THESIS

EFFECT OF AGE ON GROUND REACTION FORCE-RELATED PARAMETERS DURING STAIR NEGOTIATION

Submitted by

Andrew C. Harms

Department of Health and Exercise Science

In partial fulfillment of the requirements

For the Degree of Master of Science

Colorado State University

Fort Collins, Colorado

Fall 2013

Master's Committee:

Advisor: Raoul F. Reiser II

Brian L. Tracy
David P. Green

ABSTRACT

EFFECT OF AGE ON GROUND REACTION FORCE-RELATED PARAMETERS

DURING STAIR NEGOTIATION

Stairs are a frequently encountered obstacle in daily life. The ability to negotiate stairs without difficulty or pain is important to quality of life. Although it is a simple task for healthy adults, ascending and descending stairs can be very challenging when motor functions are diminished (e.g., in older adults, persons with physical disabilities and in persons who have experienced trauma to their lower extremities). Because many individuals with neuromuscular impairments walk and ascend/descend stairs slowly, it is important to isolate functional task from other factors such as age, muscle weakness, etc. Previous studies have presented some data on kinematics, kinetics and muscles or groups of muscles that contribute to the specific subtasks of stair negotiation. However, the complete data for Ground Reaction Force (GRF) related parameters (GRF as well as free vertical moment and center of pressure) during stair negotiation across age groups at controlled speeds has yet to be established. The purpose of this thesis was to characterize the GRF related parameters during stair negotiation in healthy young and older adults while ascending and descending stairs at a controlled speed.

Ten healthy younger adults and nine healthy older adults each performed five trials of stair ascent and descent at a controlled speed (70 steps/min). A force platform was embedded in the center of the second step of a custom-built four step staircase. Subjects performed one step on the level starting surface before ascending or descending the stairs in a step-over-step manner with the right foot always contacting the force platform.

Results showed that significant differences were exhibited between stair ascent and descent, with greater variability during descent. Further, significant differences existed across age groups, with older adult's also typically exhibiting greater variability of measures. Most notable were the age-group

differences between loading and unloading slopes in the vertical GRF and center of pressure displacements/velocities at the beginning and end of stance, respectively. The present study is novel in the fact that it was the first to examine both differences in ascent and descent as well as age-related differences in force control at a controlled speed. Differences between groups during loading suggest reduced control of force while differences during unloading suggest a more cautious strategy in the older adults. The greater variability that existed during descent compared to ascent highlights the difficulty the ability to control eccentric contraction compared to ascent contraction, regardless of age. The age-related differences that were exhibited in the greater variability in older adults supports previous reports the lack of neuromuscular control during eccentric contractions as part of the natural aging process. Characterization of the biomechanics of stair negotiation in individuals with disabilities in clinical conditions may direct innovative rehabilitative therapies to target and strengthen impaired muscle groups so that these people can negotiate stairs with increased ease, control and independence.

ACKNOWLDEDGMENTS

I would sincerely like to express my gratitude to a number of people without whom this thesis would not have been written.

First, I would like to thank Dr. Raoul Reiser II, for being my advisor during my graduate career at Colorado State University and for his support with this particular project. I am especially grateful for his guidance for the role he played as my mentor. I would also like to acknowledgethe Clinical Biomechanics Laboratory for their help. I would also like to thank the Neuromuscular

Function Laboratory and their director Dr. Brian Tracy for their help; especially Amy Grossman and Bethany Murphy.

I would also like to thank the Health and Exercise Science Department for funding my graduate education at Colorado State University. I would like to thank my fellow teaching assistants and my students for helping me find meaning and reward in my life that I couldn't have found without them. I would like to thank them for giving me their friendship and encouragement.

Thanks also to my thesis committee, Dr. Raoul Reiser II, Dr. Brian Tracy and Dr. David Green for their direction, dedication, and invaluable advice along this project. I would also like to express my deep appreciation for the College of Health & Human Sciences for supporting the project with their Mini-Grant Program.

Finally, and most importantly, I would like to express my gratitude to my family and friends, who have provided support, inspiration and encouragement throughout my life and especially during this project.

TABLE OF CONTENTS

CHAPTER 1

INTRODUCTION

Stair negotiation is one of the most hazardous activities in everyday life for older adults [1]. For older adults the risk of serious injury has been shown to be greater for a fall on stairs than a fall occurring on level ground [2]. Falls during stair ascent and descent are the leading cause of injury related-hospitalization and death in persons older than 65 years of age [3, 4]. Three- quarter of all staircase accidents occurred during stair descent [5], supporting claims that this is the most hazardous aspect of stair negotiation [6]. Stair negotiation is an important task that is often essential for independent living. Examining the mechanics of stair negotiation may be useful for not only understanding the aging process in general, but potentially for assessment of dynamic stability, as well as risk assessment and early intervention.

Stair negotiation is a demanding task and is often problematic in older adults due to deteriorations of physical capacities as part of the natural aging process (mainly strength, flexibility, proprioception, and control of force). Older adults generally work at a higher percentage of their strength capacity relative to young when ascending and descending stairs [7]. Older adults also often cite stiffness and decreased movement when they discuss their inability to perform common tasks such as stair negotiation [8]. Declines in proprioceptive function represent a fundamental aspect of the aging process, thus increasing the susceptibility to falls during stair negotiation [9]. Further, older adults exhibit a reduced ability to control force and demonstrate increased motor output variability compared to younger adults [10]. It is particularly difficult for older adults to control eccentric contractions compared with concentric contractions [10, 11]. Eccentric muscle contractions, as compared to concentric contractions, are relied upon almost exclusively to successfully descend stairs [12, 13]. Alone or in combination, these reduced capacities may be related to the increased risk for falling when descending compared to ascending stairs.

Assessment of dynamic control of force may provide important insight into the motor and sensory contribution of balance control during stair negotiation. One advantage of dynamic task assessment is the possibility to obtain information on force control that simulates situations encountered in daily-life activities, such as stair negotiation. Force platforms are instruments that may be utilized to measure the Ground Reaction Force (GRF), Center of Pressure (CoP), and Free Vertical Moment (FVM) at the foot-ground interface. The broad use of force platforms in clinical settings for force control assessment suggests that they are easy to use, becoming more affordable and readily available [14].

Vertical and Anterior-Posterior GRFs have been compared between young adults and older adults in both stair ascent and descent [1, 15, 16]. These studies show us that older adults tend to use a more cautious stepping strategy except in the case of initial loading rate during descent, which is suggestive of reduced control of force during eccentric contractions. The differences in these forces across age groups are greater during stair descent compared to stair ascent [1]. This suggests that the joint moment requirement is probably also higher [17], highlighting the demanding physical nature of this task. Also, older adults might have severe impaired dynamic balance control leading to a more unpredictable and potentially dangerous descent when unaided. Further, no investigation of medial-lateral GRFs has been examined during stair negotiation, which may help explain the phenomena on dynamic balance in the medial-lateral direction.

Additionally the CoP and FVM are expected to further our understanding of force control and dynamic stability. CoP displacement and velocity used to make inferences about neurologic and biomechanic mechanisms of postural control [18]. The FVM is the movement initiated about the body's vertical-axis. Both of these variables have been used to determine age-differences in anticipatory postural adjustments [19], and stepping down [20]. However, to date, no studies have examined CoP and FVM during stair negotiation across age groups with a controlled speed. Further, the variability of these measures needs to be examined under controlled speed conditions. Trial-to-trial variability is important for collecting an appropriate number of trials for averaging purposes as well as its potential use as a parameter of force control.

Considering the importance of stair negotiation, the risks of the task to older adults, and our limited understanding of how force control may be affected with aging, the goals of this investigation were to examine the GRF, FVM, and CoP during both stair ascent and descent in young and older adults at the same speed. Based on the available literature it was hypothesized that differences in average values, as well as variability of measures will be greater during stair descent compared to stair ascent because of the difficulty to control eccentric contractions. Further, it was hypothesized that older adults will demonstrate differences in average values as well as in variability of measures compared to young adults, which are suggestive of not only reduced strength [21], but compromised control of movement. Understanding the GRF and its associated parameters may be useful for understanding the aging process and for assessment of dynamic stability and risk assessment for early intervention. By understanding the regions of risk associated with stair negotiation in the older adult population we hope to give rehabilitation professionals a basis for early intervention protocol for this aging population.

CHAPTER 2

LITERATURE REVIEW

Thesis outline and intent

The overall focus of this thesis is to evaluate biomechanical attributes of stair ascent and descent related to normal, healthy aging. The following comprehensive literature review examines the current state of knowledge in this area of research. Of particular interest are the Ground Reaction Force (GRF), Free Vertical Moment (FVM), and Center of Pressure (CoP) characteristics of stair ascent and descent in relation to how they might be affected by a person's age and their ability to control their muscle force.

Introduction

Older adults have identified stair negotiation as one of the most difficult tasks as part of the natural aging process and are one of the leading causes of fall related injuries [3, 6, 7]. In fact, stair falls account for more than 10% of fatal fall accidents [7]. The demands that stairs place on the musculoskeletal and cardiovascular systems are compounded by the need for input from the somatosensory, visual, and vestibular systems throughout in the task. Many of these contributing systems deteriorate with age, thus increasing the difficulty and risk of failure in a task that inherently involves exposure to significant danger.

Understanding the normal, age-related alterations in movement control is important to provide a benchmark against which rehabilitation professionals working with a wide range of people with physical limitations can gauge mobility. The description of motor impairment and the comprehension of their influence on stair negotiation would provide a basis for an early intervention protocol in the older adult population. However, the capacity to produce appropriate levels of force is only one aspect of successful stair negation. Being able to control these forces to the degree necessary for successful, safe task time and time again is another. Frailty prevention programs designed to maintain independence and quality of life need to understand and integrate both strength and motor control in order to be successful.

Falls in Older Adults

One of the major problems associated with aging is increased susceptibility to falls. Falls are the leading cause of injury-related hospitalization and death in persons older than 65 years of age [3, 4]. Injuries resulting from falls include soft tissue damage and limb fractures [22]. Other consequences of falls include a downward spiral of decreased mobility, reduced confidence, and death [22].

Falls in the older adult population are one of the most important public health problems for health care services in the 21st century [23]. Ascending and descending stairs were among the top five tasks that older individuals listed as being the most difficult [24]. Epidemiological studies report that most falls occur during locomotion [16, 25] with stair negotiation among the most challenging and hazardous types of locomotion for older people. Further, stair descent with predominantly eccentric muscle control appears to be even more dangerous than the ascent phase as 80% of stairway accidents occur in the descent [25] .

