td>
<td>1111 N</td>
</tr>
<tr>
<td>Gluteal group</td>
<td>2527 N</td>
<td>2663 N</td>
</tr>
</tbody>
</table>

While there was increased maximum force in several muscles in jumps with arm swing, overall the muscles of the legs in the model did more work during the jumps without arm swing than during jumps with arm swing, 13 J less (table 4.9). This is because the muscles work over a smaller range of motion in these jumps. The muscles crossing the shoulder joint produced 199.9 Joules of work, which is more than enough to increase the jump height of the model by 8 cm.

Table 4.9. Work produced by the muscles for the simulated jumps with and without arm swing.

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>Arm Swing</th>
<th>No Arm Swing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gastrocnemius</td>
<td>91.8 J</td>
<td>77.5 J</td>
</tr>
<tr>
<td>Soleus</td>
<td>33.7 J</td>
<td>23.3 J</td>
</tr>
<tr>
<td>Vasti group</td>
<td>85.7 J</td>
<td>107.4 J</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>15.2 J</td>
<td>29.7 J</td>
</tr>
<tr>
<td>Hamstrings group</td>
<td>59.7 J</td>
<td>44.4 J</td>
</tr>
<tr>
<td>Gluteal group</td>
<td>49.5 J</td>
<td>66.3 J</td>
</tr>
<tr>
<td>Shoulder extensors</td>
<td>237.8 J</td>
<td>-</td>
</tr>
<tr>
<td>Shoulder flexors</td>
<td>-37.9 J</td>
<td>-</td>
</tr>
</tbody>
</table>
Figure 4.2. Stick figures of a typical experimental subject performing a jump with arm swing (lower sequence), and simulated jump (upper sequence). Note that the arm is represented to the elbow joint only to permit compression of the images.
4.5. Discussion

This study examined the change in jump height in maximum vertical jumps as a consequence of arm swing. The change in jump height for the experimental data and the model were comparable to that reported in other studies (e.g., Shetty & Etnyre, 1989; Feltner et al., 1999). The model was a parsimonious representation of a complex system but produced results comparable to the experimental data, and is similar in complexity to other models used to investigate jumping (e.g., Anderson & Pandy, 1993; Bobbert et al., 1996).

Results from the experimental analysis show all three proposed mechanisms for enhancement of performance in jumps performed with arm swing are feasible explanations. Indeed it could be feasible that all three mechanisms contribute to increased jump height. The jumps with arm swing took longer than jumps without. This may allow the muscles of the lower extremity more time to generate force. There was also a reduction in the maximum angular velocity of the hip for the jumps with arm swing. An induced acceleration analysis confirmed that the shoulder moment induces a flexion moment at the hip approximately two thirds of the way through the jump. The shoulder moment aided in extension of the knee and plantar flexion of the ankle at the end of the jump. Finally, the velocity of the arms relative to the shoulder was large enough to give an increase in kinetic energy that could account for the increase in jump performance.

The experimental work could not eliminate any of the proposed mechanisms. For example the velocity of the arms relative to the shoulder could produce enough kinetic energy to explain the increase in jump height, but this may not result from work done by the shoulder extensors, but rather work done by the lower extremity that is transferred to the shoulder joint. The model permitted a detailed analysis of the work generated by the muscles during the jumps. The lower extremity muscles actually produced less work in the jumps with arm swing than they did in the jumps without arm swing. The increased jump height for the jumps with arm swing can be attributed to the work by shoulder musculature. Therefore the work produced by the
shoulders accounts for the increase in performance in arm swing maximum vertical jumps, not the other mechanisms previously proposed.

The work contribution of the shoulders was much greater than that produced by the other modeled muscles. The advantage in producing work the shoulder has compared with the other joints in the model is that this joint can move through a greater range of motion during the jumps (table 3).

Model parameters were determined from sources in the literature and are reflective of parameter sets used by other researchers for simulating vertical jumps (e.g., Anderson and Pandy, 1993; van Soest et al., 1993). The exception is for the model of the shoulder musculature where modeling efforts are not as common and therefore model parameters sets less frequent. The model parameters for the shoulder musculature were selected to reflect moment-angle data, but were purely phenomenological in nature. The model shoulder peak angular velocities and moments differed from mean experimental shoulder peak angular velocities and moments by less than half the experimental standard deviation in each case, providing confidence in these model parameters.

This study examined the influence of arm swing on jump height, although of course raising the arms also raises the whole body center of mass prior to take off. Feltner et al. (1999) reported that the increase in whole body center of mass height at take off for an arm swing jump compared with a no arm swing jump was approximately 8 cm. A similar mechanism was reported for standing long jumps (Ashby & Heegaard, 2002).

Other tasks can be performed with an arm swing, for example sit to stand. It would be informative to examine the influence of arm swing on this task, particularly as this task can be so challenging to the elderly (e.g., Hughes et al., 1996). If arm swing aids sit to stand the mechanism may be somewhat different because the legs are supported at the start of the movement so this cannot be influenced by arm swing, and at the end of the movement minimum vertical velocity is the target not maximum vertical velocity. A further investigation using the protocols used in the present study to examine the influence of arm swing on sit-to-stand performance seems warranted.
4.6. References


CHAPTER 5
THE INFLUENCE OF SEAT HEIGHT ON SIT TO STAND MECHANICS: A SIMULATION STUDY

5.1 Abstract
5.2 Introduction
5.3 Methods
5.4 Results
5.5 Discussion
5.6 References
5.1 Abstract

Lowering seat height has been shown to increase the difficulty of rising from a chair, which is particularly problematic for the elderly, because rising from a chair for this population can be a near maximal strength task. Increasing anterior motion of the trunk prior to leaving the seat has been proposed as a mechanism to make this task easier, particularly from low seat heights. A simulation model was used to examine the effects of reducing seat height on sit to stand performance; the model allows balance effects to be separated from strength effects. The model used was a three link direct dynamics simulation model driven by muscle-like actuators. Sit to stand was simulated from a range of different seat heights. The model analysis suggests that employing greater trunk motion in sit to stand movements from low seat heights moves the hip extensors into an unfavorable region of their force-length curve. The rotation of the trunk forward in the model decreases as seat height lowers. Instead of changes in trunk movement decreasing the load on the knee extensors, the load on the knee extensors is actually increased by the reduced anterior trunk movement. This is particularly unfortunate as these muscles are already highly stressed as a result of increased initial knee angles accompanying reduced seat height. The simulation model analysis suggest that as seat height decreases typical subjects would not benefit from increasing anterior trunk motion before seat off during the sit to stand.
5.2. Introduction

As an increasing percentage of the America population reach old age, research into activities of daily living that challenge older adults has become an important focus. Standing from a seated position has received much of this attention; this is likely because it is a very challenging activity for older adults (e.g., Rodosky et al., 1989) and clearly needs to be maintained to allow independent living.

Lowering seat height has been shown to increase the difficulty of rising from the chair (Arborelius et al., 1992). The impact of this increasing difficulty is particularly important for elderly individuals, because rising from a chair can be a near maximal strength task (Hughes et al., 1996). There are contradictory reports as to what kinematic changes occur in the sit to stand as a consequence of lowering seat height in the elderly. Weiner et al. (1993) reported increased forward rotation of the trunk while in contact with the seat, as seat height decreased from 56 cm to 43 cm. Lower leg length was 40.4 ± 2.46 cm for the subjects in the study. Alexander et al. (1991) suggested that elderly individuals move the center of mass forward at lift off from the seat to prevent backward falls. It is also possible this is done to reduce the effort needed to stand, as the center of mass would be closer to the knees and therefore the muscular moments at this joint required to rise could be reduced.

Schenkman et al. (1996) found an increased angular velocity of the trunk while in contact with the seat, as seat height decreased from 115% to 65% of knee height. They suggested that the increased rotation of the trunk is a consequence of efforts to increase forward momentum, not a consequence of trying to move the center of mass farther forward. Such a strategy introduces the risk of forward falls, indeed Pai and Rogers (1990) have demonstrated that horizontal momentum is invariant irrespective of sit to stand speed, suggesting additional horizontal momentum can be problematic.

Hughes et al. (1994) found three different strategies were used when elderly subjects rose from a chair. One group of subjects moved their center of mass forward over the feet and would then rise. A second strategy was to generate high forward momentum while in contact with the seat and use this momentum to assist in rising.
The third strategy was a combination of the first two. However, they found that these strategies did not change based on seat height.

Experimental work with the elderly can be complicated by a number of health problems that can influence results. Rising from a chair is a balance perturbation and balance has been shown to be an important determinant of sit to stand success (Schenkman et al., 1996). Because falls can be so devastating for the elderly, fear of falling also complicates experimental analyses of the sit to stand in the elderly (Kressig et al., 2001). Pandy et al. (1995) presented and validated a computer simulation model of sit to stand performance. Such a model can represent the physical properties of a population (e.g., inertial and muscle properties), and permit the examination of their influence on coordination. Health complications can be taken out of consideration using a computer simulation model, for example balance problems, which allows focused examination of the influence of strength and coordination on performance.

Two strategies have been proposed for dealing with decreases in seat height, but experimental evidence provides conflicting evidence for the adoption of these strategies. Of course it is also feasible that both strategies are used simultaneously. The following study uses a computer simulation model with muscle properties adjusted to be reflective of the elderly to examine how kinematics are adjusted to deal with the increasing difficulty of standing from progressively lower seat heights. It is hypothesized that with decreasing seat height trunk motion while in contact with the seat will increase, to reduce the load on the knee musculature but not for momentum generation.

5.3. Methods

An optimal control direct dynamics muscle moment driven model was used to simulate rising from a chair. The model used was similar to the model previously used to examine jumping (Chapters 3 and 4). The foot was assumed to be stationary throughout the movement and was therefore considered part of the ground. The remaining three model segments (shank, thigh, and the head, arms, and trunk) were
modeled as rigid links, connected by frictionless hinge joints. Joint angles were
defined according to Figure 5.1, where $\alpha$ is the ankle angle, $\beta$ is the knee angle, and $\gamma$
is the hip angle. Trunk angle is defined as the angle between the trunk and the
vertical axis. Contact forces with the seat were modeled with an exponential spring
and a linear damper in both the horizontal and vertical directions using the model
proposed by Pandy et al. (1995). The equations of motion were formulated as mixed
differential-algebraic equations (Haug, 1989). The inertial parameters for the models
links were the same as those for a typical experimental subject from the study of
Burgess (2003) where the mean subject age was 71.8 years. The model was designed
to have a mass of 82 kg, and height of 1.65 m.

Figure 5.1. Joint angle definitions.

The model was actuated by eight muscle models representing the:
gastrocnemius, soleus, tibialis anterior, vastus group, rectus femoris, hamstrings,
gluteal group, and hip flexors. The muscle component of this simulation differs from
that used in Chapters 3 and 4 in several ways. First, the muscles were assumed to
have rigid series elastic elements. This allows the integration time step to be much
larger (0.001 s versus 0.0001 s). This reduction in simulation time is important when
simulating complex systems (e.g., Anderson et al., 1995).

To determine if using a rigid tendon to simulate sit to stand was appropriate, a
simple simulation of a single muscle moving a mass was used. Simulations were
performed with a rigid and elastic tendon. The muscle chosen was the vasti group, as
this is a primary mover in sit to stand. The muscle shortens through a range of motion similar to what is seen during sit to stand, while moving a 100 kg mass. The objective function used minimized the rate change of muscle force. The simulation was constrained such that the mass must be stationary at the end point. The two simulations were similar, with the muscle force never differed by more than two percent. See Appendix E for more details.

The original parameter values for the muscle model were those used in Chapter 3, and were based on the data presented by van Soest et al. (1993) and Friederich and Brand (1990). But no information was available for the hip flexors and the tibialis anterior. Therefore, the hip flexors were given the same parameter values as the gluteal group, and the tibialis anterior the same as the soleus. Muscle lengths and moment arms were determined using the equations of Visser et al. (1990), except for the tibialis anterior for which the data of Maganaris (2000) were used.

The original muscle model parameter set reflected those of a healthy young subject, some adjustments were made to these parameters to make them reflective of an older subject (approximate age 70 years). The models maximum isometric forces were reduced by 40 percent to reflect the strength loss, which accompanies aging (e.g., Klitgaard et al., 1990; Young et al., 1985). A loss in the number of fast-twitch muscle fibers was also modeled (Larsson et al., 1979). The force-velocity properties for the muscles modeled were determined by an optimization procedure. The values for maximal velocity of shortening and curvature of the force-velocity curve for both slow and fast twitch muscle fibers were taken from Faulkner et al. (1986). A force-velocity curve was then generated for each of the muscles using the following fiber distributions: 20 percent fast twitch and 80 percent slow twitch fibers for the soleus and 33 percent fast twitch and 67 percent slow twitch fibers for the remaining muscles. An optimization was then run to determine the whole muscle force-velocity parameters that would match this output. Table 5.1 contains a summary of the model parameters used in this study.
Table 5.1. A summary of the muscle model parameters.