The ability to ambulate on stairs can be quite demanding when motor functions are reduced with aging [26]. In the older population, declined motor functions and loss of fine motor skills, osteoarthritis, neuromuscular dysfunction, and declined visual performance occur [27, 28]. The implications of loss of fine motor functions for stair negotiation in older persons have not been studied directly but are believed to contribute to the susceptibly of falls, especially during stair descent.

Increased postural sway in older adults when standing is well-documented [29-31] and research has linked greater amounts of postural sway to increased risk of falling. For instance, Fernie *et al*. (1982) found significantly greater average speed of sway in older adults who had fallen one or more times in a year compared with those who had not fallen [32]. Nearly eight million adults in the United States report balance disorders each year [33]. In adults over age 65, balance problems are linked to falls [3].

One-third of adults in this age group and over half of people over the age of 75 years fall each year [34]. Since motor processing information related to balance control during stair negotiation is limited in the older population, a better understanding of how stair negotiation perturbs gait stability is critical for reducing the incidence of falls among older people and maintaining independence.

Effort Required to Negotiate Stairs

An older adult's ability to execute activities of daily living declines with age [7]. Activities of daily living are basic, routine tasks, such as walking, rising from a chair and stair negotiation that most people are able to perform on a daily basis without assistance. An older individual's inability to adequately perform these activities is often reflective of that person's ability to live safely and independently [3]. Assessing functional abilities in the older population helps determine that person's current needs. One possible reason for this decline is that the execution of customary motor tasks such as ascending or descending stairs requires a substantially greater effort in older adults compared to young adults relative to their available maximal capacity in strength and cardiovascular effort.

Aging is associated with progressive loss of strength that often leads to disability and loss of independence. By the seventh and eighth decade of life, maximal voluntary contractile strength is decreased, on average, by 20-40% for both men and women in proximal and distal muscles [35]. Although age-associated decreases in strength per unit muscle mass, or muscle quality, may play a role, the majority of strength loss can be accounted for by decreased muscle mass [36]. Multiple factors lead to the reduction of muscle mass (aka, Sarcopenia) and the associated impact on function. Sarcopenia is a major factor contributing to decreased functional independence and mobility [37].

Older adults operate much closer to their maximum capacities than young adults when ascending stairs due to their lower maximum musculoskeletal capacities [38]. This suggests that older adults require a greater intensity of effort, leaving less reserve muscle strength capacity, which may compromise their comfort and perceived stability while climbing stairs. During stair ascent, older adults operate at a higher level of maximum knee extensor output (75%) than young adults (53%) [39].

Intuitively it appears that the quantification of relative effort in basic daily activities in terms of joint moments should bring us closer to better understanding the causes of mobility limitations with age. The ability to ascend and descend stairs successfully requires greater strength in the lower extremity than is needed for many other activities of daily living [40]. The ability to ascend and descend stairs may require joint moments that exceed the available maximal levels in some healthy older adults and particularly in frail individuals [6]. It is therefore possible that maximum eccentric strength at the knee and/or the ankle are limiting factors in stair descent in older individuals [6].

There is a wealth of data on cardiovascular function indicating that, due to a reduction in peak oxygen uptake and impairment of the oxygen delivery system, older adults walk at a significantly higher percentage of their peak oxygen uptake. Older adults when walking operate at approximately 50% of their maximal capacity, compared with young adults who walk at about 30% [41-43]. A significant increase with age in the physiologic relative effort (i.e. the level of effort needed to execute a task as a percentage of the available maximal capacity) forces old adults to operate at high effort levels, causes premature fatigue, and in some cases leads to motor accidents [7]. Individuals over the age of 75 are four times more likely to be involved in fatigue related motor accidents [44]. Knowledge of the cardiovascular relative effort for the executions of activities of daily living is important, especially for exercise prescription.

Gait Cycle of Stair Negotiation

Walking speed is a fundamental parameter of human motion and is increasingly considered as an important indicator of health status. Biomechanical factors such as mechanical work, stability, and joint or muscle forces influence walking speeds. Faster walking requires additional external mechanical work per step [45], which includes the applied GRF. As faster walking is accomplished with both longer and faster steps, internal mechanical work also increases with increasing walking speed [46]. Internal work includes the gravity force. Therefore, both internal and external mechanical work per step increases with increasing speed. Individuals may try to reduce either external or internal mechanical work by walking more slowly, or may select a speed at which mechanical energy recovery is at a maximum [47].

Stability may be a factor influencing speed selection. Hunter *et al.* (2010) [48] showed that individuals use energetically suboptimal gaits when walking downhill. This suggests that people may instead choose gait parameters that maximize stability while walking downhill. Thus, under adverse conditions such as down hills, gait patterns may favor stability over speed [48], which might also be related to form of contraction since downhill gait relies more on eccentric force control.

Past studies of the walking speed on a flat surface have been conducted and investigated factors affecting walking speed. Himann *et al.* (1988) reported that the walking speed was associated with height before 62 years old, and with height and age after 62 years old [49]. At this age, adults exhibit deteriorations of physical capacities (mainly strength, flexibility, proprioception, and control of force).

Gait Pattern

Generally, healthy individuals use a step-over step gait pattern during stair negotiation; however older adults may be forced to adjust their stair gait pattern because of decrements in muscular strength [50, 51], decrease in proprioceptive acuity [50], and altered balance mechanisms [51] associated with age [6]. Therefore, older adults often adopt alternate gait patterns, such as increased handrail use, sideways motion, or a step-by-step pattern (placing both feet on the same step before ascending or descending) that deviate from the traditional step-over-step gait pattern [6, 52]. These deviations in stair gait patterns result in higher energy costs and lower efficiency, particularly during stair descent and step-by-step gait pattern [6, 52, 53]. Maintenance of the step-over-step stair negotiation pattern is of primary concern for independence and mobility.

During stair negotiation, as during walking, the legs move in a cyclical pattern. The cycle for both ascent and descent is divided into two distinct phases; the stance phase and the swing phase (Figure 2.1) [12, 13]. In ascent, the stance phase has three sub-phases: weight acceptance (shifting the body into an optimal position to be pulled up); pull-up (progression to full support on the next step); and forward continuance (ascent of a step has been completed and progression continues). The swing phase is subdivided into two sub-phases: foot clearance (the leg is raised to clear of the intermediate step); and foot placement (the swing leg is positioned for foot placement on the next step) (Figure 2.1a).

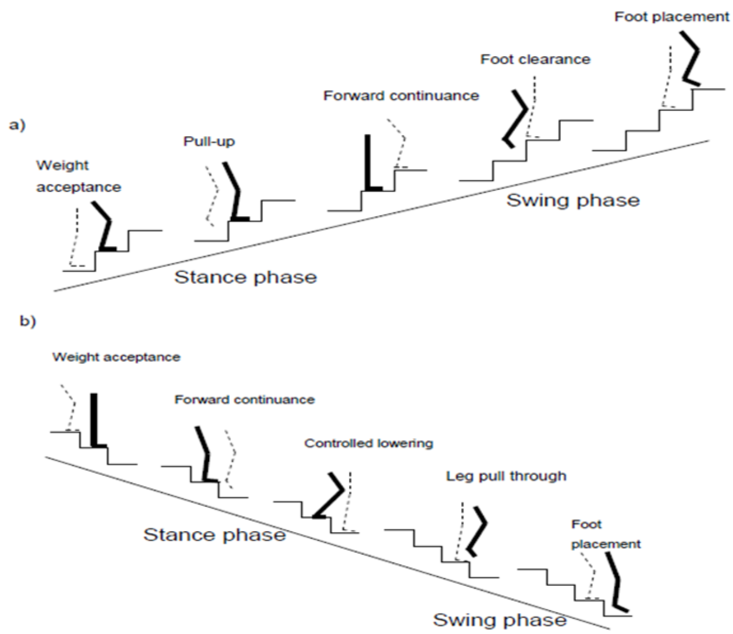

Figure 2.1 A schematic of the gait cycle of step over step stair ascent (a) and stair descent (b) [13].

The stance phase of descent is divided into three sub-phases: weight acceptance; forward continuance (the start of single limb support and forward body movement); and controlled lowering (the body's mass is lowered onto the support limb) [12, 13]. The swing phase has two sub-phases: leg-pull through (the swing limb is pulled forward); and foot placement [12, 13] (Figure 2.1b).

The Kinematics of Stair Negotiation

As with walking, the greatest range of motion (ROM) required at lower limb joints occurs in the plane of progression (sagittal) [12, 13, 54, 55]. The ROM associated with stair negotiation, however, is greater than that associated with level walking and particularly so during ascent compared to descent. In general, upwards of 10° to 20° of additional mobility is needed at each lower limb joint in healthy adults (Table 2.1) when compared to walking [13, 55, 56].

Table 2.1 Comparison of maximum sagittal plane angular displacements (in degrees) required for walking and stair negotiation* over the entire cycle (stance and swing phases) [13]

Joint	Walking[57]		Stair ascent		Stair descent	
	Flexion	Extension	Flexion	Extension	Flexion	Extension
Hip	20	20	60-65[58, 59]	5[58]	30-40[12, 59]	5[12]
Knee	60	0	94[58, 59]	10[58]	90[12, 59]	0[12]
Ankle	7	25	11-29[58, 59]	9-31[58, 59]	21-35[12,	15-40[12]

*Standard stair dimensions (rise/run) apply to stair ascent and descent values

Research has reported that healthy, older adults show reduced sagittal plane movement at the ankle and knee compared to young adults when descending stairs, which is accompanied by increased frontal plane motion at the hip and pelvis [60]. When climbing stairs, age-related reductions in ankle dorsiflexion could result in a trip if the toes fail to clear the step [61], althoughindividuals may take advantage of compensatory movements such as hip flexion or abduction to successfully negotiate the step.

Staircase slope (rise over run) proves to be an important characteristic affecting temporal [54] and kinematic gait parameters and hip/knee extensor activity [62]. Modest slope conditions can induce noticeable changes in the lower-limb locomotor patterns and represent a substantial physical obstacle to populations with restricted mobility [63]. There is significant dependency of most gait parameters during stair negotiation. However, the intensity of the dependency is different. For example, there is only a small to moderate influence of the slope on joint angle patterns, gait phase parameters and joint moment patterns [55]. Angular ranges of all lower limb joints increase with increasing slope angles [55]. This is consistent with the need for a higher elevation of the foot at increased step heights as shown previously [54].

The largest differences on stair inclination are observed in the joint power patterns. Maximum joint powers in the hip and ankle change with inclination up to ~67% [55]. This change can be related to the varying amount of potential energy that has to be produced (during ascent) or absorbed (during descent) by the muscles in order to surmount stairs with different slopes.

The increased toe clearance and cycle durations at minimum and maximum slopes can be interpreted as signs that stair climbing at extreme inclinations is a particular challenge at extreme inclinations for the human motor control system compared to climbing normal stairs. Typically, the rise (height of one stair in a staircase) is between 12.7 cm and 19.7 cm with the ideal rise being ~17.8 cm, while the run (horizontal distance of one stair) is ~30.5 cm [64].

Muscular Demands of Stair Negotiation

Stair ascent requires considerable positive power to lift the body against gravity in addition to generating movements which advance the body in a forward progression [58]. Concentric contractions of the lower limb muscles predominate during stair ascent to lift the body's mass against gravity [12, 13, 59, 65]. In stair ascent as with walking, the plantar flexors are main contributors to the work produced to complete the task [12, 13, 59], including the maintenance of upright support (the net support moment) [13]. During stair ascent at the ankle, a plantar flexion moment predominates reaching a maximum magnitude during late stance as the body is raised upward and forward to complete the rise onto the step [13, 59]. Further, older adults showed reduced ankle plantar flexor moments compared to young adults (1.24 Nm/BWkg and 1.48 Nm/BWkg) [39]. Similarly, older adults demonstrated lower magnitude knee extensor moments than younger adults (0.89 Nm/kg and 1.19 Nm/kg) [39]. This is notable considering that there was no difference in cadence between older and young adults in this investigation. This suggests that the hip extensors likely contribute to performing the work of this task to a greater extent in older than younger adults.

Unlike stair ascent, the net muscle powers are predominately negative during stair descent, representing energy absorption to control the lowering of the body when descending from one step to the next. In healthy young adults, the ankle plantar flexors absorb a significant amount of energy in addition to the moderate absorption occurring at the knee during weight acceptance. During these eccentric contractions through stair descent, older adults generated lower peak ankle plantar flexor moments than young adults (1.03 Nm/kg versus 1.32 Nm/kg); a strategy enabling older adults to operate at a similar relative proportion of their maximum capacity of the ankle compared to young adults (about 75%) [66]. Through late stance or controlled lowering, the knee extensors provide significant energy absorption [13], reflecting the gravity assisted nature of stair descent. In contrast, older adults generated knee extensor moments of a similar magnitude as those of young adults (0.83 Nm/kg versus 0.91 Nm/kg) such that they performed at a higher proportion of their maximal capacity (42%) than their younger counterparts (30%) [66]. Only the hip flexors generate some energy in late stance during stair descent [13, 55]. Thus, older adults redistribute the relative extensor moment outputs at the hip, knee and ankle joints as a strategy to keep the costs within their physical capabilities and safe limits during stair descent [66].

Demands during Stair Negotiation

Previous researchers found that successful stair descent required both accurate visualization of the stairs and proper kinesthetic feedback [67, 68]. The roles of the somatic and visual systems during stair descent are of special consequence to older adults, as both systems may be impaired in this population [69]. Nevertheless, a smooth stair descent requires good coordination of the agonist and antagonist muscles that regulate joint stiffness during the descent. Since both muscle strength and proprioceptive feedback are diminished with aging, the older individual may use a different control strategy to regulate leg stiffness than the younger individual.

It is noteworthy that the demands on the hip muscles can be quite variable in response to even small alterations in trunk position [13, 59]. For example, anterior-posterior (A/P) shifts in trunk position during ascent or descent require compensating hip extensor or flexor moments to maintain dynamic stability of the combined mass of the head, arms and trunk over the base of support [70]; a challenging prospect in the presence of instability and muscle weakness. Further, older adults presumably look down more, which shifts their weight, truck etc.

Movement Control

The success by which tasks are completed with precision depends on our capacity to control movement. Even the simplest movements in life demand the appropriate application of force, including the magnitude, direction, and timing of force, for both the agonist and antagonist muscles [1]. Injury to a joint affects the magnitude of maximal strength as well as the control of force produced by the muscles surrounding the joint [2, 3]. Adequate control of submaximal muscle force is especially important in ADLs that are normally executed at a fraction of the available maximal muscle strength in older adults [4]. Steadiness of force production and force control has been used previously to assess the quality of force production in the older adult population [5-7].

Muscle is a tension-producing tissue that is comprised of small contractile units referred to as sarcomeres which contain thick (myosin) and thin (actin) myofilaments (muscle filaments). There is an overlap of the myofilaments that allows for the formation of a cross-bridge bond (attachment). When this bond is formed a conformational change occurs within the myosin heads causing the sarcomere to shorten and produce tension. For voluntary muscles, all contractions (excluding reflexes) occur as a result of conscious effort originating in the brain [8]. The brain sends signals in the form of action potentials through the nervous system to motor neurons that innervate multiple muscle fibers. A single motor neuron and the innervated muscle fibers are termed a motor unit. A typical muscle is comprised of 100+ motor units [9].

A motor neuron receives information from both cognitive and reflexive sources. The spatial and temporal sum of these two components on the dendrites of the motor neuron will determine if an action potential is produced to generate a muscle contraction [10].

Skeletal muscles are able to produce varying levels of contractile force based on the number of motor units that are active and the amount of twitch summation that occurs within each motor unit. Twitch summation describes the addition of individual twitch contractions to increase the force of overall muscle contraction [12]. Twitch is the term used to describe the force generated from a single motor neuron action potential. Twitch summation occurs by increasing the frequency of muscle fiber activation. In this way the twitch force produced by an activation potential that occurs before the force from the previous twitch drops to zero is additive. This twitch summation, however, is not infinite since the peak tension that can be maintained by the myosin-actin bonds is finite. Once peak tension has been achieved the force levels off, termed tetanus.

In skeletal muscle the method of excitation contraction coupling relies on the ryanodine receptor being activated by a domain spanning the space between the T-tubules and the sarcoplasmic reticulum to produce the calcium transient responsible for allowing contraction [13]. The general scheme is that an action potential arrives to depolarize the cell membrane. By mechanisms specific to the muscle type, this depolarization results in an increase in cytosolic calcium [13]. This increase in calcium activates calcium-sensitive contractile proteins that then use ATP to cause cell shortening through interaction of the myosin and actin cross-bridges [14].

While muscles always try to shorten when active, this may not always be the case. The level of force produced in the muscle relative to the external resistance will determine if the muscle actually lengthens, shortens, or stays the same length when activated [15]. In concentric contractions, the force generated is sufficient to overcome the resistance, and the muscle shortens as it contracts. In eccentric contractions the force generated is insufficient to overcome the external load on the muscle and the muscle fibers lengthen as they contract [14]. Further, contractions could be isometric where muscle length does not change during contraction. The ability to produce force varies with mode of lengthening/shortening. Muscles can produce the greatest force eccentrically, followed by isometric condition, and finally the concentric condition [12].

While isometric contractions are used to hold joint positions, and concentric contractions are used to actively produce movements, eccentric contractions fundamentally oppose joint motion and act as a braking force. Eccentric contractions are also often present to protect the joint from damage when the overall motion is dominated by concentric contractions of other muscles. Rather than working to pull a joint in the direction of the muscle contraction, the muscle acts to decelerate the joint at the end of the movement or otherwise control the repositioning of a load. This can occur involuntarily (e.g., when attempting to move a weight too heavy for the muscle to lift) or voluntarily (e.g., when the muscle is "smoothing out" a movement). While muscles can produce more force eccentrically than concentrically, the magnitude of the muscle activation, as indicated by electromyogram, EMG during a maximum eccentric contraction is often much less than that recorded during a maximum concentric contraction due to the lower levels of voluntary activation by the nervous system [16].

Effects of Age on Force Control

Physical inactivity contributes to muscle wasting and weakness, which often puts elderly persons at high risk for serious life-threatening falls [13]. Falls, especially while negotiating stairs result from numerous factors [14]. However, they are often directly attributed to impairments of force control and strength) in the lower extremity musculature [15].

Sarcopenia

Loss of muscle mass (Sarcopenia) is a process that starts around age 30 and progresses throughout life [17]. In this process, the amount of muscle tissue and the number and size of muscle fibers gradually decrease. With the gradual loss of muscle mass there is an accompanying loss of muscle strength. The loss of muscle strength places increased stress on certain joints (such as the knees) because they have to operate at higher levels of their maximum strength, which may predispose a person to falling [18].

Muscle mass and strength decrease approximately 10% per decade after the age of 50 [19]. Such deficits profoundly impact quality of life, even for healthy older people. Recent research indicates that the observed loss of force production in older adults is primarily due to the result of muscle atrophy and alteration in the percentage of contractile tissue within muscle [21-24], rather than deficits in muscle activation (motor unit recruitment and firing rates) [25, 26]. There is a preferential loss of type II fibers (aka fast twitch fibers) in older adults, to include the motor units that controlled them [21]. With fewer motor units, it may be more difficult to control force level. With fewer fast twitch fibers, which tend to be used at higher levels of force output, it may preferentially make control of higher levels of force more difficult. Furthermore, since there is evidence that fast twitch fibers are preferentially used for eccentric contractions [32], the loss of fast twitch fibers may be partially responsible for the evidence that shows older adults lose eccentric control faster than concentric control [23].

Age-related changes in joints

As people age, their joints are affected by changes in cartilage and in connective tissue [27]. The cartilage inside a joint becomes thinner, and components of the cartilage (the proteoglycans) become altered, which may make the joint less resilient and more susceptible to damage [28]. Thus, in some people, the surfaces of the joint do not slide as well over each other as they used to. This process may lead to osteoarthritis [24]. Additionally, joints become stiffer because the connective tissue within ligaments and tendons becomes more rigid and brittle [22]. This change affects the range of motion of joints [28]. Ligaments also tend to shorten and lose flexibility, making joints feel stiff with reduced ranges of motion.

Reduced joint proprioceptors will reduce a person's knowledge of body position (kinesthesia) and reduced peripheral proprioceptors, such as in the skin, will reduce knowledge of the interaction with the environment. Alone or in combination these loses will likely reduce force control [25]. Pain will alter what muscles are used, playing a role in force control [25]. The lack of joint proprioceptors and pain receptors causing a lack of mobility due to the stiffness of the joints may decrease the ability to control force, thus increasing the susceptibility of falls. This is particularly during stair negotiation as this demanding task requires a greater ROM compared to level walking [29].