<table>
<thead>
<tr>
<th></th>
<th>$F_{\text{max}}$ (N)</th>
<th>$L_{\text{fopt}}$ (m)</th>
<th>$W$ (m)</th>
<th>$L_{\text{r}}$ (m)</th>
<th>$V_{\text{max}}$ (L_{\text{fopt}}/s)</th>
<th>$K$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Soleus</td>
<td>8100</td>
<td>0.076</td>
<td>0.56</td>
<td>0.226</td>
<td>3.82</td>
<td>10.76</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>2700</td>
<td>0.050</td>
<td>0.56</td>
<td>0.350</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>8100</td>
<td>0.076</td>
<td>0.56</td>
<td>0.226</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Vasti group</td>
<td>9000</td>
<td>0.128</td>
<td>0.56</td>
<td>0.14</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Rectus femoris</td>
<td>1800</td>
<td>0.104</td>
<td>0.56</td>
<td>0.371</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Hamstrings group</td>
<td>4050</td>
<td>0.245</td>
<td>0.56</td>
<td>0.154</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Gluteal group</td>
<td>6300</td>
<td>0.171</td>
<td>0.56</td>
<td>0.128</td>
<td>5.26</td>
<td>12.45</td>
</tr>
<tr>
<td>Hip flexors</td>
<td>6300</td>
<td>0.171</td>
<td>0.56</td>
<td>0.128</td>
<td>5.26</td>
<td>12.45</td>
</tr>
</tbody>
</table>

Note - $F_{\text{max}}$ – maximum isometric force (values are for both legs), $L_{\text{fopt}}$ – optimal fiber length, $W$ – spread of the force-length curve, $L_{\text{r}}$ – resting length of tendon, $V_{\text{max}}$ – maximum unloaded shortening velocity, $K$ – force-velocity curvature constant.

In the simulations a criterion was adopted to determine when upright stance was achieved, the stand score (ss). Upright stance was defined as a stand score of less than 0.25, which was determined by Equation 5.1. The coefficients were selected to allow motion that is seen during quite stance in the elderly (Burgess, 2003). The stand score was used as an inequality constraint for the optimization, because using ground reaction force only would allow any final position in equilibrium.

$$ss = a \Delta p + b \cdot v$$  \hspace{1cm} (5.1)

Where:

- $\Delta p$ - the distance between the center of mass and its position in upright stance
- $v$ - absolute value of the magnitude of the center of mass velocity vector
- $a, b$ - constants with values of 10 m$^{-1}$ and 1 s.m$^{-1}$ respectively
The performance criterion for the sit to stand task was that presented by Pandy et al. (1995). Here the task is divided into two stages: one prior to seat off where the sum of the muscle stresses squared is minimized, and after seat off where the sum of the rates of change of the muscle forces is minimized. A genetic search algorithm (Goldberg, 1989) was used to select sequences of muscle model neural excitations for each of the muscles so that the following objective function was minimized,

\[
\begin{align*}
\mathbf{f} &= \frac{1}{c} \int_{0}^{t_1} \left( \frac{Fm_i}{F_{\text{max}_i}} \right)^2 dt + \frac{1}{d} \int_{t_i}^{t_{\text{max}}} \left( \frac{\dot{Fm}_i}{F_{\text{max}_i}} \right)^2 dt \\
\end{align*}
\]

(5.2)

Where:

- \( \mathbf{f} \) - function to be minimized
- \( t_1 \) - time at which seat off occurs
- \( Fm_i \) - force for a given muscle
- \( F_{\text{max}_i} \) - maximum isometric force for a given muscle
- \( \dot{Fm}_i \) - rate of change of muscle force for a given muscle with respect to time
- \( c, d \) - constants proposed by Pandy et al. (1995) with values of 0.6 s and 3 s\(^{-1}\).

The optimal control problem was converted into a static optimization problem (Goh & Teo, 1988) by representing the time histories of neural excitation for each of the eight muscles as a series of control nodes separated by 0.1s. The neural excitations are related to muscle active states by a first order ordinary differential equation which creates a time delay between excitation and development of muscle force (Zajac, 1989). The active state directly scales muscle force. Muscle neural excitation between these nodes was determined by linear interpolation. Initial neural excitations were set at 0.005. Under this initial condition the force required to support the body was produced by the seat force, and ground reaction force. The simulations were allowed a maximum of two seconds to achieve upright stance; this was done to constrain the solution space to feasible movement times, and therefore speed simulation.

Sit to stand was simulated from four different seat heights. The shank and the trunk were vertical at the start of each of the simulations. Changing seat height was
accommodated by rotation of the thigh segment. Seat heights were selected so that the thigh angle was 100, 90, and 80 degrees; these corresponded to seat heights of 34.8 cm, 42 cm, and 49.6 cm respectively. At a thigh angle of 100 degrees, corresponding to a seat height of 34.8 cm, the model was not strong enough to be able to stand. An additional trial with the thigh angle at 95 degrees (seat height 38.4 cm) was therefore added.

Experimentally a number of different criteria have been used to determine when upright stance has been achieved during a sit to stand. Studies using strict kinematic criteria have found movement times much larger than what has been reported elsewhere in the literature (Gross et al., 1998), therefore to allow better comparison with values from the literature sit to stand movement termination was determined when whole body center of mass acceleration crossed zero with a positive gradient for the second time. This criterion has been successfully validated by comparing force plate data derived movement termination with those from a motion analysis recording (Burgess, 2003). Seat off was determined when the force applied to the seat fell to zero.

The model was validated by comparing basic kinematics and the time histories of the vertical ground reaction force of a typical subject from Burgess (2003) and the simulation from a seat height 42 cm. This is the seat height that corresponded to knee height.

5.4. Results

The simulation from the 42 cm seat produced a similar vertical ground reaction force as a typical subject from Burgess (2003). Table 5.2 presents mean characteristics of the ground reaction force curves from Burgess and from the model. Figure 5.1 shows a ground reaction force curve for a typical subject and for the model. Movement times were also similar. The model took 1.05 seconds to transition from seat off until upright stance, 11 adults with a mean age of 71.8 ± 3.6 in the study of Burgess (2003) took 1.13 ± 0.38 seconds.
The times in contact with the seat were similar for all three seat heights (Table 5.3), although time was slightly decreased from the highest seat height. The time to achieve upright stance as defined by the ground reaction force criteria increased as seat height decreased (Table 5.3).

**Table 5.2.** Mean characteristics of the ground reaction force curves from Burgess (2003) and from the model from a seat height of 42 cm. Forces are normalized with respect to body weight.

<table>
<thead>
<tr>
<th></th>
<th>Model</th>
<th>Experimental data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum Force</td>
<td>1.18</td>
<td>1.19 ± 0.07</td>
</tr>
<tr>
<td>Timing of Maximum Force</td>
<td>0.69 s</td>
<td>0.60 ± 0.23 s</td>
</tr>
<tr>
<td>Minimum Force</td>
<td>0.76</td>
<td>0.82 ± 0.09</td>
</tr>
<tr>
<td>Timing of Minimum Force</td>
<td>1.08 s</td>
<td>1.22 s ± 0.42 s</td>
</tr>
</tbody>
</table>

**Table 5.3.** Time in contact with the seat, and time to achieve upright stance for the simulations from the three seat heights.

<table>
<thead>
<tr>
<th></th>
<th>Seat Height</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>38.4 cm</td>
</tr>
<tr>
<td>Seat Contact Time</td>
<td>0.35 s</td>
</tr>
<tr>
<td>Time to Upright</td>
<td>1.47 s</td>
</tr>
</tbody>
</table>

Although the time in contact with the seat was similar for the three conditions, there were considerable differences in trunk movement prior to seat off. The hip angle at seat off increased as seat height decreased (Table 5.4). However because of the higher initial hip angle the resulting anterior trunk rotation decreased as seat height decreased (Table 5.4).
Figure 5.2. Vertical ground reaction force for a typical subject (top) from Burgess (2003), and the model (bottom) performing the sit to stand from a seat height of 42 cm. Vertical lines indicate when upright stance was achieved.

Table 5.4. Hip and trunk angles at seat off for the simulations from the three seat heights.

<table>
<thead>
<tr>
<th>Seat Height</th>
<th>Hip Angle</th>
<th>Trunk Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>38.4 cm</td>
<td>102°</td>
<td>7°</td>
</tr>
<tr>
<td>42 cm</td>
<td>101°</td>
<td>10°</td>
</tr>
<tr>
<td>49.6 cm</td>
<td>95°</td>
<td>15°</td>
</tr>
</tbody>
</table>

Maximum hip flexion velocity decreased as seat height decreased. The timing of the maximum hip flexion velocity was shortly before seat off in all trials (Table 5.5). The maximum knee and hip extension velocities increased as seat height decreased (Table 5.6).
Table 5.5. Maximum hip flexion velocity and its timing for the simulations from the three seat heights.

<table>
<thead>
<tr>
<th>Seat Height</th>
<th>38.4 cm</th>
<th>42 cm</th>
<th>49.6 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum Hip Flexion Velocity</td>
<td>37 °/s</td>
<td>54 °/s</td>
<td>85 °/s</td>
</tr>
<tr>
<td>Time</td>
<td>0.32 s</td>
<td>0.29 s</td>
<td>0.28 s</td>
</tr>
</tbody>
</table>

Table 5.6. Maximum hip and knee extension velocities for the simulations from the three seat heights.

<table>
<thead>
<tr>
<th>Seat Height</th>
<th>38.4 cm</th>
<th>42 cm</th>
<th>49.6 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum Hip Extension Velocity</td>
<td>194 °/s</td>
<td>134 °/s</td>
<td>113 °/s</td>
</tr>
<tr>
<td>Maximum Knee Extension Velocity</td>
<td>154 °/s</td>
<td>107 °/s</td>
<td>97 °/s</td>
</tr>
</tbody>
</table>

Maximum hip extension moment decreased as seat height decreased (Table 5.7). Maximum knee extension moment increased when decreasing seat height from 50.5 cm to 43 cm, however it slightly increased when moving to the lowest seat height, while the percentage of maximum moment increased with decreasing seat height. The maximum knee moments occurred after seat off. The knee moment at seat off increased as seat height decreased. Very large dorsiflexion moments were required shortly after seat off. They follow the same pattern as the knee extension moments.
Table 5.7. The maximum joint moments for the simulations from the three seat heights presented in Newton meters and as a percentage of maximum moment available at given joint angles and velocities in parenthesis.

<table>
<thead>
<tr>
<th></th>
<th>Seat Height</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>38.4 cm</td>
</tr>
<tr>
<td>Maximum Hip Extension Moment</td>
<td>70 Nm (67%)</td>
</tr>
<tr>
<td>Maximum Knee Extension Moment</td>
<td>266Nm (98%)</td>
</tr>
<tr>
<td>Knee Extension Moment at Seat Off</td>
<td>245Nm (93%)</td>
</tr>
<tr>
<td>Maximum Dorsiflexion Moment</td>
<td>266Nm (99%)</td>
</tr>
</tbody>
</table>

Note - Moments presented are for both legs.

5.5. Discussion

It has been suggested that increased forward movement of the trunk before seat off during the sit to stand is a possible strategy to help elderly individuals cope with the increased difficulty associated with standing from low seat heights. Weiner et al. (1993) suggested that increased anterior trunk movement would move the center of mass closer to the knees and decrease the required effort. Schenkman et al. (1996) suggested that a high forward velocity of the trunk at seat off could be used to provide momentum that would assist in the rest of the stand. Evidence from this study does not support adoption of either of these two strategies. Here as seat height decreased both the forward rotation and angular velocity of the trunk decreased. To accommodate the lower seat heights initial hip flexion had to increase; this additional hip flexion moved the hip extensors to a less favorable region of the force-length curve so they were less able to produce force, greater trunk motion would have exasperated this effect. In the simulation from the lowest seat height the gluteal muscles are stretched far enough at the point of maximum forward trunk movement that they can only produce 12 percent of their maximum isometric force. This is complicated by the knee extensors being near maximally stressed at this time as well, therefore activation of the hamstrings is undesirable. The gluteal muscles are therefore the only muscles available for hip extension. In the simulation from the
The results of the simulation model indicate that as seat height is reduced the moment producing capabilities of the hip musculature are compromised. The knee is also placed in a less favorable position, but while the subject is in contact with the seat there is little which can be done to alleviate this challenge to the knee musculature. Indeed the strategies involving anterior motion of the trunk would only place greater requirements on the knee musculature. Generating momentum with the trunk prior to seat off was not used a strategy by the model, and is one which elderly subjects should consider avoiding. Pai and Rogers (1990) have demonstrated that horizontal momentum is invariant irrespective of sit to stand speed, suggesting precise control of this variable is important to maintain stability particularly once upright. Maintenance of postural stability in the elderly can be a challenge (e.g., Maki and McIlroy, 1996), as such generating large amounts of trunk momentum is a potentially dangerous strategy for individuals with balance difficulties, as the increased forward momentum could lead to an increased incidence of forward falls.

The time to complete the movement increased as seat height decreased. However, even with the increase in time, there were increased maximum hip and knee extension velocities with decreasing seat height. These velocities are still greater even if they are normalized with respect to the range of motion. These increased velocities are accompanied by increased muscle active states, these two combinations make error correction during the sit to stand problematic. These factors could lead to an increased incidence of forward falls in sit to stand movements from low seat heights; they warrant further investigation.

The knee moment at seat off has been used to determine the knee extensor strength required to stand from a given seat height (e.g., Schultz et al., 1992). Maximum knee extension moment is actually greater at other instants in the movement, and the ratio of the seat off knee moment to the maximum knee moment in the simulations changed with seat height. The knee moment produced at seat off in the 39 cm trial was 93% of the maximum isometric moment available at that given joint angle. This is comparable with the work of Hughes et al. (1996) who reported
An unexpected finding from this study was the large dorsiflexion moment required shortly after seat off. This is necessary because of the large moment arm of the center of mass of the body relative to the ankle joint. In many laboratory situations the location of the feet is dictated to the subject, and often constrained by the location of a force plate. Kawagoe et al. (2000) and Shepard and Koh (1996) have both reported the hip extension moment is reduced with posterior foot placement. These studies examined only young subjects, so it is not possible to generalize these results to the elderly. The results of the present study suggest more work should be performed examining the effect of foot placement on sit to stand performance in an elderly population.

Increasing forward trunk movement prior to seat off has been proposed as a mechanism that may be used to reduce load of the knee extensors when standing from low seat heights. This is a potentially dangerous strategy as it could increase the likelihood of forward falls. The results from this study suggest that another complication in employing this strategy is that in sit to stand movements from low seat heights increased forward trunk movement moves the hip extensors into an unfavorable region of their force-length curve. Therefore the model demonstrated forward trunk movement actually decreased as seat height decreased. Instead of changes in trunk movement decreasing the load on the knee extensors, the load on the knee extensors is actually increased by the reduced anterior trunk movement. This is particularly unfortunate as these muscles are already highly stressed as a result of increased initial knee angles, which occurs with decreasing seat height.

All models are by definition simplifications of the systems they are designed to represent. The model used here is comparable in complexity to other models used for maximum vertical jumping (e.g., Bobbert and Van Zandwijk, 1999; Nagano and Gerritsen, 2001), and is based on the validated model of sit to stand of Pandy et al. (1995). The results presented here in large part depend on the accurate representation of the properties of the muscles. The moment-angle curves produced by the model correspond well with those in the literature (e.g. Kulig et al., 1984). But variations between subjects do exist, for example Savelberg and Meijer (2003) have shown that
the force-length curve of the rectus femoris varies between specific populations. However, the results of the simulation model suggest that as seat height decreases typical subjects would not benefit from increasing anterior trunk motion before seat off during the sit to stand.
5.6. References


CHAPTER 6
THE EFFECT OF STRENGTH TRAINING INDIVIDUAL MUSCLE GROUPS ON SIT TO STAND PERFORMANCE: A SIMULATION STUDY

6.1 Abstract
6.2 Introduction
6.2 Methods
6.3 Results
6.4 Discussion
6.5 References
6.1 Abstract

Rising from a seated position is a challenging activity of daily living for the elderly, with strength being an important determinant of sit to stand success. Strength training has been shown to produce significant strength gains in the elderly. It was hypothesized that there is one muscle group which limits sit to stand performance in the elderly. The purpose of this study was to determine what the limiting muscle group may be using a simulation model of sit to stand. Sit to stand was simulated using an optimal control direct dynamics muscle moment driven model. The model had three rigid links (shank, thigh, and a combined head/arms/trunk), actuated by eight muscle models representing the major muscle groups of the lower extremities. Muscle activations for each of the muscles were selected so that the time to rise was minimized. The model was “strength trained” by increasing the maximal isometric force produced by the muscles; and then the sit to stand simulated. Strength training the hip extensors and the knee extensors produced similar improvements in performance. These improvements were approximately one half of the improvement seen when training all muscles. There was potentially better control of the movement when all muscle groups were trained; as demonstrated by a reduction in peak joint extension angular velocities and a change in the timing of the maximum horizontal momentum. The results of these simulations suggest it is important to train all of the muscles rather than one specific muscle group for improving sit to stand performance.
6.2 Introduction

With an increasing proportion of the America population reaching old age, research improving functional ability in activities of daily living has become an important research focus. It has been shown that rising from a seated position is a challenging activity of daily living (e.g., Rodosky et al., 1989). Dawson et al. (1987) reported that more than two million people in the United States over the age of 64 years had difficulty in performing sit to stand. In fact many elderly are unable to rise from a chair without assistance (Alexander et al., 2000). Rising from a chair is clearly a precursor to gait and therefore is a significant determinant of independence.

Strength has been shown to be an important determinant of sit to stand ability. Hughes et al. (1996) showed that elderly subjects used 97 percent of their available knee strength to rise from the lowest chair height they were able to stand from. This is in agreement with simulation model results in Chapter 5, which showed the knee extensors are stressed heavily as seat height is lowered. Shenkman et al. (1996) and Corrigan and Bohannon (2001) showed strong correlations between strength, particularly knee extensor strength, and sit to stand performance, while Gross et al. (1998) showed that hip extensor strength may be a limiting factor in the time to complete sit to stand in elderly subjects.

Studies have shown that significant strength gains (100-200 percent) are possible in the sedentary and the frail elderly (Frontera et al., 1988; Fiatarone et al., 1990). While, smaller strength gains (20-42 percent) have been shown in the healthy elderly (Schlicht et al., 2001). The degree to which this strength gain can affect sit to stand performance is debatable. Alexander et al. (2001) found only small improvements in reducing sit to stand time following a 12-week lower body strength training program. Sullivan et al. (2001) showed substantial improvement in sit to stand ability in a group of frail elderly subjects following 10 weeks of training performing leg presses. Fifteen out of 19 subjects progressed from unable to stand to able, or decreased their sit to stand time by at least 25 percent. Recently Schot et al. (2003) showed a change in the kinematics of sit to stand following an eight-week
whole body strength training program in healthy elderly subjects. Following the training the kinematics progressed to being more like that of young subjects.

Adherence rates for exercise training programs in the elderly vary greatly (Pollock et al., 1991). Many studies examining adherence are of supervised exercise, which take place in a laboratory setting. These studies are likely to have high adherence rates as a result of the high level of supervision. King et al. (1991) compared adherence rates for exercise in a group based program, with adherence rates for a home based program of similar intensity in older adults. They found the home based program had a much higher adherence rate (79% compared with 53% for group based). When comparing individuals who had at least a 75 percent attendance, they also found larger performance gains in the home based training group. King and colleagues were examining aerobic exercise, but similar adherence rates for home based strength training have been reported (Mikesky et al., 1994; Capodaglio et al. 2002).

It was hypothesized that there is one muscle group in the elderly that limits sit to stand performance during maximum speed sit to stand. The purpose of this study was to determine what the limiting muscle group may be, and if identified to recommend a simple, home-based strength training program which could improve sit to stand performance in the elderly. The study is performed using a computer simulation model of rising from a chair.

6.3 Methods

Sit to stand was simulated using an optimal control direct dynamics muscle moment driven model. The model had three rigid links (shank, thigh, and the head arms and trunk), connected by frictionless hinge joints. The foot was assumed to be stationary throughout the movement and was therefore considered part of the ground. The chair was modeled as paired springs and dampers in vertical and horizontal directions (Pandy et al., 1995). The equations of motion of the three link system were formulated as mixed differential-algebraic equations (Haug, 1989). The model was actuated by eight muscle-like actuators representing the major muscle groups of the
lower extremity. The muscle models contained force-length and force-velocity properties as well as activation dynamics (Gallucci & Challis, 2002). Muscle model parameters were originally representative of a young healthy population (Chapter 3). Maximum isometric force and force-velocities parameters were adjusted to reflect elderly muscle (approximate age 70 years). Similarly body segment inertial parameters were representative of an elderly subject (height – 1.65 m, mass – 82kg). Additional model details can be found in Chapter 5.

Two potential performance criteria were identified that would allow for assessment of functional improvement: minimizing time to stand, and determining the lowest seat height the simulation was able to stand from at a self selected pace. Simulation time for determining the lowest possible seat height was prohibitive compared with the time required for minimizing stand time, therefore minimizing time to stand was selected as the performance criterion for the study. Time to rise is a very common experimental measure used to evaluate the relationship between strength and sit to stand ability (e.g., Gross et al., 1998; Corrigan and Bohannon, 2001; Alexander et al., 2001; Schlicht et al., 2001) and use of this objective function allows for easy comparison with the literature. Simulations were performed from a seat height of 42 cm corresponding to 100 percent of model knee height.

A genetic search algorithm (Goldberg, 1989) was used to select sequences of muscle model neural excitations for each of the modeled muscles which minimized model time to rise. The time series of muscle neural excitations was represented as a series of control nodes separated by 0.1s. This allows for the optimal control problem to be converted into a static optimization problem (Goh & Teo, 1988). Muscle neural excitation between these nodes was determined by linear interpolation. The neural excitations are related to muscle active states by a first order ordinary differential equation which creates a time delay between excitation and development of muscle force (Zajac, 1989). The active state directly scales muscle force. Initial body configuration was set so that the trunk and shank were vertical and the thigh was horizontal. In its initial position the force required to support the body was produced entirely by the seat force.

A stand score previously described in Chapter 5 was used to determine when upright stance was achieved. Upright stance was defined as a stand score of less than
0.25. The stand score was determined by equation 6.1. The coefficients were selected to allow motion that is seen during quite stance in the elderly (Burgess, 2003).

\[ ss = a \Delta p + b \cdot v \]  \hspace{1cm} (6.1)

Where:

- \( ss \) - stand score
- \( \Delta p \) - the distance between the center of mass and its position in upright stance
- \( v \) - absolute value of the magnitude of the center of mass velocity vector
- \( a, b \) - constants with values of 10 m\(^{-1}\) and 1 s.m\(^{-1}\) respectively

The effect individual muscles groups have on performance was analyzed by individually “strength training” the muscles in the model. This was done by increasing the maximal isometric force produced by the muscles. Simulations were performed with no increases in force, 25 percent increases in all of the lower extremity muscles, and a 25 percent increase in each of the following muscle groups: plantarflexors, dorsiflexors, knee extensors, and hip extensors. This strength gain was selected based on the low end of strength gains seen in Schlicht et al. (2001). Theirs was an eight-week training study with healthy subjects of a similar age (mean age 72 years) to that represented by the model.

Improvement in performance was evaluated by comparing time to rise values. Additional model outputs, joint angular velocities, muscles active states, and total body center of mass momentum, were evaluated to determine how increases in strength were exploited.

### 6.4 Results

The model has been previously validated for sit to stand at a self-selected pace (Chapter 5). Table 1 compares time to rise for the model, before strength training, with studies in the literature. The greater time required in Mourey et al. (2000) is likely because the subjects are slightly older the subjects in Vander Linden et al. (1994) (mean ages 75 and 69 years respectively). The time to rise found by Vander
Linden and colleagues was similar to the simulation model without any strength increases. Figure 6.1 shows a stick figure of the model standing from a chair without any strength increases.

**Table 6.1.** Time to rise for the simulation model compared with means (±standard deviations) reported in the literature for maximum speed sit to stands in the elderly.

<table>
<thead>
<tr>
<th>Study</th>
<th>Time to rise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Current study</td>
<td>1.19 s</td>
</tr>
<tr>
<td>Vander Linden et al. (1994)</td>
<td>1.20 ± 0.21 s</td>
</tr>
<tr>
<td>Mourey et al. (2000) (young group)</td>
<td>0.98 ± 0.14 s</td>
</tr>
<tr>
<td>Mourey et al. (2000) (elderly group)</td>
<td>1.40 ± 0.36 s</td>
</tr>
</tbody>
</table>

**Figure 6.1.** Stick figure of the model standing without any strength increases.
Strength training all of the muscles produced the largest improvement in sit to stand time (Table 6.2). Strength training the hip extensors and the knee extensors produced similar improvements. These improvements were about half the size of that elicited by strength training all of the muscles. Strength training the ankle musculature only produced very small improvements in sit to stand time. The time in contact with the seat was similar for all simulations (Table 6.2), but there were slight reductions here in the trials with increased hip extension and dorsiflexion strength.