Decreased Bone Mineral Density

As we age, the density of bones begins to diminish in men and women. This loss of bone density accelerates in women after menopause [29]. As a result, bones become more fragile and are more likely to break [30]. Further, bones become more porous, less resistant to stress, and more prone to fractures. This can be caused by hormonal changes, calcium and vitamin D deficiency and a lack of physical activity, all of which are associated with aging [10]. With aging, the body absorbs back calcium and other minerals from the bones which makes the bones weaker and leads to osteopenia and osteoporosis [31]. This reduced strength of the bones makes falling more dangerous as bones are more likely to break. Older adults may consciously and/or subconsciously alter movements to reduce the risk for falling. This may produce less efficient movement patterns that are harder to control [32].

Aging Effect on the Nervous System

Aging causes normal changes in the nervous system that can affect physical and mental abilities. When a nerve cell in the central nervous system dies, it is usually not replaced [32]. As cells die normally with aging, the brain weight gradually decreases [33]. These changes can result in a gradual loss of cognitive and motor function. These changes result in slower thinking, memory, and lack of physical activities. Nerve impulses have been shown to lose speed with age, leading to slower reflexes and responses. Further, as previously discussed, proprioceptors tend to die off with aging and are less sensitive, making proper force control more difficult. Thus, reaction time is slower because of the changes in the central nervous system and peripheral nervous system, causing the time lapse between the brain receiving signals and the person response to the signal to increase with aging [33]. The older individual is thus not as readily able to control force and maintain balance following a situational stress. Thus, these changes in the central nervous system could play a contributing role in the maintenance of balance and control of force during stair negotiation, leading to an increased susceptibility to falls in older adults.

Muscle Co-activation

Previous findings suggest that the level of co-activation of the flexor and extensor muscles during knee extension is higher in older adults compared to younger adults [44]. Co-activation of both agonists and antagonists may protect and stabilize the joint at the end-range motion [44], to ensure homogenous distribution of compression forces over the articular surfaces of the joint [8] and to increase joint stiffness thereby providing protection against external impact forces as well as enhancing the stiffness of the entire limb during forceful contractions [44]. Further, it has been shown that antagonist muscle activation during eccentric and dynamic knee extension is greater in men over the age of 65 compared to younger adult men [45]. This could reflect a loss of control and lower coordination or skill [12] and may be considered as an additional factor to explain the reduced level of net torque production across a joint and control in older adults. It has been shown that antagonist co-activation decreases as a result of resistance training in older [46] and younger adults [12].

Strength and Power during Stair Negotiation

Both strength and power are a contributor for successful completion of stair negotiation, especially in the ability to react quickly to a perturbation and prevent a fall. The ability to ascend and descend stairs successfully requires greater strength in the lower extremity than is needed for many other activities of daily living [34]. It is believed that a given individual may have adequate strength for level walking but not have sufficient strength to walk up and down stairs [35]. Joint moments calculated during motion using inverse dynamics can be considered as instantaneous estimates of the required strength in that they represent the product of instantaneous equivalent muscle force multiplied by the equivalent lever arm at a joint [36]. Power is the product of force and velocity (or torque and angular velocity in angular situations). Reductions in strength therefore contribute to losses in power. Considering that older adults preferentially lose fast twitch fibers, they lose power faster than they lose strength. The ability to generate power is important for stair negotiation stemming from the need to generate high levels of force quickly when contact is made with the ground as well as the need to move quickly if balance is lost to prevent a fall [48, 50].

Basssey and colleagues [38] examined the contribution of muscle power to functional tasks and found that leg extensor power was predictive of stair climbing performance in older adults. Another study [39] has shown that peak muscle power of the leg extensors and ankle plantar flexors are even more predictive of functional task performance than muscle strength. When tests across a range of external resistances (40%-90% of the 1RM), peak power typically occurs at approximately 70% of the 1RM. In addition, the independent effect of the velocity component of muscle power (at 40% of the 1RM) has been shown to demonstrate stronger associations with functional performance in older adults than maximal strength [40].

Eccentric Control

While older adults have impaired force control compared to younger adults, in general, previous studies have shown that older adults have a particularly difficult time controlling lengthening (eccentric) contractions during slow movement of inertial loads with the knee extensors [1] and during slow torque-matching contractions [7].

As previously discussed, past studies have shown that there is a preferential loss of type II fibers (aka fast twitch fibers) in older adults [21, 24, 27]. With fewer fast twitch fibers there is also a reduced number of motor units that controlled them [21]. With fewer motor units, it may be more difficult to control force level. With fewer fast twitch fibers, which tend to be used at higher levels of force output, it may preferentially make control of higher levels of force more difficult. Furthermore, since there is evidence that fast twitch fibers are preferentially used for eccentric contractions [32], the loss of fast twitch fibers may be partially responsible for the evidence that shows older adults lose eccentric control faster than concentric control [23]. Further, possible reasons for the loss of eccentric control include other age-related effects including muscle atrophy [17], increased muscle co-activation [21] and a reduced number of nervous cells [25]. These factors combined may contribute to the difficulty in controlling eccentric lengthening.

It has been presented that muscle strength during isometric contractions declines after the age of 60 years old at a rate of 10-15% per decade [42]. When this same maximal force capacity is assessed in a non-isometric contraction, there is less of a decline with age in force for eccentric contractions compared with concentric contractions [9, 43].

Eccentric Training

High-force production in appropriate doses is the stimulus for increasing muscle size and strength. Therefore, the elevated forces produced during eccentric contractions could be the most powerful stimulus to promote muscle growth and strength [51]. A further benefit of eccentric contractions is that the energy required (i.e., oxygen consumed) to produce negative work is relative to the equivalent magnitude (i.e., same force production) of positive work, i.e., when a muscle shortens, displacing an external load [51]. This low energy requirement results in a perception of much "less effort" to those participating in this exercise. Therefore, the "high force, low cost" abilities of eccentric contractions are thought to be ideally suited to the older adult populations engaging in resistance exercise [48].

However, eccentric training is not advised for all individuals, especially in certain untrained older adults [52]. While all types of muscle contractions can cause delayed onset muscle soreness (DOMS), it is especially noticed during eccentric exercises. DOMS is typically characterized as the muscle soreness and swelling that becomes evident 8-10 hours after exercise. There are several theories explaining the multifactor causes of DOMS. One hypothesis is the connective tissue theory that emphasizes the disruption of the non-contractile elements (i.e., connective tissue) in the sarcomere (such as the sarcoplasmic reticulum) and the connective tissues surrounding muscle proteins (i.e., sarcolemma) [30]. A cellular theory of DOMS focuses on the irreversible strain placed upon the sarcomeres during an eccentric contraction, resulting in disruption of components of the sarcomere. Still, most recently, a newer theory spotlights that an additional contribution to DOMS is with the excitation-contraction (E-C) coupling mechanism of the myosin cross-bridges attaching to actin proteins [52]. Previous research has hypothesized that the release of calcium ions (from the sarcoplasmic reticulum), which initiates the power stroke movement (i.e. the sliding of actin over myosin proteins), can be 'stretched' significantly with eccentric contractions (as compared to concentric actions) [16]. This E-C coupling elongation disruption, followed by substantial calcium ion release, results in a disruption of the voltage regulating sensors in the sarcomeres (which regulate neural input in the muscle), which also contributes to DOMS occurring from the eccentric exercise. It should be stated that older adults should not overdo eccentric exercises since they may not be able to recover as quickly [53]. A recent study has suggested that older adults may have reduced abilities to rebuild proteins after eccentric exercise [53]. Thus, older untrained adults should meet with a physician before beginning eccentric training and introduce it into their training programs judiciously.

The use of eccentric exercise in rehabilitation has increasingly gained attention in the literature as a specific training modality. Eccentric exercise has been primarily described in the rehabilitation literature for the management of tendinopathies [43]. However, evidence is mounting to support its use in the treatment of muscle strains, with most of the rehabilitation literature relating to the use of eccentric training in rehabilitation after lower limb injuries [8, 54-56]. Current research indicates that eccentric exercise is an effective form of treatment for lower extremity tendinoses, but little evidence suggests that it is superior to other forms of therapeutic exercises, such as concentric exercise or stretching [57]. Eccentric exercise may provide better outcomes than some treatments, such as non-thermal ultrasound, friction massage and splinting, and may be most effective during a respite from activity-related loading [58].

Eccentric training is particularly effective for rehabilitating certain muscles and tendons. During eccentric contractions there is an increase in the stiffness of the titin protein, whose functions as a molecular spring which is responsible for the passive elasticity of muscle. Titin adds a passive force enhancement to the muscle’s force production while being lengthened under load [9]. It has been speculated that other, not fully elucidated, metabolic force enhancement changes in the sarcomere are also occurring during eccentric muscle actions [59]. Therefore, if an older adult is able to include eccentric exercise in their training, it may have many beneficial effects.

Ground Reaction Forces during Stair Negotiation

The reaction force exerted by the ground against the person is typically called the Ground Reaction Force (GRF). The GRF combines both gravity's effect on the body and the effects of the body's movement and acceleration (change in velocity) in three planes of reference: vertical (GRFv), anterior-posterior (GRFap) and medial-lateral (GRFml). Therefore, the GRF is highly reflective of the muscular forces acting within the body.

Vertical Ground Reaction Force (GRFv)

GRFv both support the body weight and play an important role in the exchange of potential and kinetic energy [114]. Because of the heavy dependence of the GRF on bodyweight (BW), they are typically normalized to this for comparative purposes. Parameterization of distinct points on the GRFv curve has been pursued by a number of authors examining the biomechanical attributes of stair negotiation [12, 13, 114-116]. During stair negotiation the GRFv has a characteristic biphasic "M" shaped curve during both stair ascent and descent [70] (Figure 2.2). Similarly shaped to the GRFv of walking, it is typically split in two halves, with a maximum in each. These two maximum values are often defined as Fz2 and Fz4 (Figure 2.2). Fz2 is present in the phase of weight acceptance after touchdown, or loading, Fz4 during push-off or unloading. The first maximum, often denoted as impact peak, or Fz1, is typically not detected in stair negotiation at submaximal speeds [62]. Research on Fz1 shows that its magnitude depends mainly on the position of the foot during touchdown (heel-strike versus forefoot touchdown) and on the material properties of the shoe sole [117], aspects that are not considered to be important for the present study. During stair negotiation the biphasic "M" shaped GRFv curve is slightly different between stair ascent and descent as the second peak (Fz4) becoming larger than the first (Fz2) during stair ascent and the first peak becoming larger than the second during stair descent [16, 55, 62]. Stacoff *et al.* (2005) found that during stair decent the curves showed large variations with or without a second maximum [62].