**Table 6.2.** Time to rise and time to seat off for the model of sit to stand performance before strength training and as consequence of strength training.

<table>
<thead>
<tr>
<th>Type of strengthening</th>
<th>Time to rise</th>
<th>Time to seat off</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before Strength Training</td>
<td>1.19 s</td>
<td>0.36 s</td>
</tr>
<tr>
<td>All Muscles</td>
<td>1.11 s</td>
<td>0.37 s</td>
</tr>
<tr>
<td>Hip Extensors</td>
<td>1.15 s</td>
<td>0.34 s</td>
</tr>
<tr>
<td>Knee Extensors</td>
<td>1.16 s</td>
<td>0.36 s</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>1.18 s</td>
<td>0.33 s</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>1.19 s</td>
<td>0.36 s</td>
</tr>
</tbody>
</table>

There were no noteworthy changes in anterior rotation of the trunk prior to seat off (range 10-12 degrees of anterior trunk movement). However, there were changes in the trunk angular flexion velocity (Table 6.3). There was a slightly increased velocity in the trials with the strength trained hip extensors and dorsiflexors. There was a large decrease in trunk angular velocity in the simulation with all the muscles strength trained.

Maximum hip and knee extension angular velocities changed considerably for the different simulations (Table 6.4). Two things should be noted from these changes, first the simulation with hip extensors trained and the simulation with knee extensors trained are very similar and have the largest velocities. Second the simulation with all muscles trained has the smallest maximum knee extension velocity and a fairly small maximum hip extension velocity.
Table 6.3. Trunk flexion angular velocity at seat off during the sit to stand simulations, before strength training and as consequence of strength training.

<table>
<thead>
<tr>
<th>Type of strengthening</th>
<th>Trunk flexion velocity at seat off</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before Strength Training</td>
<td>43 °/s</td>
</tr>
<tr>
<td>All Muscles</td>
<td>28 °/s</td>
</tr>
<tr>
<td>Hip Extensors</td>
<td>48 °/s</td>
</tr>
<tr>
<td>Knee Extensors</td>
<td>42 °/s</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>50 °/s</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>44 °/s</td>
</tr>
</tbody>
</table>

Table 6.4. Maximum hip and knee extension angular velocities during the sit to stand simulations, before strength training and as consequence of strength training.

<table>
<thead>
<tr>
<th>Type of strengthening</th>
<th>Maximum hip extension velocity</th>
<th>Maximum knee extension velocity</th>
<th>Maximum dorsiflexion velocity</th>
<th>Maximum plantarflexion velocity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before Strength Training</td>
<td>244 °/s (1.11s)</td>
<td>405 °/s (1.12s)</td>
<td>77.3 °/s (0.70s)</td>
<td>172.4 °/s (1.12s)</td>
</tr>
<tr>
<td>All Muscles</td>
<td>258 °/s (0.66s)</td>
<td>355 °/s (1.03s)</td>
<td>99.5 °/s (0.54s)</td>
<td>155.5 °/s (1.04s)</td>
</tr>
<tr>
<td>Hip Extensors</td>
<td>306 °/s (1.08s)</td>
<td>491 °/s (1.09s)</td>
<td>81.1 °/s (0.68s)</td>
<td>209.7 °/s (1.09s)</td>
</tr>
<tr>
<td>Knee Extensors</td>
<td>308 °/s (1.09s)</td>
<td>485 °/s (1.10s)</td>
<td>62.4 °/s (0.71s)</td>
<td>206.6 °/s (1.10s)</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>237 °/s (1.09s)</td>
<td>431 °/s (1.13s)</td>
<td>71.2 °/s (0.63s)</td>
<td>165.8 °/s (1.11s)</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>259 °/s (1.12s)</td>
<td>390 °/s (1.11s)</td>
<td>72.1 °/s (0.70s)</td>
<td>184.1 °/s (1.13s)</td>
</tr>
</tbody>
</table>

Note: the timing of the velocity is in parentheses.
The values for maximal horizontal momentum are approximately one standard deviation above the mean from Pai and Rogers (1990), however the timing of maximum horizontal momentum occurs after loss of contact with the seat in the simulation model (Table 6.5), whereas the maximum horizontal momentum occurred before seat off in Pai and Rogers (1990). There were no large changes in the maximum forward center of mass momentum as a consequence of strength training. The largest increase (less than ten percent) was when all muscles were trained. The timing of the maximum momentum was nearly the same for all simulations performed, except when all muscles were trained. In this simulation the maximum forward momentum occurs much earlier.

Table 6.5. Maximum center of mass momentum and timing during the sit to stand simulations, before strength training and as consequence of strength training.

<table>
<thead>
<tr>
<th>Type of strengthening</th>
<th>Maximum center of mass momentum</th>
<th>Timing of maximum center of mass momentum</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pai and Rogers (1990)</td>
<td>0.56 ± 1.12 m/s</td>
<td>-0.09 ± 0.05 s</td>
</tr>
<tr>
<td>Before Strength Training</td>
<td>0.53 m/s</td>
<td>0.36 s</td>
</tr>
<tr>
<td>All Muscles</td>
<td>0.58 m/s</td>
<td>0.16 s</td>
</tr>
<tr>
<td>Hip Extensors</td>
<td>0.56 m/s</td>
<td>0.36 s</td>
</tr>
<tr>
<td>Knee Extensors</td>
<td>0.50 m/s</td>
<td>0.38 s</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>0.51 m/s</td>
<td>0.37 s</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>0.49 m/s</td>
<td>0.35 s</td>
</tr>
</tbody>
</table>

*Note* – Momentum is normalized with respect to body mass. Timing is expressed as time after seat off.

Examination of maximal active states shows that in general when an individual muscle group is trained the maximum active state for that muscle is reduced (Table 6.6). Therefore the strength gain is not entirely utilized. Of all the muscle groups the strengthened hip extensors had the maximal active states closest to their pre-strengthened values.
Table 6.6. Maximum muscle active states during the sit to stand simulations, before strength training and as consequence of strength training.

<table>
<thead>
<tr>
<th>Type of strengthening</th>
<th>Gastrocnemius</th>
<th>Soleus</th>
<th>Tibialis Anterior</th>
<th>Vasti Group</th>
<th>Rectus Femoris</th>
<th>Hamstrings</th>
<th>Gluteal Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before Strength Training</td>
<td>0.88</td>
<td>0.14</td>
<td>0.94</td>
<td>0.99</td>
<td>0.95</td>
<td>0.61</td>
<td>0.97</td>
</tr>
<tr>
<td>All Muscles</td>
<td>0.71</td>
<td>0.12</td>
<td>0.96</td>
<td>0.95</td>
<td>0.84</td>
<td>0.71</td>
<td>0.77</td>
</tr>
<tr>
<td>Hip Extensors</td>
<td>0.89</td>
<td>0.18</td>
<td>0.99</td>
<td>0.99</td>
<td>0.94</td>
<td>0.56</td>
<td>0.87</td>
</tr>
<tr>
<td>Knee Extensors</td>
<td>0.82</td>
<td>0.22</td>
<td>0.91</td>
<td>0.79</td>
<td>0.81</td>
<td>0.8</td>
<td>0.97</td>
</tr>
<tr>
<td>Dorsiflexors</td>
<td>0.95</td>
<td>0.14</td>
<td>0.80</td>
<td>1</td>
<td>0.93</td>
<td>0.75</td>
<td>0.97</td>
</tr>
<tr>
<td>Plantarflexors</td>
<td>0.69</td>
<td>0.14</td>
<td>0.96</td>
<td>1</td>
<td>0.92</td>
<td>0.61</td>
<td>0.99</td>
</tr>
</tbody>
</table>
6.5 Discussion

Training all muscle groups was shown to be more effective than training any individual group in reducing time to stand. Training the hip extensors had a larger effect on performance than training any other muscle group. The effect was half the size as training all of the muscle groups. This is a somewhat unexpected result as most of the literature has focused on the importance of knee extensor strength in sit to stand movements (Hughes et al., 1996; Shenkman et al., 1996; Corrigan and Bohannon, 2001). However, this is in agreement with Gross et al. (1998), who found a negative correlation between hip extensor strength and movement time during maximum speed sit to stand in elderly subjects. While training the hip extensors proved to be the most effective training program, training the knee extensors resulted in a similar improvement in performance. Training the dorsiflexors produced a small improvement in sit to stand performance, and training the plantarflexors produced no improvement suggesting the plantar flexors are not important in the sit to stand movement.

There are two components to the mechanism by which increased strength of the hip extensors was able to increase performance. First there was an increase in maximum hip flexion velocity compared with the simulation without any strength gains, however there was no increase compared with the simulation with the knee extensors trained. Secondly increasing hip extensor strength allowed for a slightly larger forward momentum of the trunk at seat off, by being better able to reverse this motion. This increased forward momentum reduced the load on the knee extensors and therefore permitted higher knee extension velocities.

The dorsiflexors are highly stressed (over 90 percent active) just after seat off to help prevent a backward fall. The time required to reach the high level of force required by the dorsiflexors is likely the reason for the similar times in contact with the seat for all conditions. Training the dorsiflexors increased performance primarily by decreasing the time in contact with the seat. The increased dorsiflexor strength allowed for this muscle group to reach the required force level sooner.
The finding that maximum active state of a muscle group is reduced in simulations where that muscle group was strength trained suggests that the entire strength gain cannot be utilized because of deficiencies in other muscles. The hip extensors were the muscle group least effected by this, and this may explain why training these muscles had the largest effect on movement time.

Perhaps the most interesting result from the study is the way in which the total body strength trained simulation decreased sit to stand time. There was a reduction in maximum knee extension velocity and only a small increase in maximum hip extension velocity when compare to the initial strength level simulation. Possibly suggesting the movement was easier to control following the strength training.

Maximum horizontal momentum of the center of mass has been identified as an important quantity to control to prevent falls following sit to stand (Pai and Rogers, 1990). The maximum horizontal momentum occurred later in the simulation model compared with Pai and Rogers. This is likely a consequence of the model not having any fear of falling. There were no large changes in maximum horizontal momentum among the different strength training strategies tested. This is consistent with the finding from Pai and Rogers that maximum horizontal momentum does not increase as movement speed increases. The timing of the maximum momentum was similar for all trials except for the trial with all muscles strengthened. The maximum horizontal momentum occurred much earlier in this trial. This is clearly an important aspect for control as it allows for a much longer time for reducing momentum.

The simulation was run again with strength increases in both the hip and knee extensor muscle groups, to determine if this could match what was seen when training all muscle groups. While the improvement in time to stand was equal to what was seen when all muscle groups were trained, the improved control of the movement, measured either by decreased joint extension angular velocities or by the timing of the horizontal momentum, was not seen.

The minimizing of the time to complete the sit to stand makes this a potentially challenging task for some of the elderly. This challenge and the importance of this
performance criterion has been recognized previously, as maximum speed sit to stand has been incorporated into a number of tests used to assess physical function in the elderly (e.g. Guralnik et al., 1994; Lin et al., 2001). The change in performance due to strength training was modest, less than 7 percent of movement time (< 0.1 seconds), while strength increases were 20%. The nature of the sit to stand task is different to other tasks often simulated (e.g., jumping - Van Soest et al., 1993; walking - Anderson, & Pandy, 2001) as it starts and finishes with all joint angular velocities of virtually zero. This nature of the task presents different challenges to the musculo-skeletal system, and makes it interesting to investigate. This study has highlighted that increasing strength can influence the ease of control of the movement, and the advantage of strengthening all relevant muscle groups not one specific muscle groups.

This model was designed to examine changes in functional improvement as a consequence of strength training, not to examine control. The finding of improved ease of control may be an important result, however this finding would be much more important if found in movements at a self selected pace, as these movements are much more commonly performed.

Although it was hoped this study would identify a single muscle group that when trained would greatly improve sit to stand performance, the results do not support this training strategy. Training both the hip and knee extensors allowed for similar improvements in minimizing time to stand, but it did not match the increased control that was accomplished by training all muscle groups. For elderly subjects increased control is likely to be of much large benefit than increasing performance, because falls can be so devastating. Additional study is required to determine if the increased ease of control seen when standing at maximum speed will carry over to movements performed at a self-selected pace. Following that, a recommendation needs to be made to the elderly that strength training should be performed to decrease the risk of falls following sit to stand.
6.4 References


CHAPTER 7
DISCUSSION

7.1 Summary
7.2 Limitations
7.3 Conclusions
7.4 Directions for Future Research
7.1. Summary

The purpose of this dissertation was to develop a simulation of sit to stand that can be used to guide strength training as a means to improve performance in the elderly. This dissertation was composed of four reports detailing separate investigations. The fourth investigation specifically dealt with the purpose of the dissertation. The first three investigations were designed to be steps in developing the model.

The first investigation examined the effect of squat depth on jumping performance. It was hypothesized that the subjects would jump highest from their self-selected squat depth and that the simulation would jump highest from a very deep squat. This hypothesis was found to be correct. The reason being that the model was optimally coordinated from any initial posture, whereas the subjects are likely to be optimally coordinated only from postures they have practiced from. If subjects were to train jumping from a very deep squat, their jump performance should improve.

The second report detailed an investigation into the mechanism by which arm swing improves vertical jump performance. Several mechanisms have been proposed to contribute to jump performance enhancement that accompanies arm swing. The first is that the arms have a positive vertical momentum relative to the shoulder at takeoff and this adds to jump height. Two other proposed mechanisms involve the moment at the shoulder joint slowing the extension of the legs. This would allow the muscles more time to develop force as well as operate on a more advantageous region of the force-velocity curve. It was hypothesized that each of the proposed mechanisms would contribute, but that the largest contribution would be from the kinetic energy of the arms at takeoff. Experimental work could not rule out any of the proposed mechanisms. A simulation model was used to calculate the work produced by the muscles under both jumping conditions. It was found that the work produced by the muscles of the lower extremity was actually reduced in jumps with arm swing and that the work done by the shoulder musculature was the entire cause for the increased jump performance.
The third study in the dissertation examined changes in movement coordination that accompany decreasing seat height in sit to stand in the elderly. A simulation model was used. Experimental work has suggested two strategies by which elderly individuals can cope with the increased difficulty in standing from low seat heights. The first is to increase the forward trunk rotation while in contact with the chair, thereby decreasing the distance between the center of mass of the body and the knee joint. The second strategy is to increase the forward momentum while in contact with the seat and allow this momentum to assist in standing. It was hypothesized that with decreasing seat height, trunk motion while in contact with the seat will increase, to reduce the load on the knee musculature but not for momentum generation. The results showed this hypothesis to be incorrect. The forward movement of the trunk before seat off decreased as seat height was lowered. This is because there is an increased initial hip angle when standing from lower seat heights. Forward trunk movement was decreased to keep the hip extensors from moving into an unfavorable region of their force-length curve.

The final report in the dissertation directly addressed the purpose of the dissertation. The effects of strength training individual muscle groups on maximum speed sit to stand were investigated. It was hypothesized that sit to stand was limited by an individual muscle group and that strength training these muscles would have large effects on performance. It was that found training the dorsiflexors, the knee extensors and the hip extensors all improved performance. The largest performance gains for any individual muscle groups were associated with training the hip extensors. A likely more important consequence of training all muscle groups is that while maximum speed was increased, maximum joint angular velocities were actually decreased. This suggests that the movement is potentially more easily controlled. The idea that balanced strength training improved control while rising from a chair is further supported by the finding that the maximum horizontal momentum occurs much earlier when all muscles are strength trained. This is an important finding because it should mean a decreased likelihood of falls.
7.2. Limitations

There were several limitations to the work in this dissertation which will be outlined in this section. The first limitation is common to all four studies and all studies using simulations in general. Simulation models are by definition simple representations of complex biological phenomena. This simplification is necessary to keep computational time manageable (Anderson et al., 1995). However, models need to be complex enough so that they are accurate enough to examine the movements investigated. Validations are performed by comparing output to experimental data to determine the model’s accuracy, however a perfect validation does not exist (Panjabi, 1979).

There are three major limitations that are a result of efforts to keep simulation time manageable. The first is that simulations as well as experimental analysis were limited to the sagital plane. This is a fairly common practice for both jumping and sit to stand studies, as there is very little out of plane movement in either of these activities (Lundin et al., 1995) and it greatly simplifies analysis. However, muscles generally do not cause movement in a single plane. Stabilizing muscles are required to keep motion planar. A two dimensional analysis does not allow for investigation of these effects.

The second limitation that is a direct result of reducing simulation time is the use of rigid tendons in the models for Chapters 5 and 6. The simulations of sit to stand are of similar complexity as the simulations of jumping. However they take much longer for two reasons. First the movement time is six to eight times longer. This translates directly to increases in simulation time. Secondly, as a result of the longer movement time, far more control nodes are necessary to accurately represent neural input. The increase in the number of parameters to be optimized results in far more iterations being required to find an optimal solution. Estimated completion time for one sit to stand simulation using an elastic tendon model was over 200 days under the programming environment used in this dissertation. The use of a rigid tendon model was required to make the study feasible.
and the findings from appendix E suggest the effect of using the rigid tendon should be small.

The last major limitation that is a function of keeping simulation time manageable is the objective function chosen for Chapter 6. This objective function needs to create a maximal movement so that functional improvement can be measured. The objective function chosen was to perform the movement as quickly as possible. Determining the lowest seat height from which the simulation could stand may be a more relevant objective function for an elderly population. However, determining lowest seat height would require multiple simulations for each strength increase tested. Simulations at a preferred speed using the objective function presented in Chapter 5 took approximately 20 days to complete. If five simulations for each strength increased tested were necessary to find the lowest seat height from which the model was able to stand the resulting simulation time would be nearly two years.

There are a fairly large number of sources for muscle model parameters for the lower extremity. However, very few of these sources offer enough information for a complete model. It is typically necessary to draw parameters from more than one source. As a result parameters often need to be adjusted to ensure compatibility. It is difficult to determine which parameters need to be adjusted, because compensating errors can produce moment profiles similar to those seen experimentally. A best effort was made to ensure muscle model parameters for this dissertation were representative of a normal individual, however compensating errors could still exist.

While there is difficulty in determining muscle model parameters for the lower extremity, the availability of data for the upper extremity or for an elderly population is much more scarce. For the shoulder muscles modeled in Chapter 4, this is not a large problem because only a phenomenological model was needed here. For Chapters 5 and 6 the lack of data on model parameters in the elderly was a bigger problem. Model parameters for these chapters were determined by adjusting values of the parameters used in Chapters 3 and 4. This procedure could not account for potential differences in muscle tissue loss amongst different muscles. If differential loss is seen, this could affect results,
particularly in Chapter 6. Strength training a particular muscle group could be advantageous if losses in this group are larger than losses in other muscle groups.

7.3. Conclusions

In this section the hypotheses for each of the studies will be restated and the specific conclusion for each of these hypotheses will be given.

**Study 1 hypothesis.**
The subjects would jump highest from their self-selected squat depth and that the simulation would jump highest from a very deep squat.

**Study 1 conclusion.**
This hypothesis was found to be correct. The reason being that the model was optimally coordinated from any initial posture, whereas the subjects are likely to be optimally coordinated only from postures they have practiced from.

**Study 2 hypothesis.**
Three mechanisms have been proposed to contribute to jump performance enhancement that accompanies arm swing. The first is that the arms have a positive vertical momentum relative to the shoulder at takeoff and this adds to jump height. Two other proposed mechanisms involve the moment at the shoulder joint slowing the extension of the legs. This would allow the muscles more time to develop force as well as operate on a more advantageous region of the force-velocity curve. The hypothesis for the second study was each of the proposed mechanisms by which arm swing can enhance performance would contribute, but that the largest contribution would be from the kinetic energy of the arms at takeoff.
Study 2 conclusion.
The hypothesis was found to be incorrect. The work produced by the muscles of the lower extremity was actually reduced in jumps with arm swing and that the work done by the shoulder musculature was the entire cause for the increased jump performance.

Study 3 hypothesis.
With decreasing seat height in the elderly, trunk motion while in contact with the seat will increase, to reduce the load on the knee musculature but not for momentum generation.

Study 3 conclusion.
The results showed this hypothesis to be incorrect. The forward movement of the trunk before seat off decreased as seat height was lowered. This is because there is an increased initial hip angle when standing from lower seat heights. Forward trunk movement was decreased to keep the hip extensors from moving into an unfavorable region of their force-length curve.

Study 4 hypothesis.
Sit to stand performance in the elderly is limited by an individual muscle group and that strength training these muscles would have large effects on performance.

Study 4 conclusion
It was found training the dorsiflexors, the knee extensors and the hip extensors all improved performance, however each of these improvements were much smaller than the improvement seen when training all groups. A likely more important consequence of training all muscle groups is the movement becomes potentially easier to control.
Dissertation Purpose

The purpose of this study was to develop a simulation of sit to stand that can be used to examine performance of the sit to stand in the elderly, and exploited to examine the influence of strength training on sit to stand performance in the elderly.

It was hoped that this dissertation could discover a single muscle group responsible for limiting sit to stand performance in the elderly. A recommendation could then be made to perform a simple training program that could be performed at home that could produce improvement in sit to stand performance similar to what could be accomplished with full body training. It was found that, while the knee extensors are the muscle group most highly stressed during sit to stand, training several muscle groups resulted in improved performance. None of the improvements seen when training individual muscle groups were as large as what was seen when training all lower extremity muscles. It was also found that whole body training improved ease of control during the movement, while training individual muscle groups did not improve control. The final recommendation regarding strength training is that elderly individuals should have a strength training program that works all muscles of the lower extremity to help ease the control of sit to stand movements.

7.4. Directions for future research

The results of the first investigation left an important unanswered question. It was shown that there is an advantage to jumping from a very deep squat. It is not clear if this would translate to improvements in performance in more complex sports skills. A study with athletes should be performed to determine if jumping from a deep squat can be incorporated into sport skills such as high jumping. There are also a number of future unrelated questions about jumping performance that can be answered using the simulation model used in this dissertation. The role of the biarticular gastrocnemius in jumping has been examined previously using the gastrocnemius (van Soest et al., 1993).
The effects of the biarticular hamstrings have not been investigated and could be done using this model.

The final two studies also left unanswered questions that merit additional research. The high dorsiflexion moment required during sit to stand was an unexpected finding. It is likely a result of the feet being placed directly under the knees at the beginning of the movement. This is fairly common for an experimental design because it allows the seat to be placed on one force plate and the feet on another so that inverse dynamics could be performed while there is contact with the seat. A more natural movement would move the feet posterior to the knees before beginning to stand. Kawagoe et al. (2000) showed that with posterior foot placement there is a decrease in Tibialis Anterior activity. It has also been shown that hip extension moment is reduced with posterior foot placement (Shephard and Koh, 1996; Kawagoe et al., 2000). An investigation on the effects of foot placement on sit to stand in the elderly has yet to be conducted. This needs to be done as the effects in an elderly population where the movement is near maximal may be very different from the effects in a young healthy population. Using a simulation for this investigation is a logical choice because of the difficulty in experimentally measuring ground reaction forces with posterior foot placement. The simulation model used in chapter 5 could be modified for this investigation.

The result that strength training improved ease of control while standing from a chair is an important finding, however this result was for maximum speed sit to stand. An investigation into the effects of strength training on sit to stand at preferred speed needs to be undertaken to determine if increased ease of control occurs here as well. As with the first proposed study, the simulation model from chapter 5 could be used to conduct this study, however if the simulation yields positive results an experimental study should examine this as well. It would also be interesting to determine if strength training has similar effects in other activities of daily living (e.g., walking, and stair climbing).
7.5. References


Appendix A
Muscle Model

This appendix describes the muscle model used in the simulations described in Chapters 3, 4, 5, and 6. It is based on the model presented in Challis and Kerwin (1994). The model consists of a contractile element in series with an elastic element. The model represents paralleled fibered muscle fibers. Section A.1 describes the contractile component of the muscle model, which represents the properties of the muscle fibers. The series elastic component is described in section A.2, this broadly represents the properties of the tendon in series with the muscle fibers. The simulation procedure is presented in section A.3. Details on the model parameters are found in appendix C.

A.1 The Contractile Component

The force produced by the muscle model \( F_m \) can be described using the following function (Challis, 1998),

\[
F_m = q F_{max} F_1 F_v
\]  

(A.1)

Where
- \( q \) – normalized active state of the muscle fibers
- \( F_{max} \) – maximum isometric force the muscle can produce
- \( F_1 \) – fraction of maximum isometric force muscle can produce at current length and \( F_v \) fraction of maximum isometric force muscle can produce for current length at current velocity.

Therefore the force produced by the muscle model was a function of the active state, maximum isometric force, and the force-length and force-velocity properties of the contractile element.
The muscle active state is then determined by solving the following ordinary differential equation from Pandy et. al. (1992),

\[ \dot{q} = t_{\text{rise}}(u - q_0)u + t_{\text{fall}}(u - (q_0 - q_{\text{min}}) - (u - q_0)u) \] (A.2)

Where,
- \( q \) - active state (0 \( \leq q \leq 1 \))
- \( t_{\text{rise}} \) - constant for time of rise of active state
- \( t_{\text{fall}} \) - constant for time of fall of active state
- \( q_{\text{min}} \) - minimum active state
- \( u \) – neural excitation (0 \( \leq u \leq 1 \))

and \( q_0 \) is the initial muscle activation.