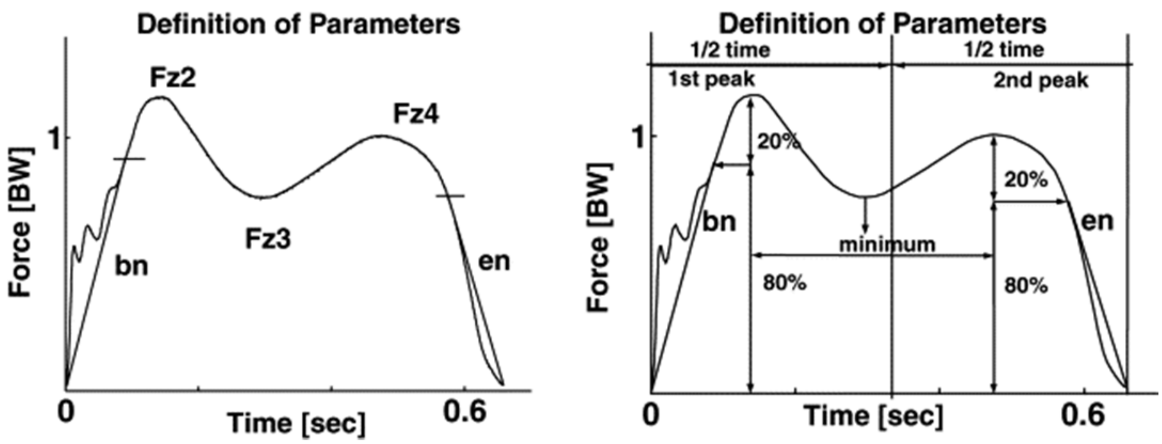

Figure 2.2 Detection and definition of selected parameters of the vertical GRF measured during stride to stride stair negotiation [62].

Between Fz2 and Fz4, the minimum (Fz3) is detected, representing unloading during mid-stance. By analyzing the GRFv signal relative to these events it is possible to calculate the mean value of the entire stance phase, the mean force of the mid phase between Fz2 and Fz4, the mean force during the loading phase from onset of force to Fz2, and the mean force during the unloading phase from Fz4 to foot take-off. Time duration of each phase is also a potential consideration.

In addition, the loading rate at touchdown (Loading Slope) and the unloading rate at takeoff (Unloading Slope) may be calculated [118]; note that Unloading Slope carries a minus sign (-) because it is a negative slope. These parameters are thought to describe the intensity with which the force develops at touchdown or ceases at take-off [117]. Reduced Loading and Unloading Slopes indicate a lower intensity in the weight transfer [117]. Stussi and Debrunner (1995) tested several definitions for Loading and Unloading Slope and found that the slope which represented the rate of Loading and Unloading best was that as shown in Figure 2.2, which uses the 80% value of Fz2 and Fz4, respectively [62]. This definition has the advantage that neither Fz1 nor very slow slopes (which may occur during very slow walking and/or in patients) would produce values for Loading and Unloading slopes which would lead to an overestimation or underestimation.

For stair ascent, the maximum values of GRFv were reported to be between 1.2 and 1.7BW. For stair descent, we find these values to be between 1.4 and 2.6 BW. In stair descent, there is a fast increase in GRFv reaching Fz2 at the start of single limb support (14% of stair descent cycle), then the GRFv decreases as in stair ascent until Fz3 (32% of stair descent cycle). Thereafter, the GRFv increases reaching Fz4 in the middle of the second double support phase (53% of stair descent cycle) [119].

Anterior-Posterior Ground Reaction Force (GRFap)

During gait, external forces act in three dimensions to support and accelerate the body. GRFap are involved in braking (deceleration) and propulsion (acceleration). The early portion of the GRFap profile during stair ascent is dissimilar to the stair descent profile at the same portion of the stance phase (Figure 2.3). During stair ascent the curve has two small peaks within the first 15% of the stance phase, which is absent during stair descent (Figure 2.3). This has the effect of lengthening the "region of risk" in the early support phase, during which a fall may occur [15]. The GRFap may be of particular interest relative to control of force as it is a parameter that switches direction of application within the stance phase.

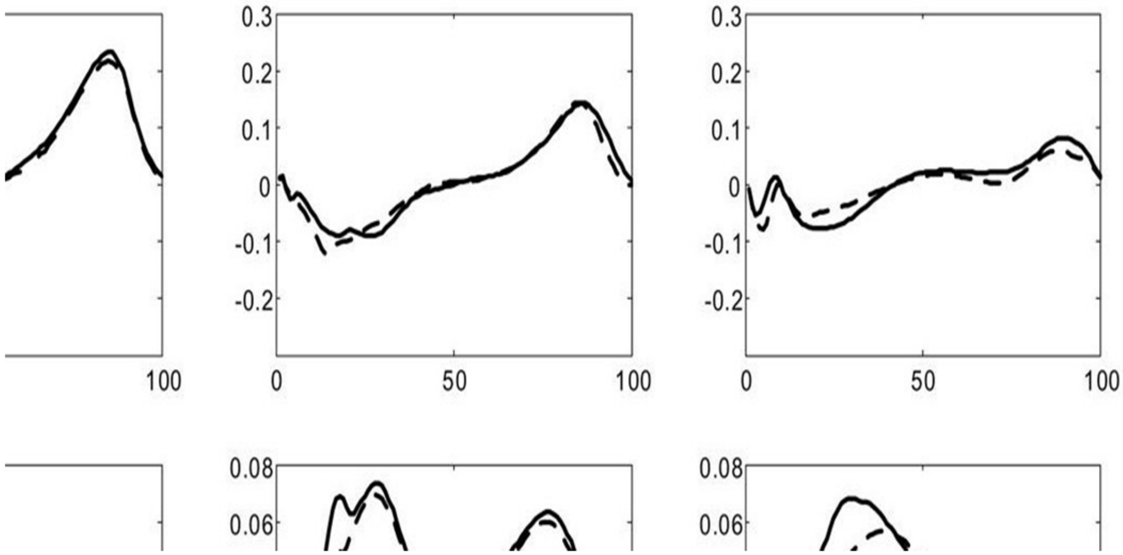

Figure 2.3 GRFap profiles from stair descent (left) and stair ascent (right) [15]. Note, a positive GRFap propulsive in nature while a negative GRFap is braking.

Medial-Lateral Ground Reaction Force (GRFml)

The GRFml profile is also biphasic with an "M" shape with two medially directed peaks of similar magnitude [120] (Figure 2.4). Excessive motion in the frontal plane can cause significant shifts in the location of the body's center of mass relative to the small base of support. Frontal plane motion will directly influence the GRFml. Medial-lateral control requires adequate muscle strength to maintain stability, which are known to decline with increasing age [70] and likely contribute to a reduction in the ability to safely ascend and descend stairs [60].

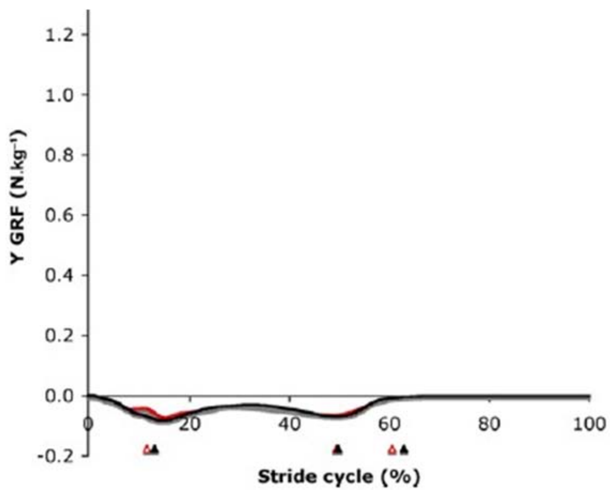

Figure 2.4 GRFml profiles of young and older adults during stair descent [120]. Note, a negative GRFml is acting medially on the foot.

Effects of Age on the Ground Reaction Force (GRF)

Larsen *et al.* (2008) [1] found that during freely ascending velocity during stair negotiation there was a significantly greater GRFv observed for young subjects compared to older during Fz2 peak (12.4%), as well as the average over the entire stance ($Fmean_{stance}$) (5.1%) and the average during loading ($Fmean_{load}$) (9.8%) (Figure 2.5) [1]. Contrarily, reduced Fz3 (13.7%) was demonstrated for the young subjects during stair ascent. Significantly lower values were observed in the older subjects for Loading Slope (72.3%) and Unloading Slope (43.0%) [1]. During descending freely chosen velocity Fz3 (5.5%) and the average force during unloading ($Fmean_{unload}$) (10.1%) were significantly higher for older subjects compared to the young (Figure 2.6) [1]. However, speed was not controlled and the young adults walked significantly faster during ascent and descent compared to the older subjects.

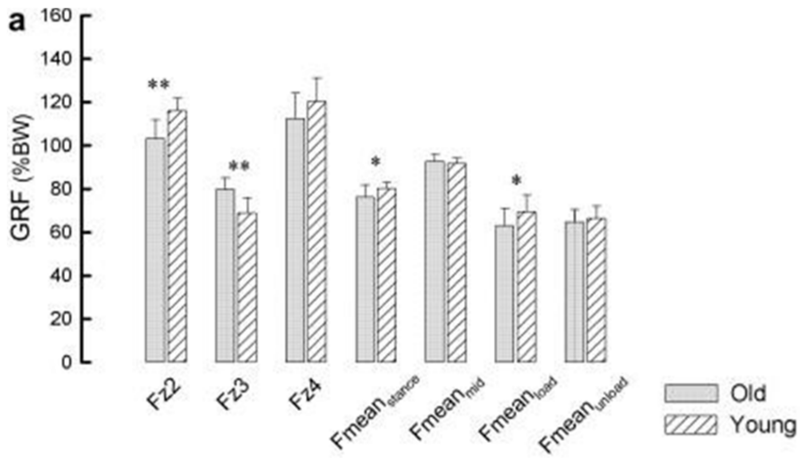

Figure 2.5 GRF v profiles ascending at freely chosen velocity GRFv [1].