Values for \( t_{\text{rise}} \) and \( t_{\text{fall}} \) were set at 0.05 and 0.2 respectively (Bobbert and van Zandwijk, 1999). \( q_{\text{min}} \) was set at 0.005 so that there is always some tension in the muscle, which provides greater stability in the ordinary differential equation (equation A.2). Figure A.1 shows how active state varies with neural excitation for a muscle with a \( t_{\text{rise}} \) of 0.05 and a \( t_{\text{fall}} \) of 0.20. Neural excitation is maximum for the first second and is then zero for another second. In this model, neural excitation effectively governs the active state, it can be thought of as effort.

The force a muscle fiber can generate is a function of its length (Gordon et. al., 1966). To characterize this relationship, it is necessary to know the optimal fiber length and a shape parameter, which specifies the working range of the muscle. Optimal fiber length is the length at which the fibers can produce maximum isometric force. The fraction of the maximum isometric force that a muscle can produce at any given fiber length is represented by the following equation,

\[ F_i = 1 - \left[ \frac{\text{If} - \text{If}_{\text{opt}}}{\text{w}.\text{If}_{\text{opt}}} \right]^2 \] (A.3)
Where,

\( l_f \) – current length of fiber
\( l_f_{\text{opt}} \) - optimal fiber length

and \( w \) indicates the maximum fraction of the optimum length at which the fiber can still produce force.

**Figure A.1.** The active state of a muscle plotted against time for a \( t_{\text{rise}} \) of 0.05 and a \( t_{\text{fall}} \) of 0.020.

As an example this equation with a \( w \) value of 0.56 gives the force-length relationship for a muscle shown in Figure A.2, where the muscle ceases to produce force at 44% of its optimum length, and 156% of its optimum length.
As the shortening velocity of a muscle increases, the force it can generate decreases with the force equal to zero at the maximum velocity of unloaded shortening. Muscle shortening was designated as a positive velocity. The force-velocity relationship for concentric muscle actions was determined using the equation of Hill (1938),

\[ F_v = \frac{V_{\text{max}} - V_f}{V_{\text{max}} + K \cdot V_f} \quad V_f \geq 0 \quad (A.4) \]

Where

- \( F_v \) – the fraction of the maximum isometric force a muscle can produce at a given fiber velocity
- \( V_{\text{max}} \) – maximum velocity of shortening of the fibers
- \( V_f \) – current velocity of shortening of the fibers
- \( K \) – constant specifying the degree of curvature of the force-velocity curve
As the magnitude of the lengthening velocity increases, the force a muscle can generate increases. The peak force a muscle can produce while lengthening at high velocities is typically around one and a half times the maximum isometric force (Fitzhugh, 1977). The relationship for eccentric muscle action was determined using the equation of Fitzhugh (1977),

\[
F_v = 1.5 - 0.5 \frac{V_{\text{max}} - V_f}{V_{\text{max}} - (2K.V_f)} \quad V_f < 0
\]  

(A.5)

A typical force-velocity relationship for a muscle is shown in Figure A.3.
A.2 The Series Elastic Component

There is elasticity in both the cross-bridges (Huxley and Simmons, 1971) and the tendon (Bennett et. al., 1986). Practically, it is of little importance if some of the elasticity in the system comes from the muscle fibers as this elasticity can be incorporated within the model series elastic component. In the following model description this element will be referred to as the tendon.

The stress stain curve of tendon is nonlinear (Fung, 1981). However outside of the so called “toe region”, which is exceeded for moderate loads, the curve is very close to linear for the range of forces expected in the simulations (Bennett et al., 1986). The stress-strain relationship of the tendon was assumed to be linear, therefore,

![Figure A.3. A normalized force-velocity relationship for a muscle.](image)
\[ L_i = L_{tr} \left[ 1.0 + \frac{F_m}{A_t \cdot E} \right] \]  
(A.6)

Where,

- \( L_i \) - length of the tendon
- \( L_{tr} \) - resting length of the tendon
- \( A_t \) - cross-sectional area of the tendon
- \( E \) - Young's modulus of elasticity for tendon.

By introducing the constant \( C \), which represents the fraction of its resting length to which the maximum isometric muscle force extends the tendon, the equation can be simplified as,

\[ C = \frac{L_i - L_{tr}}{L_{tr}} = \frac{F_{max}}{A_t \cdot E} \]  
(A.7)

The final equation representing the tendon properties is then:

\[ L_i = L_{tr} + L_{tr} \cdot C \frac{F_m}{F_{max}} \]  
(A.8)

For example, Figure A.4 shows a force length curve for a tendon with 0.04 as a value for \( C \). The stiffness of the tendon is constant, and can then be determined by dividing muscle force by the change in length of the tendon at that force.

**A.3 Simulation Procedure**

The first step in simulation was determining active state using Equation A.2. The initial values for neural excitation were selected so that the model would be in static equilibrium. This was done by computing the initial resultant moments at the ankle, knee, and hip joints. The muscle forces required to produce these moments were computed in the following fashion. First the contribution of the biarticular muscles to
each joints resultant moment were allocated based on the fraction each of these muscles could contribute to a maximum isometric moment at the joints they crossed. The contributions of the mono-articular muscles were then computed so that the initial squat resultant joint moments were produced.

![Figure A.4](image)

**Figure A.4.** A normalized strain versus muscle force plot for a tendon.

The second step in simulation was determining the normalized force-velocity relationship by rearranging Equation A.1 to the following form:

\[
F_v = \frac{F_m}{q.F_{max}.F_i}
\]  

(A.9)

The velocity of the muscle fibers could then be determined by rearranging Equations A.4 and A.5 to the following,

\[
V_f = \frac{V_{max} \cdot F_v - V_{max}}{F_v \cdot K + 1} \quad V_f \geq 0
\]  

(A.10)
The velocity of the muscle-tendon complex was determined by taking the first order finite difference of the muscle-tendon length determined in the direct dynamics simulation. The velocity of the tendon could then be determined by the following equation,

\[ V_t = V_{mt} - V_f \]  \hspace{1cm} (A.12)

Where,

\( V_t \) – velocity of the tendon

\( V_{mt} \) – velocity of the muscle-tendon complex

Muscle force is related to tendon length according to the following equation,

\[ F_m = k \Delta L_t \]  \hspace{1cm} (A.13)

Where,

\( k \) – stiffness of the tendon

\( L_t \) – length of the tendon

The rate of change of the muscle force is then determined by differentiating Equation A.13 with respect to time,

\[ \dot{F}_m = k V_t \]  \hspace{1cm} (A.14)

Where,

\( \dot{F}_m \) – first derivative of the muscle force

The muscle force is then determined by numerically integrating this first order ordinary differential equation. The muscle force is then input into the direct dynamics simulation (Appendix B). A summary of the muscle model is presented in Figure A.5.
Figure A.5. The flow of variables in the muscle model.
A.4. References


Appendix B
Rigid Body Model

The following appendix describes the formulation of the direct dynamics
equations of motion and the equations used to predict external forces acting on the
system.

B.1. Direct Dynamics Equations of Motion

The direct dynamics portion of the model takes the muscle forces from the muscle
models and predicts the generalized coordinates for each time step. Then from these
coordinates the muscle lengths and velocities are computed and then input back into the
muscle model.

Examples are given using a planar two-link system connected to the ground by a
pin joint with another point joint connecting the segments (Figure B.1). Local reference
frames are attached to each segment at the segment center of mass with the y-axis
directed along the segment. Segment angles are defined as the angle between the local x-
axis and the global x-axis.

The direct dynamics are formulated using a mixed algebraic differential approach
(Haug, 1989). Using Haug’s method there are three generalized coordinates for each
segment for a planar analysis. For the example system the vector of generalized
coordinates (q) are given in Equation B.1. Generalized coordinates are given in the
global reference frame.

\[
q = \begin{bmatrix}
x_1 \\
y_1 \\
\phi_1 \\
x_2 \\
y_2 \\
\phi_2 
\end{bmatrix}
\]  

\text{(B.1)}
Where: $X_1$ – x coordinate of segment 1
$Y_1$ – y coordinate of segment 1
$\phi_1$ – orientation of segment 1
$X_2$ – x coordinate of segment 2
$Y_2$ – y coordinate of segment 2
$\phi_2$ – orientation of segment 2

Figure B.1. Example of a two-link system connected to the ground by a pin joint with another point joint connecting the segments
A two-link system has a total of six degrees of freedom, however four are restricted by the pin joints in the example system. A joint constraint for every limited degree of freedom must also be defined. The joint constraint vector for the example system is presented in Equation B.2. Joint constraints for pin joints specify that locations of the joint on connected bodies be at the same global location.

\[
\Phi = \begin{bmatrix}
x_1^A = 0 \\
y_1^A = 0 \\
x_1^B - x_2^B = 0 \\
y_1^B - y_2^B = 0 
\end{bmatrix}
\]  

(B.2)

Where:  
- \( x_1^A \) – x coordinate of point A on segment 1 in a global reference
- \( y_1^A \) – y coordinate of point A on segment 1 in a global reference
- \( x_1^B \) – x coordinate of point B on segment 1 in a global reference
- \( y_1^B \) – y coordinate of point B on segment 1 in a global reference
- \( x_2^B \) – x coordinate of point B on segment 2 in a global reference
- \( y_2^B \) – y coordinate of point B on segment 2 in a global reference

The basis for the model is Equation B.3

\[
\begin{bmatrix}
\ddot{q} \\
\lambda
\end{bmatrix} = \begin{bmatrix}
[M] & [J^T]^{-1} \\
[J] & [0]
\end{bmatrix} \cdot \begin{bmatrix}
Q \\
\gamma
\end{bmatrix}
\]  

(B.3)
where: $\ddot{\mathbf{q}}$ – Generalized accelerations \([nq \times 1]\) 

$\lambda$ - Joint constraint forces \([nc \times 1]\) 

$\mathbf{M}$ - Mass matrix \([nq \times nq]\) (B.4) 

$\mathbf{J}$ - Jacobian \([nc \times nq]\) (B.5) 

$\mathbf{0}$ - Matrix full of zeros \([nc \times nc]\) 

$\mathbf{Q}$ - Vector containing externally applied forces and moments \([nq \times 1]\) 

$\gamma$ - Acceleration right hand side vector \([nc \times 1]\) (B.6) 

*note*: \(nq\) is the number of generalized coordinates and \(nc\) is the number of joint constraints. 

The mass matrix was made by forming a square matrix of zeros of order number of segments times three. The diagonal of this matrix is then filled with the masses and moments of inertia of the segments. A mass matrix for the example system is presented in Equation B.4.

$$
\begin{bmatrix}
m_1 & 0 & 0 & 0 & 0 & 0 \\
0 & m_1 & 0 & 0 & 0 & 0 \\
0 & 0 & j_1 & 0 & 0 & 0 \\
0 & 0 & 0 & m_2 & 0 & 0 \\
0 & 0 & 0 & 0 & m_2 & 0 \\
0 & 0 & 0 & 0 & 0 & j_2
\end{bmatrix}
$$

(B.4)

where: $m_1$ – Mass of segment 1 

$j_1$ - Moment of inertia about the center of mass of segment 1 

$m_2$ – Mass of segment 2 

$j_2$ - Moment of inertia about the center of mass of segment 2
The Jacobian is a matrix made of partial derivatives of the joint constraints with respect to the generalized coordinates. The Jacobian does two things in this formulation. First it determines how intersegmental dynamics affect generalized accelerations, and second it determines what joint constraint force are caused by external forces. A Jacobian for the example system is presented in Equation B.5.

The applied forces and moments come from four potential sources. The weight of each segment is an external force applied in the vertical direction. The second source is muscle moments. These moments are the product of the muscle moment arms and the forces coming from the muscle model. The muscle moment arms are computed by inputting the current joint angles into the equations of Visser et al. (1990). An external moment is applied by the rotational spring damper at the model’s metatarsal phalangeal joint in the jumping simulations. In the sit to stand simulations an external force is applied at the hip by the chair.