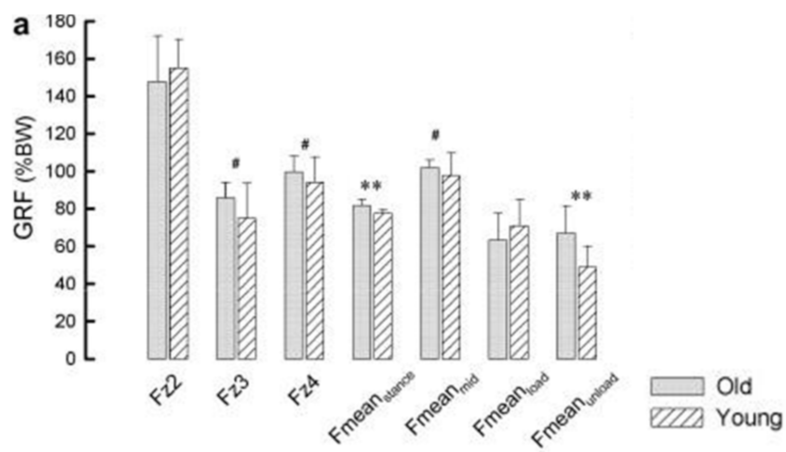

Figure 2.6 GRFv profiles descending at freely chosen velocity [1]

When speed is controlled for during stair ascent there are fewer differences between young and older adults. Reduced Loading Slope (25.3%) and Unloading Slope (17.5%) have been reported for older adults [1, 15]. In addition, $Fmean_{stance}$ has been found to be different between old and young subjects, showing a 4.2% higher mean average for younger adults (Figure 2.7) [1].

During descent at controlled velocity across subjects there is a significantly greater Fz3 peak (5.5%) and $Fmean_{unload}$ (10.1%) observed for older subjects compared with young subjects (Figure 2.8) [1]. Further, Hamel *et al* (2005) found a significantly greater Fz4 peak which is present during unloading in the old compared to young adults [15]. The age effects can be interpreted as a more vigorous push off and a more cautious use of the available friction at foot strike and at push off in older adults. This generally more conservative profile is similar to that found in other contexts (such as step clearance) in which older adults adopt more cautious strategies [27]. However, Loading Slope was shown to be significantly greater for the older adults [1, 15], which may reflect a lack of control during loading when compared with the young, or an increase in joint stiffness.

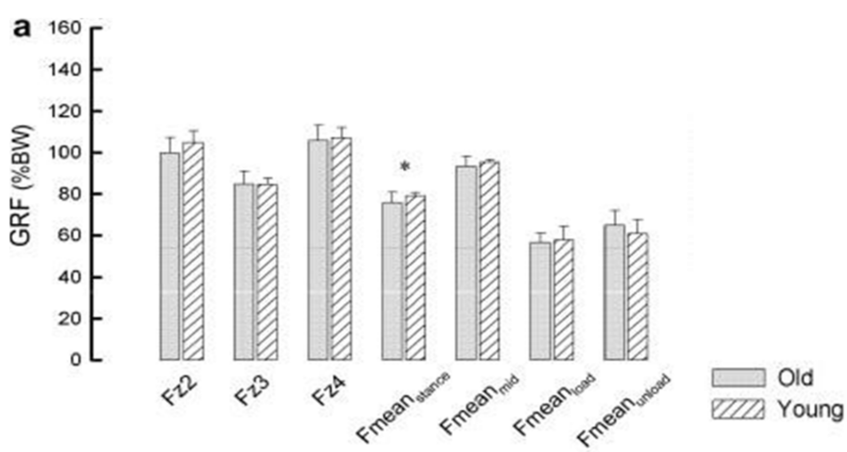

Figure 2.7 GRFv ascending at standardized velocity (Cycle frequency = 35 min^{-1}). [1]

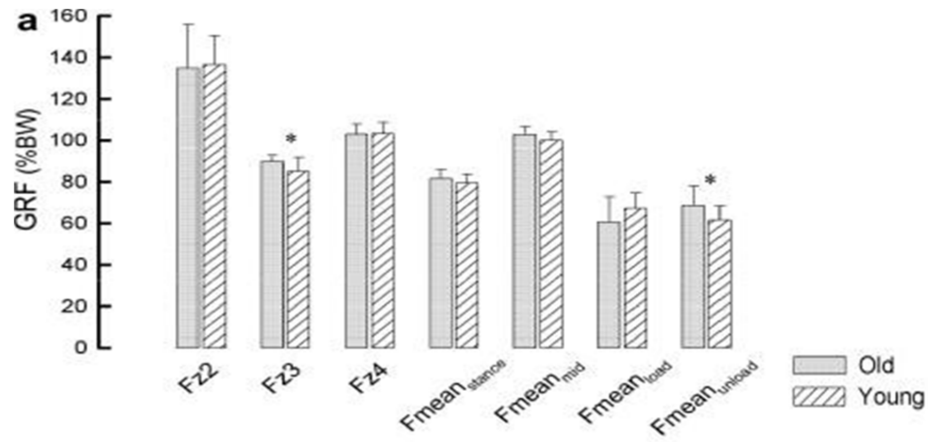

Figure 2.8 GRFv descending at standardized velocity (Cycle frequency = 35 min^{-1}). [1]

During stair negotiation at a controlled speed GRFap are similar in profile between the young and older adults and show two peaks, a posterior directed (braking) peak and an anteriorly directed propulsive peak [120] (Figure 2.3). However, during stair descent older adults exhibit higher peak posterior GRFap at loading compared to younger adults [15]. During stair ascent older adults exhibit significantly lower anteriorly directed GRFap during Unloading when compared to younger adults [15].

During stair descent the GRFml has been shown to be similar between the older and younger adults, showing a "M" shaped peak that is directed medially throughout the stance (Figure 2.4), with no significant differences between the two groups [120].

Thus, previous studies have observed more age-dependent differences between young and older individuals during stair descent compared to stair ascent at a controlled speed. This suggests that the joint moment requirement is probably higher in older adults highlighting the demanding physical nature of this task [17]. This may cause older adults to have less dynamic balance control leading to a more unpredictable and potentially dangerous unaided descent.

Free Vertical Moment during Stair Negotiation

Besides GRF, another output parameter from force platforms is the Free Vertical Moment (FVM). The FVM is a force couple and, as such, has the same action on any vertical axis regardless of its position. However, the rotational effect of the horizontal forces (GRFap and GRFml) will depend on the position of the axis [119]. In layman's terms, the FVM is the friction torque under the foot related to twisting and turning.

Following movement initiation when stepping down, the lead-limb is moved forwards and downwards. This can be achieved through a combination of localized limb movements and by rotation of the body about its vertical-axis, which causes movement of the out-stretched limb. Because of the action-reaction principal, such body rotation will be reflected by changes in the FVM. During gait this moment will have a biphasic pattern with the FVM generated during support phase being opposite in direction to that produced in the movement execution phase [19].

During self-paced step-downs older adults show a reduced peak FVM and FVM-impulse compared to younger adults [20]. Older adults show the FVM acting to slow and halt the body's vertical-axis rotation away from the lead-limb side. As a result, older adults land with significantly increased vertical-axis pelvis angular displacement and velocity at instant of landing, even though movement duration times indicate they take longer to complete step-downs [20]. This perhaps provides further support for why stair descent by older adults being described as a "controlled fall" [121].

The significant age differences in FVMs during step down-execution suggest older adults are unable to exert the same vertical-axis control during single-support compared to young adults. This highlights that older individuals have to undergo greater vertical-axis body rotation, back towards the lead-limb side following ground contact. This reduced FVM in older adults may be a consequence of reduced strength and flexibility, and/or inferior sensorimotor control; or simply a consequence of using a different/more cautious stepping strategy [20]. However, this study was conducted by having subjects on top of one of three blocks placed on top of a force platform and having them step-down. Further, speed was self-selected and was not controlled. Therefore, differences between young and older adults in the FVM during stair descent and ascent are not completely clear.

Center of Pressure during Stair Negotiation

Center of Pressure (CoP) is another variable that can be calculated and analyzed from a force platform. CoP is defined as the point of application of the GRF under the foot. The location of the CoP is the outcome of the inertial forces of the body and the restoring equilibrium forces of the postural control system. CoP displacement is used to make inferences about neurological and biomechanical mechanisms of postural control.

CoP Displacement

During level walking and stepping over obstacles, older adults show a shorter A/P displacement of CoP (CoPap) compared to young adults [122]. The shorter CoPap distance in older adults has been interpreted as a more conservative strategy, enabling them to reduce the required moments about the lower limb.

Another study demonstrated that older adults had a significantly decreased M/L displacement of CoP (CoPml) during obstacle crossing as compared to young adults [123]. A greater displacement of CoPml has also been reported for healthy older adults during obstacle crossing, as compared to older adults with balance impairments [124]. Furthermore, previous studies have reported that the control of whole body CoPml through manipulation of CoP during walking is closely related to the maintenance of lateral balance, which is highly related to increase risk of lateral falls in older adults [124].

During stair descent, it has been shown that the CoPap and CoPml displacements are significantly less for older adults as compared to young adults [90]. The mean average CoPap displacement for young adults was 143% greater than the mean average CoPap displacement for the older adults [90]. These results are similar to observations made during level walking and while stepping over an obstacle on the ground [125]. This reduction in the magni interpreted as a more cautious strategy and less efficient, resulting from reduced inhibition of soleus and gastrocnemius activity and increased ankle joint stiffness that is related to aging [126].

Center of Pressure (CoP) Velocity

The velocity of CoP provides valuable information about how individuals modulate gait when negotiating stairs. A greater CoP velocity may present a challenge to the maintenance of balance while a slower speed may be suggestive of a more stable position. Data on the effect of aging on velocity of CoP during stair negotiation is limited. In a single study examining CoP velocity during stair descent, the mean average velocity was shown to be significantly lower in older adults than that of younger adults [90]. The slower speed of CoP may be potentially beneficial for maintaining the dynamic balance that is necessary for supporting an upright posture during stair descent. This may indicate that older adults display a more cautious strategy for optimizing postural stability during stair negotiation.

However, during this single study examining CoP during stair negotiation, subjects walked at a self-selected pace, which could attribute to the differences between the groups. Typically, during self-selected pace trials older adults walk at a slower pace than younger adults [127]. A study examining these parameters while having subjects walk at a constant pace across groups has not been conducted to this point. Further, these parameters have not been examined during stair ascent. Thus, it is inconclusive if this is true when the cadence is consistent across groups.

Center of Mass (CoM) - Center of Pressure (CoP)

Additionally, gait stability can be assessed using the motion of the whole body Center of Mass (CoM) and its relative position to the CoP of the supporting leg. In a stable, non- accelerating system, CoP is located directly below the CoM. In an accelerating system it will not always be aligned with the

CoM when equilibrium is disrupted. In contrast to level walking, a study reported that when ascending stairs, both young and older adults demonstrated a significantly greater CoM-CoP medial inclination angle compared to the floor-to stair transition phase [128]. Inclination angles are the angle formed by the interaction of the line connecting CoP and CoM with a vertical line through the CoP. However, a significant reduction in medial inclination is only detected in young adults when transferring from stair descent to level ground walking since stair descent is a kinematically constrained task, which may limit any CoPap differences between groups. Furthermore, a greater CoM-CoP separation has been reported during stair descent compared to stair ascent and level walking in younger adults [60]. While the CoM-CoP relationship is intriguing relative to balance and force control it requires more than just force platforms. The CoM location is computed from kinematics that are usually collected with cameras and reflective markers.

Variability of Parameters

Past studies have reported that the central nervous system takes advantage of motor abundance when performing whole-body tasks [129, 130]. It allows for the flexible control of limb movements to achieve the body stability. This may be important for on-line error corrections. Thus, movement variability can be interpreted as the outcome of a continuous control process dynamically interacting with the environment. This higher variability may represent an attempt to improve compensations in motor control caused by alterations in gait due to external and internal changes [131].

Reliable GRF, FVM and CoP parameters are a crucial factor in the meaningful qualification of a movement task. Repeated trial-to-trial measurements could be used to evaluate the response of therapeutic interventions such as physical therapy or surgery. These results will be beneficial for the identification of further predictive parameters and the evaluation of treatment interventions targeting the improvement of stair climbing performance.

Comparing GRF values of trial-to-trial variability, expressed in coefficient of variation (COV), during stair negotiation have observed greater variability during stair descent compared to ascent [132], however speed was not controlled for in this study. Loading and Unloading Slopes show considerably higher values between 10-15% for stair ascent, and 15-20% for stair descent [62]. In addition, it has been shown that time parameters (in which gait speed ins not controlled), variability during stair descent has been shown to be high [132].

Previous findings have been interpreted as an increased coordinative demand during stair descent [133]. Therefore, it seems that descending is a more challenging task and requires greater variability from person to person in movement strategies. The results on trial-to-trial variability, COV, has been suggested to have a normal range from 5 to 10% for the GRFv peaks Fz2 and Fz4, respectively [62, 133]. These parameters indicate changes in force and thus presented the least stable movement patterns. It stands to be noted that speed was not consistent across groups during these studies.

Variability of GRF parameters during stair negotiation at a controlled speed reveals that testing variability shows no systematical differences between young and older adults [1]. However consistently lower COV are observed for stair ascent (4-7%) compared with stair descent (6-8%) in the GRFv [1]. Furthermore, variability of the Loading Slope and Unloading Slope shows a similar pattern, but with considerably higher COV (9-13% for ascent and 15-16% for descent) [62]. Further, FVM, GRFap and GRFml variability during stair negotiation has yet to be examined.

CoP variability is an indicator of balance control in gait. During level gait, CoP_{ml} displacement appears to exhibit a greater COV value compared to CoP_{ap}, suggesting poor symmetry in the M/L direction. Often, gait parameters in the M/L direction are characterized by substantial asymmetries in CoPml profiles due to the high variability [134]. To date, no studies have examined CoP age-related trial-to-trial variability during stair negotiation.

Significance of Study

As discussed, stair negotiation is one of the most difficult tasks attributed to aging and is one of the leading causes of fall related injuries [7]. Past studies show us that stair negotiation has age related differences in GRF, the CoP and most likely the FVM between older and younger individuals even when speed is controlled.

Specific to GRF during stair ascent, older adults have lower Loading and Unloading GRFv compared to younger adults as evidence from the reduced Fz2 and Fz4 peaks [62]. During stair descent , older adults demonstrate a higher Fz2 peak and GRFv Loading Slope compared to younger adults [15]. By examining the slope of the loading and unloading phase we can determine the control of force and how it changes with aging. During stair ascent GRFap is similar between the young and older adults and show two peaks, a posterior directed (braking) peak and an anteriorly directed propulsive peak [120]. During stair descent older adults exhibit higher peak posterior directed GRFap during loading compared to younger adults [15]. During stair ascent older adults exhibit significantly lower anteriorly directed GRFap at push off when compared to younger adults [15]. The GRFml profiles between young and older adults have not been examined at a controlled speed.

Data on FVM reveal older individuals show a reduced peak FVM and FVM-impulse compared to younger adults when stepping down at a freely chosen comfortable speed [20]. As a result, older adults land with significantly increased vertical-axis angular displacement and velocity, contributing to the support of a “controlled fall” during stair descent. However, only a single study has only been conducted during stepping down and age-related differences are inconclusive.

Research on CoP during stair negotiation is limited. During stair descent a single study has shown that the A/P and M/L displacements of CoP, as well as the average velocity of CoP for these two directions are significantly reduced for older adults as compared to young adults [90]. However, these studies did not control for cadence which could explain the differences. Nor did they analyze differences during stair ascent. Therefore, further investigation into age related changes in the CoP are warranted during both stair ascent and descent at a consistent pace.

Studies examining trial-to-trial variability during stair negotiation with GRF forces and CoP is scares. It has been shown that there is more variability in GRF forces during descent compared to ascent [132]. A single study has analyzed GRF variability during stair negotiation in the elderly population, however speed was not controlled. Further, no previous findings can be find that examines CoP variability during stair negotiation.

The GRFs, FVM and CoP are of particular interest because of their relationship to muscle force both in terms of magnitude and control. Considering that force control is impaired to a greater degree in older adults during graded contractions eccentrically more so than concentrically, force control may be effected more during stair descent than ascent, contributing to the high rate of falls. Therefore, the goal of this study is to examine GRF, FVM and CoP differences during stair ascent and descent between young and older healthy adults at the same cadence.

CHAPTER 3

METHODS AND PROCEDURES

Subjects

Eleven healthy older adults (OA) between the ages of 65-95 years old and ten young healthy adults (YA) between the ages of 18-30 years old with equal sex ratios between groups were recruited and gave written informed consent to participate. All subjects were free of any condition or medication known to affect the measures and were no more than moderately active (exercise <30 min/day, 3 days/ week), with no regular strength training in the past year and no history of falls. Subjects were carefully screened for orthopedic conditions (arthritis, etc.) that could prevent pain-free stair descent or ascent. Height, weight, as well as right foot length and width were gathered from each subject. Project approval was gained from the Colorado State University Human Subjects Committee.

Staircase Configuration

All measurements were conducted on a custom-built four-step wooden staircase with an isolated Bertec 4060-10 (Columbus, OH) force-measuring platform embedded in the center of the second step. The riser height of the staircase was 18 cm, with a tread depth of 28 cm and a width of 100 cm. A handrail was independently mounted on both sides and present during all tests. A non-slip adhesive top was across each step to prevent subjects from slipping. The landing at the top of the 4th step was 100 cm by 100 cm and surrounded by handrail.

Stair Walking Protocol

All participants performed the stepping tasks with provided anti-slip socks in order to standardize the footwear and for security purposes when walking without shoes. Participants were asked to wear loose fitting clothing. Participants ascended and descended the stairs at a predetermined pace of 70 steps/min (coordinated by an audible tone metronome) in a step-over- step manner. Participants started at the bottom of the platform and performed one step on level ground leading with their right foot followed by step wise progression through the four step staircase such that the right foot always contacted the force platform. The descent consisted of one step on the landing leading with their right foot followed by step wise progression through the four step staircase, such that the right foot contacted the force platform. This ensured "steady–state" when crossing the force platform [13], with the force platform the second step when both ascending and descending the staircase. Handrails were present as a safety precaution and participants were asked not to use them unless necessary. In instances where participants did use the handrails, the trial was discarded and not used in the analysis. A minimum of one initial test trial for each subject was performed to condition the subject for the selected pace and familiarity of the staircase and equipment. The time between the five trials was determined by the subject and was typically fewer than 30 seconds. All subjects performed 5-6 trials in which ascent and descent data was collected for further analysis. Though not used in the analysis, an electrogoniometer (SG150, Biometrics, Ltd, Gwent, UK) was attached across the right knee with hypoallergenic adhesive tape.

Parameterization of Ground Reaction Forces

Force platform GRFs and moments were sampled at 1000 Hz through a motion capture system (Motus 9.0, Vicon, Centennial, CO). CoP and FVM were subsequently calculated in the software using the force and moment data. Raw signals were Butterworth low-pass filtered at 20 Hz (4^{th}-order, recursive). The start and end of stance was detected with a 5N threshold. The GRFv was analyzed similar to Stacoff et al. [62] and Larsen et al. [1]. In brief, forces were normalized to percent body weight (%BW) and the "M" shape curve was divided in two halves (Figure 2.3). The peak force was detected on both sides of the midline as Fz2 and Fz4, respectively. Fz2 was present in the loading phase after touchdown and Fz4 prior to the unloading phase. An initial contact peak (Fz1) prior to Fz2 was not extracted in the present study as it was not always present.

Between Fz2 and Fz4, a minimum GRF was detected, Fz3 [118]. Furthermore, the Loading Slope (from onset of force production to 80% of Fz2) and Unloading Slope (from 80% Fz4 to toe- off) were calculated. Additionally, the slope of the 1st and 2nd half of each of these slopes was also calculated in order to more fully explore the control of force. The time duration of each phase and the entire stance were also extracted. Further, we examined the GRFap maximum peak during propulsion and the minimum peak during breaking as well as the average during the stance. For GRFml the two maximums during propulsion and breaking in addition to the averages were examined.

Parameterization of Free Vertical Moment

From the data collected from the force platform, free vertical moment (FVM) was calculated as the moment about the vertical-axis of force-platform, resulting from shear forces around the CoP [135]. To enable comparisons across subjects, FVM data was normalized to a percentage of body weight by height [135]. FVM Max, Min, as well as total Stance Avg and Impulse values were analyzed.

Parameterization of Center of Pressure

CoP was calculated using the force and moment data from the force platform. CoP velocity was then calculated with finite differences using the CoP position one time step ahead and one behind the instantaneous location. Due to instability in CoP calculations at initial ground contact and toe-off, the first 40% of the loading and last 40% unloading were not included in the COP analysis.