\( \gamma \) is known as the acceleration right hand side vector (Haug 1989). It can be derived from Equation B.6.

\[
\gamma = -([J] \ddot{q})_q \ddot{q} - 2[J]_T \dot{q} - \Phi_{TT} \tag{B.6}
\]

where:

- \( \ddot{q} \) – Vector of generalized velocities
- \( T \) - Time

note: subscripts indicate derivatives with respect to

For pin joint constraints this vector is made up of terms that represent the normal accelerations. The acceleration right hand vector for the example system is presented in Equation B.7.

\[
\gamma = \begin{bmatrix}
0 - \cos \phi_1 \cdot x_1^A \dot{\phi}_1^2 + \sin \phi_1 \cdot y_1^A \dot{\phi}_1^2 \\
0 - \sin \phi_1 \cdot x_1^A \dot{\phi}_1^2 - \cos \phi_1 \cdot y_1^A \dot{\phi}_1^2 \\
\cos \phi_1 \cdot x_1^B \dot{\phi}_1^2 - \sin \phi_1 \cdot y_1^B \dot{\phi}_1^2 - \cos \phi_2 \cdot x_2^B \dot{\phi}_2^2 + \sin \phi_2 \cdot y_2^B \dot{\phi}_2^2 \\
\sin \phi_1 \cdot x_1^B \dot{\phi}_1^2 + \cos \phi_1 \cdot y_1^B \dot{\phi}_1^2 - \sin \phi_2 \cdot x_2^B \dot{\phi}_2^2 - \cos \phi_2 \cdot y_2^B \dot{\phi}_2^2 
\end{bmatrix} \tag{B.7}
\]
\[ [J] = \begin{bmatrix}
-1 & 0 & x'_A \cdot \sin \phi_1 + y'_A \cdot \cos \phi_1 & 0 & 0 & 0 \\
0 & -1 & y'_A \cdot \sin \phi_1 - x'_A \cdot \cos \phi_1 & 0 & 0 & 0 \\
1 & 0 & -x'_B \cdot \sin \phi_1 - y'_B \cdot \cos \phi_1 & -1 & 0 & x'_B \cdot \sin \phi_2 + y'_B \cdot \cos \phi_2 \\
0 & 1 & x'_B \cdot \cos \phi_1 - y'_B \cdot \sin \phi_1 & 0 & -1 & y'_B \cdot \sin \phi_2 - x'_B \cdot \cos \phi_2 
\end{bmatrix} \]  

where:  
\( x'_A, y'_A \) – Coordinates of pin joint A with respect to local frame 1  
\( x'_B, y'_B \) – Coordinates of pin joint B with respect to local frame 1  
\( x'_B, y'_B \) – Coordinates of pin joint B with respect to local frame 2
The angular accelerations are taken from the generalized accelerations and integrated twice to get segment angles. This is less computationally demanding than integrating for all of the generalized coordinates and it ensures kinematic consistency. The rest of the generalized coordinates were then determined using forward kinematics.

The improved Euler method (Ross, 1989) was used for numerical integration. Initially a variable step size fifth order Runge-Kutta algorithm was used, however because the time step required by the muscle model was very small (0.0001) there was no difference in accuracy between these integrators for the direct dynamics. The improved Euler method was chosen because it was significantly faster.

A summary of the direct dynamics simulation is presented in Figure B.2.
Figure B.2. Flow chart summarizing the direct dynamics simulation.
B.2. References


Appendix C
Model Parameters

The muscle model parameters used in the simulations described in Chapters 3, 4, 5, and 6 are given as tables in each chapter. This appendix provides additional detail as well as a graphical representation of the changes in muscle length, moment arm, and maximum isometric moment producing capabilities that occur with changes in joint angle. Section C.1 gives the body segment inertial parameters used in Chapters 3, 4, 5, and 6. Muscle lengths and moment arms are presented in section C.2. Section C.3 presents the moment angle relationship for each of the muscles modeled.

C.1 Body Segment Inertial Parameters

Table C.1 presents the body segment inertial parameters used in chapters 3 and 4. They are based on a typical subject from chapter 3, representing a subject of 1.65 m height and 65.7 kg mass. The foot length represents the distance between the ankle and the metatarsal phalangeal joint. Table C.2 presents the body segment inertial parameters used in chapters 5 and 6 to represent an older subject. They are based on a typical subject from Burgess (2003), representing a subject of 1.65 m height and 82 kg mass.
Table C.1. Body segment inertial parameters used in chapters 3 and 4 to represent a young subject in the simulations of jumping.

<table>
<thead>
<tr>
<th></th>
<th>Length</th>
<th>Center of Mass Location</th>
<th>Mass</th>
<th>Moment of Inertia</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot</td>
<td>0.15 m</td>
<td>45.8 %</td>
<td>2.3 kg</td>
<td>0.01 kg.m²</td>
</tr>
<tr>
<td>Shank</td>
<td>0.42 m</td>
<td>42.3 %</td>
<td>6.6 kg</td>
<td>0.10 kg.m²</td>
</tr>
<tr>
<td>Thigh</td>
<td>0.415 m</td>
<td>43.9 %</td>
<td>18.3 kg</td>
<td>0.27 kg.m²</td>
</tr>
<tr>
<td>Trunk and Head</td>
<td>0.757 m</td>
<td>60.7 %</td>
<td>38.5 kg</td>
<td>1.60 kg.m²</td>
</tr>
<tr>
<td>Arms</td>
<td>0.753 m</td>
<td>39.3 %</td>
<td>8.3 kg</td>
<td>0.33 kg.m²</td>
</tr>
<tr>
<td>HAT</td>
<td>0.757 m</td>
<td>62.3 %</td>
<td>46.8 kg</td>
<td>2.01 kg.m²</td>
</tr>
</tbody>
</table>

Note – HAT refers to a single segment the head arms and trunk. Center of mass locations are expressed as the percentage of the segment length and are measured from the proximal joint. Moments of inertia are about a transverse axis through the segments center of mass.

Table C.2. Body segment inertial parameters used in chapters 5 and 6 to represent an older subject in the simulations of sit to stand.

<table>
<thead>
<tr>
<th></th>
<th>Length</th>
<th>Center of Mass Location</th>
<th>Mass</th>
<th>Moment of Inertia</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shank</td>
<td>0.42 m</td>
<td>43%</td>
<td>7.1 kg</td>
<td>0.14 kg.m²</td>
</tr>
<tr>
<td>Thigh</td>
<td>0.415 m</td>
<td>43%</td>
<td>16.9 kg</td>
<td>0.42 kg.m²</td>
</tr>
<tr>
<td>HAT</td>
<td>0.757 m</td>
<td>68%</td>
<td>55.7 kg</td>
<td>3.9 kg.m²</td>
</tr>
</tbody>
</table>

Note – HAT refers to a single segment the head arms and trunk. Center of mass locations are expressed as the percentage of the segment length and are measured from the proximal joint. Moments of inertia are about a transverse axis through the segments center of mass.
C.2 Muscle Lengths and Moment Arms

Change in muscle length as a result of change in joint angle was computed using modified equations from Visser et al. (1990). Change in length is added to a reference length (Lr) to get total muscle tendon length for a given joint angle. Biarticular muscles have a $\Delta L_{mt}$ for each joint crossed. The coefficients of the equations were adjusted so that the angle definitions agreed with the current study and so that the equations were scaled to the limb lengths used. All angles must be input in radians. The general form of the equations is presented in equation C.1. The coefficients used are presented in table C.3.

$$\Delta L_{mt} = c_0 + c_1 \theta + c_2 \theta^2$$

(C.1)

Where

$\Delta L_{mt}$ - change in length of the muscle tendon complex

$c_0, c_1, c_2$ - coefficients

$\theta$ - joint angle in radians

The first derivative of the change of length of the muscle with respect to the joint angle gives the moment arm of the muscle, therefore the moment arm can then be computed by equation C.2.

$$r = \frac{dL_{mt}}{d\theta} = c_1 + 2c_2 \theta$$

(C.2)

Where
Muscles, which were ankle plantar flexors, knee flexors, hip flexors, and shoulder flexors, were designated negative moments, and therefore have negative moment arms.

**Table C.3.** Parameters for determining muscle length and moment arm.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Distal Joint</th>
<th>Proximal Joint</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lr  c2 c1 c0</td>
<td>c2 c1 c0</td>
</tr>
<tr>
<td>Soleus</td>
<td>0.302</td>
<td>-8.411 -46.11  0.005 - - -</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>0.391</td>
<td>-8.411 -46.11  0.005 1.517 -19.22 27.14</td>
</tr>
<tr>
<td>Tibialis Anterior</td>
<td>0.296</td>
<td>8.411 46.11 -0.005 - - -</td>
</tr>
<tr>
<td>Vasti Group</td>
<td>0.199</td>
<td>-9.537 56.01 -2.121 - - -</td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>0.457</td>
<td>-8.038 57.60 -0.098 5.994 -35.76 -81.76</td>
</tr>
<tr>
<td>Hamstrings Group</td>
<td>0.448</td>
<td>0 -34.82 8.23 0.823 8.31 65.60 0.691</td>
</tr>
<tr>
<td>Gluteal Group</td>
<td>0.228</td>
<td>8.310 73.90 0.691 - - -</td>
</tr>
<tr>
<td>Hip Flexor</td>
<td>0.353</td>
<td>-8.310 -73.90 -0.691 - - -</td>
</tr>
<tr>
<td>Shoulder Extensor</td>
<td>0.248</td>
<td>0 21.11 -0.691 - - -</td>
</tr>
<tr>
<td>Shoulder Flexor</td>
<td>0.248</td>
<td>0 -21.11 0.691 - - -</td>
</tr>
</tbody>
</table>

Note: All coefficients are multiplied by $10^3$

Figures C.1 through C.4 present muscle length plotted against joint angle for each muscle modeled. Figures C.5 through C.8 present the moment arms. Biarticular muscles have plots for both joints crossed. In these plots the secondary joint is kept stationary at zero degrees.
Figure C.1. Muscle length plotted against ankle angle for muscles crossing the ankle joint.
Figure C.2. Muscle length plotted against knee angle for muscles crossing the knee joint.
Figure C.3. Muscle length plotted against hip angle for muscles crossing the hip joint.
Figure C.4. Muscle length plotted against shoulder angle for muscles crossing the shoulder joint.
Figure C.5. Moment arm plotted against ankle angle for muscles crossing the ankle joint. Note: Soleus and Gastrocnemius have the same moment arm.
Figure C.6. Moment arm plotted against knee angle for muscles crossing the knee joint.
Figure C.7. Moment arm plotted against hip angle for muscles crossing the hip joint.
Figure C.8. Moment arm plotted against shoulder angle for muscles crossing the shoulder joint.

C.3 Moment angle curves

The muscles modeled in chapters 3 and 4 are representations of young healthy muscles. These parameters will be described first. Then the adjustments made to make the model reflective of an elderly population will be presented.

Shape parameters for the force-length and force-velocity curves as well as tendon strain under maximum isometric force are taken directly from van Soest et al. (1993). Optimal fiber length and tendon length were originally based on the values presented by van Soest and colleagues. These values were first scaled to limb length and then joint
angle curves were plotted using these parameters. Adjustments in both optimal fiber length and resting tendon length were necessary to ensure the muscles produced force over a reasonable range of motion. The only muscle requiring major changes was the hamstrings group. The hamstrings produce force over a very small range of motion using parameters as presented by van Soest and colleagues. The value for optimal fiber length was increased by over 100% and corresponding decreases were made in the resting tendon length to bring muscle tendon length back to a reasonable value. Small increases in optimal fiber length were also made for the knee extensors. The ratio of maximum isometric forces for the different muscles was determined from Friedrich and Brand (1990). The magnitude was set so that the maximum moment producing capability was similar to values reported in Kulig et al. (1984).

Maximum isometric moment was calculated from equation C.3. Details on the force-length equation can be found in Appendix A.

\[ M_{\text{max}}(\theta) = F_{\text{max}} \cdot F_{l} \cdot r(\theta) \]  
(C.3)

Where

- \( M_{\text{max}}(\theta) \) – maximum isometric moment at joint angle \( \theta \)
- \( F_{\text{max}} \) - maximum isometric force the muscle can produce
- \( F_{l} \) – fraction of maximum isometric force muscle can produce at current length
- \( r(\theta) \) – moment arm of muscle at current joint angle \( \theta \)

The length of the fibers was determined by an iterative procedure. First fiber length was estimated by subtracting resting tendon length from the muscle tendon length at a given joint angle. This value was used to determine a maximum isometric force. The tendon length at that isometric force was then used to calculate a new estimate for fiber length. This procedure continued until a consistent value for maximum isometric force was produced.

The maximum isometric moment angle curves for muscles modeled in chapters 3 and 4 are presented first. Maximum isometric moment angle curves for the plantar
flexors are presented in Figure C.9. The curves for the knee extensors are presented in Figure C.10. The knee flexors are found in Figure C.11. Figure C.12 shows the curves for the hip extensors. The shoulder extensors and flexors are found in figures C.13 and C.14 respectively.

![Figure C.9. Moment versus angle curves for the young model plantarflexors.](image)

**Figure C.9.** Moment versus angle curves for the young model plantarflexors.
Figure C.10. Moment versus angle curves for the young model knee extensors.
Figure C.11. Moment versus angle curves for the young model knee flexors.
Figure C.12. Moment versus angle curves for the young model hip extensors.
Figure C.13. Moment versus angle curves for the young model shoulder extensors.
Figure C.14. Moment versus angle curves for the young model shoulder flexors.
The muscle models in chapters 5 and 6 were adjusted to reflect elderly muscle. The maximum isometric force was reduced by 40% and adjusting the force-velocity parameters to reflect a loss of fast-twitch muscle fibers. Details on this procedure are presented in Chapter 5. There was also a slight additional increase in the optimal fiber length of the hamstrings. This was done, because during sit to stand, the hamstrings were operating at a length beyond which they could produce force for much of the beginning of the movement. The iterative procedure for determining muscle length was not required for the elderly muscle model, as the tendon is rigid. The primary change in the isometric curves of the elderly muscle model compared to the young muscle model is a reduced maximum moment. There is a slight shape change in the curve as a result of the rigid tendon. The force-velocity parameters have no effects on these curves.

Maximum isometric joint angle curves for the plantar flexors are presented in Figure C.15. The dorsiflexors are presented in Figure C.16. The curves for the knee extensors are presented in Figure C.17. The knee flexors are found in Figure C.18. The hip extensors and flexors are found in figures C.19 and C.20 respectively.
Figure C.15. Moment versus angle curves for the old model plantarflexors.
Figure C.16. Moment versus angle curves for the old model dorsiflexors.
Figure C.17. Moment versus angle curves for the old model knee extensors.
Figure C.18. Moment versus angle curves for the old model knee flexors.
Figure C.19. Moment versus angle curves for the old model hip extensors.
Figure C.20. Moment versus angle curves for the old model hip flexors.
C.4. References


Appendix D
Optimal Controller

This appendix describes the genetic algorithm used to determine muscle activation patterns. Section D.1 describes the basic ideas of a genetic algorithm. The specific algorithm used in this dissertation is described in section D.2. Section D.3 describes the objective functions used.

D.1 Genetic Algorithms

Genetic algorithms are designed to mimic the process of natural selection. An objective function must be formulated, and assuming the function is to be maximized, the fitness of a potential solution to the problem is determined by the value it produces for the objective function. Given a set of potential solutions, solutions that produce higher values for the objective function are more likely to “survive” and pass on their “genes” to the next generation. Genetic algorithms have been shown to be very robust means of finding the optimal solution to complex problems (Goldberg, 1989). The simple genetic algorithm (SGA) was proposed by Holland (1975). There are four main components to this algorithm: encoding, selection, recombination, and mutation. These will be discussed with reference to Holland’s original implementation.

The process of encoding involves turning the variable to be optimized into a string of ones and zeros. This requires the values to have upper and lower bounds and fixed number of decimal points must be used. A different string is assigned to each possible value within the bounds. The length of the sting required is the base two logarithm of the number of possible values.

The selection process determines which variables are going to “survive”. The SGA uses roulette wheel selection. Each individual is assigned a slot of the wheel. The size of the slot for each individual is proportional to the value it produces for the objective function. Random numbers are generated between 0 and $2\pi$, indicating a slot, selection of a slot gives that individual one “offspring” in the next generation.
Crossover is the process of “mating” two individuals in the population and determining the resulting “offspring”. Individuals are paired randomly. The SGA uses single point crossover. This method will combine two strings of length \( L \). A random number \( x \) between 1 and \( L-1 \) is generated. The first \( x \) values of the resulting string come from the first “parent” and the remaining values from the second.

The final process of the SGA is mutation. Mutation changes a certain percentage of the bits in a string from a one to a zero or vice versa. If all of the individuals in a population ended up with the same value at any point along their string, it would be impossible for crossover to generate individuals with a different value.

### D.2 Genetic and Evolutionary Algorithm Toolbox (GEATbx)

GEATbx is designed for use in Matlab, and was written by Hartmut Pohlheim (http://www.geatbx.com/). The four main components of a genetic algorithm will be discussed: encoding, selection, recombination, and mutation. There are a number of options in this toolbox for selection, recombination, and mutation, only the options used for the work described in this dissertation will be outlined. In GEATbx encoding was not used. Each potential solution was represented by a string representing the timings of the muscle activations.

Selection is done by stochastic universal sampling (Baker, 1987). This is a random selection process similar to the roulette wheel selection in the SGA, however stochastic universal sampling produces results closer to the expected distribution (Baker, 1987). Individuals are placed along a line. As with roulette wheel selection the chance an individual will be selected for an offspring is proportional to the value it produces for the objective function. In stochastic universal sampling this is done by varying the amount of space each individual occupies along the line. For each offspring to be produced a pointer is placed on the line. These pointers are placed at a fixed distance apart. The location of the first pointer is selected at random. For each pointer that falls within an individual’s space on the line an offspring is assigned.
Pohlheim’s toolbox uses discrete value recombination (Mühlenbein and Schlierkamp-Voosen, 1993). As in the SGA individuals are mated randomly after selection. Each “parent” consists of a string of numerical values being optimized. For each “offspring” a string of random ones and twos of the same length as the “parents” is generated. For every space along the string occupied by a one the value is replaced by the corresponding value from the first “parent”. Spaces occupied by twos are replaced by corresponding values from the second “parent”.

As in the SGA mutation is important to increase the chances of finding a global optimum. A mutation rate is set that determines the likelihood of a mutation occurring (Bäck, 1993). The variable that is mutated can become any value within the upper and lower bounds. For all of the studies in this dissertation the mutation rate was set so that on average one variable per individual is mutated.

The stopping criterion for the algorithm was 100 generations without changing the value of the objective function. The algorithm would also stop if none of the muscle activation control nodes changed by at least 0.01 over 100 generations. Approximately 700 generations were typically required for jumping simulations. The sit to stand simulations took considerably longer, because of the additional control nodes. More than 2000 generations were required for each of the sit to stand simulations.

The time to complete the optimization was considerable. Each generation consisted of 80 individuals and took 10-15 minutes to finish on a 2.8 GHz Pentium 4 PC. Total simulation time for a jumping simulation was approximately one week. For sit to stand the simulation time was 15-20 days.
D.3. References


Appendix E

The influence of an elastic tendon on the force producing capabilities of a muscle during a dynamic movement

Chapters 5 and 6 contained simulations of sit to stand using a rigid tendon model. This was done to simplify the muscle model and therefore reduce simulation time. This appendix describes the experiment used to justify this simplification.

E.1 Introduction

Computer simulation has become a very powerful research tool in biomechanics. Simulation allows for the examination of factors which cannot be examined experimentally, for example changing the origin of the gastrocnemius so that it becomes an uniarticular muscle (van Soest et al., 1993). Simulation time for complex models can be extremely long even on powerful parallel processing computers (Anderson et al., 1995).

One possibility for reducing simulation time for models of the musculo-skeletal system is to perform the simulations with a rigid tendon. For many models of the musculo-skeletal system, integration step size is determined by the model of the muscles, not the model representing the rigid body mechanics. By using a rigid tendon one ordinary differential equation per modeled muscle is eliminated from the model, and integration step size can be larger. This is because muscle fiber length can be calculated directly from muscle tendon length and muscle fiber velocity is equal to muscle tendon velocity.

The effect tendon has on a muscles force producing capabilities depends on the activity and muscle architecture. Van Soest et al. (1995) examined isometric muscle action and showed that the tendon resting length relative to the optimal fiber length influenced how large an effect the tendon had on force output. However, tendon had very little effect on the force a muscle produces unless the neural input is varying greatly.
Bobbert (2001) showed that squat jump height is enhanced when simulated with more compliant tendons.

This study will use a simple one muscle model to examine the effects of tendon on the force producing capabilities of the muscle under different dynamic conditions. The effect of different muscle fiber to tendon length ratios will also be examined.

E.2 Methods

The model consists of a single muscle with force-length and force-velocity properties as well as activation dynamics (Gallucci & Challis, 2002). The muscle was given parameters similar to the vasti group (Chapter 5). The muscle model parameters are given in Table E.1.

<table>
<thead>
<tr>
<th></th>
<th>Fmax (N)</th>
<th>Lfopt (m)</th>
<th>W</th>
<th>Ltr (m)</th>
<th>C</th>
<th>Vmax (Lfopt/s)</th>
<th>K</th>
</tr>
</thead>
<tbody>
<tr>
<td>Vasti group</td>
<td>9000</td>
<td>0.128</td>
<td>0.56</td>
<td>0.14</td>
<td>0.04</td>
<td>5.26</td>
<td>12.45</td>
</tr>
</tbody>
</table>


Simulations were conducted where the muscle had to move a 100 kg point mass a distance that corresponds to the muscle moving from 90 degrees flexion to full extension. Initial muscle activation for all cases was selected to put the system in static equilibrium.

Three different tasks were simulated, the second and third were designed to represent what a muscle might do during sit to stand or a reaching movement. The tasks were,
**Task 1** - the movement was simulated with the muscle maximally activated throughout the range of motion. This would be a similar activation that may be seen during jumping.

**Task 2** - the movement was simulated with a performance criterion to minimize, it was the rate change of muscle force. This criterion has been used to simulate sit to stand (Pandy et al., 1995). In this simple model this objective function is the same as minimizing jerk, which has been proposed as an objective function for reaching (Flash and Hogan, 1985). The muscle was given a maximum time of one second in which to complete the movement.

**Task 3** - the movement was simulated with a performance criterion to minimize, it was time to complete the movement. This would be similar to standing or reaching as quickly as possible.

All simulations were performed first with a rigid tendon. For tasks 2 and 3 the neural excitation was selected by a genetic search algorithm (Goldberg, 1989). In both cases the optimization was constrained to find neural excitation patterns that resulted in the mass being stationary at the end point of the movement. After determining muscle neural excitation for the movement with a rigid tendon the same neural excitation was used to simulate the movement with an elastic tendon that would stretch four percent under maximum isometric force (van Soest et al., 1993). The changes in the force production throughout the range of motion were investigated. To determine the effects of different muscle fiber to tendon length ratios, the experiment was repeated with the resting length of the tendon increased by 0.5m.

Simulations were compared by computing the maximum difference between force output.

**E.3 Results**

An elastic tendon enhanced muscle force output during the fast movements and slightly suppressed force during the slow movement. The size of the effect depends on
both the ratio of muscle fiber tendon length and the type of movement performed. The effect is increased by lengthening the tendon or by increasing the speed of the movement. Table E.1 gives the maximum force increase as a result of an elastic tendon in each of the movements.

**Table E.2.** Maximum force enhancement as a consequence of tendon elasticity in each of the movements examined.

<table>
<thead>
<tr>
<th>Tendon length</th>
<th><strong>Task 1</strong> Maximal activation</th>
<th><strong>Task 2</strong> Rate change of muscle force minimized</th>
<th><strong>Task 3</strong> Movement time minimized</th>
</tr>
</thead>
<tbody>
<tr>
<td>Short Tendon</td>
<td>2.9 %</td>
<td>-0.04 %</td>
<td>1.0 %</td>
</tr>
<tr>
<td>Long Tendon</td>
<td>7.2 %</td>
<td>-2.0 %</td>
<td>3.3 %</td>
</tr>
</tbody>
</table>

The maximum force enhancement occurs fairly early in the movement. The force enhancement quickly drops and throughout much of the movement there is a slight reduction in force with an elastic tendon. Figures E.1, E.2, and E.3 show muscle force plotted against mass position for the maximal activation movement, the movement minimizing the rate change of muscle force, and the movement minimizing the time of the movement respectively.
Figure E.1. Muscle force plotted against mass position for the maximal activation movement.
Figure E.2. Muscle force plotted against mass position for the movement minimizing the rate change of muscle force.
E.4 Discussion

Tendon velocity is equal to the rate change of muscle force divided by the stiffness of the tendon, and in fast concentric movements there is a negative rate change of muscle force as velocity of movement increases. Therefore, during these movements tendon is shortening. This allows the muscle fibers to shorten at a reduced velocity muscle force is enhanced. The amount of force enhancement was therefore increased as the velocity of movement increased.

Tendon compliance is given in this study as a strain under maximum isometric force. The use of longer tendons increased the tendon compliance in absolute terms. This means that for any given rate change of muscle force tendon shortening velocity is
increased, and muscle force enhancement is also increased. These results are in agreement with the finding of increased jump heights with compliant tendons from Bobbert (2001).

Reaching and sit to stand can be very time consuming movements to simulate. A primary reason for this is that these movements take much longer than an explosive movement such as jumping. Because these movements take a long time to perform, muscle velocities are low. The results from this study show that given these conditions the effect of tendon on muscle force output is minimal. When simulating these movements, the use of a rigid tendon should be considered as a means of reducing simulation time. In explosive movements such as running or jumping an elastic tendon has larger effects on force output and simulations of these movements should include tendon elasticity.
E.5. References


References

Alexander RM. 1990. Optimum take-off techniques for high and long jumps. Philosophical Transactions of the Royal Society, Series B 329: 3-10


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