A/P COP (CoP_{ap}) and M/L COP (CoP_{ml}) data were normalized to foot length (%FtL) and width (%FtW), respectively. CoP variables were assessed within the Loading, Mid-stance, and Unloading phase as well as over the entire stance phase in the A/P and M/L directions. Variables extracted included:

- CoP Max – Min (M-M)
- CoP End – Start (E-S)
- CoP Standard Deviation (SD)
- CoP Path Length (PL)
- CoP Maximum Velocity (Vmax)
- CoP Minimum Velocity (Vmin)
- CoP Average Velocity (Vavg)
- CoP Standard Deviation of Velocity (VSD)

Data Analysis

The values extracted from the first five usable steps in ascent and descent, respectively, were averaged to create a representative value for each subject. Additionally, the coefficient of variance (COV) of the five trials was calculated for assessment of step-to-step variability. The individual subject averages and COVs were pooled for assessment of differences between groups and ascent/descent with a 2 x 2 repeated measure ANOVA on trials within and between subjects. Post-hoc t-tests for equality of means between groups were performed if significant differences existed in the interaction of main effects. All statistics were performed in IBM SPSS Statistics 21 (Chicago, IL). Statistical significance was set at $p < 0.05$.

CHAPTER 4

RESULTS

Note: For brevity, statistical results are presented without mention of directionality. I.E., there is no indication whether a value was greater in ascent or descent or in the YA compared to the OA. Directionality is available by examination of the values in the tables. Also, the Discussion is written with explicit indication of the direction of the observed differences.

Subject Characteristics

Nineteen subjects (10 YA, 9 OA) completed the data collection. An additional two OA subjects passed the eligibility screening, but were unable to negotiate the stairs at the required pace without using the handrails. These two subjects were excluded from the study. Besides age ($p<0.001$), there were no differences in mass or anthropometric dimensions between the two groups ($p \geq 0.217$) (Table 4.1). All subjects reported their right leg as their preferred kicking leg except for one subject in the YA group. The groups were equally balanced for sex (YA: 5 men, 5 women; OA: 5 men, 4 women).

Table 4.1: Subject Characteristics

	YA		OA	
	Mean	SD	Mean	SD
Age (yrs)*	22.6	(1.6)	76.87	(5.3)
Mass (kg)	69.8	(11.2)	75.2	(11.4)
Foot Length (cm)	25.1	(1.8)	25.4	(1.3)
Foot Width (cm)	9.1	(0.6)	9.5	(0.7)
Height (cm)	171.1	(8.1)	167.4	(8.0)

*$p<0.05$ between YA and OA

Stance Time

A significant main effect for ascent/descent existed in all the time parameters ($p \leq 0.003$) with no significant main effects for age group ($p \geq 0.056$) (Table 4.2). However, significant interactions between groups and ascent/descent existed for Load time and Mid time ($p \leq 0.042$) with differences between the age groups in Load time during stair descent ($p=0.008$) and Mid time during stair ascent ($p=0.004$).

When examining subject variability, there were significant main effects in the average COV for ascent/descent in total Stance, Mid and Unload times ($p \leq 0.038$) and a main effect for age group in total Stance time ($p=0.026$) (Table 4.3). There were no significant interactions between groups and ascent/descent in time parameter variability ($p \geq 0.086$).

Vertical Ground Reaction Forces (GRFv)

GRFv profiles during ascent and descent showed an "M" shaped biphasic curve (Figure 4.1). A significant main effect for ascent/descent existed in the Fz2 and Fz4 peaks ($p < 0.001$) with no significant main effects for age group for any maximum or minimum GRFv ($p \geq 0.086$) (Table 4.2). However, significant interactions between groups and ascent/descent existed for Fz2 ($p \leq 0.001$) with differences between the age groups during both stair ascent and descent ($p \leq 0.023$). When examining subject variability, there were significant main effects in the average COV for ascent/descent in Fz2 and Fz3 peaks ($p \leq 0.042$). There were no significant interactions between groups and ascent/descent in maximum or minimum GRFv variability ($p \geq 0.103$) (Table 4.3).

A significant main effect for ascent/descent existed in all Loading and Unloading Slopes ($p < 0.001$) with significant main effects for age group for Loading Slope 1 and all of the Unloading Slopes ($p \leq 0.017$) (Table 4.2). However, significant interactions between groups and ascent/descent existed only for stair ascent in Loading Slope 2 and Unloading Slope 1 ($p \leq 0.001$). When examining subject variability, there were significant main effects for average COV for ascent/descent in Loading Slope 1 and 2 and

Unloading Slope 2 (p<0.010) and significant main effects for age groups in Loading Slope 1 and 2 and Unloading Slope 1 (p<0.031). However, there were significant interactions between ascent/descent and age group only in Loading Slope 1 in ascent and Loading Slope 2 in descent (p<0.040) (Table 4.3).

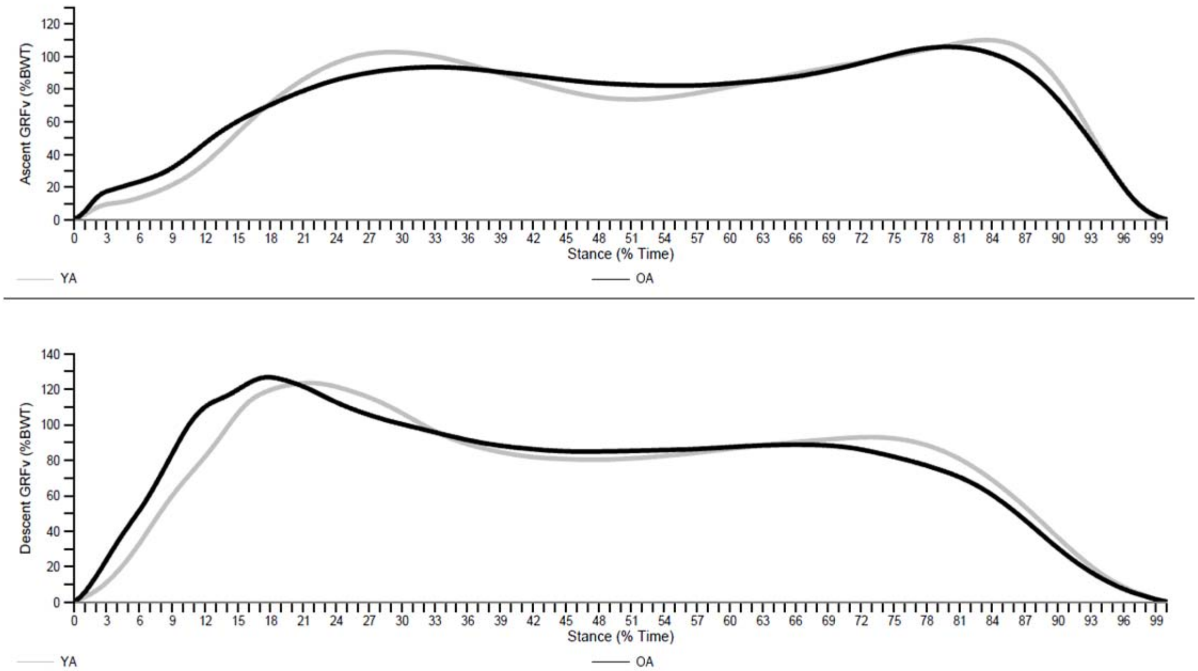

Figure 4.1 Group average Stair Ascent (top) and Descent (bottom) profiles for the GRFv.

Anterior/Posterior Ground Reaction Forces

GRFap profiles were highly biphasic with an initial posterior braking phase followed by anterior propulsion (Figure 4.2). A significant main effect for ascent/descent existed in all three GRFap variables (p<0.001) with no significant main effects for age groups (p≥0.121) (Table 4.2). However, a significant interaction existed between groups and ascent/descent for GRFap Min (p<0.031) though post-hoc tests for age group differences were not statistically significant (p≥0.092). When examining subject variability, there were no significant main effects or interactions related to GRFap (p≥0.061) (Table 4.3).

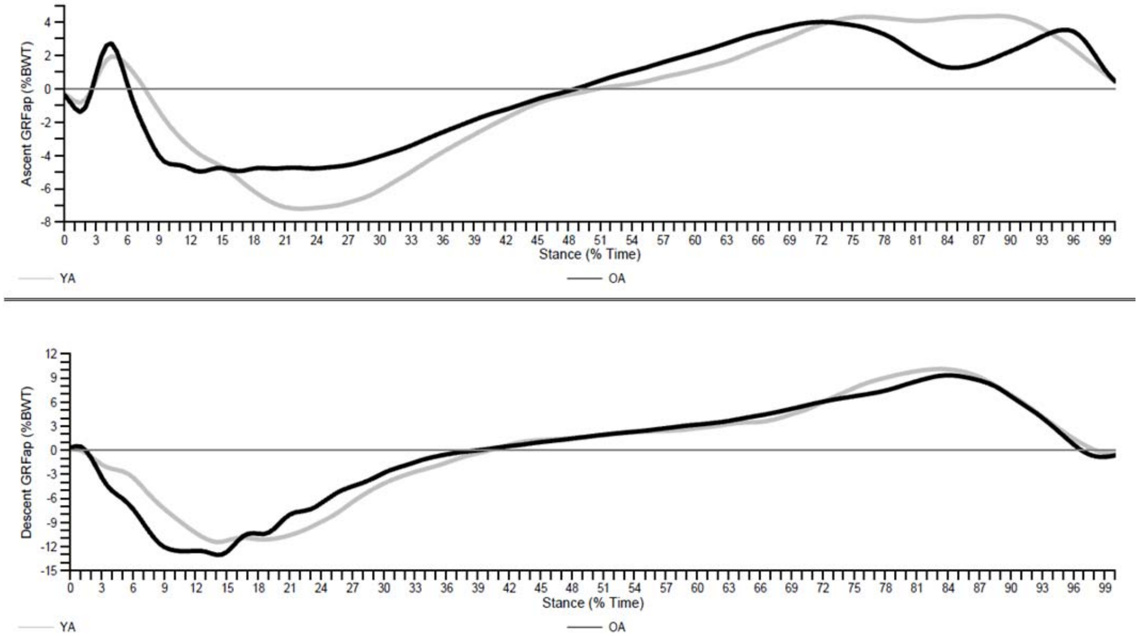

Figure 4.2 Group average profiles during Stair Ascent (top) and Descent (bottom) in the GRFap.

Medial/Lateral Ground Reaction Forces

GRFml profiles exhibited "M" shape biphasic profiles (Figure 4.3). A significant main effect for ascent/descent existed in all three GRFml variables ($p \leq 0.001$) with no significant main effects for age groups ($p \geq 0.11$) or any interaction ($p \geq 0.327$) (Table 4.2). When examining subject variability, a significant main effect existed for ascent/descent with GRFml Unloading Max ($p \leq 0.034$), but not for age groups ($p \geq 0.13$) or any interaction ($p \geq 0.094$) (Table 4.3).

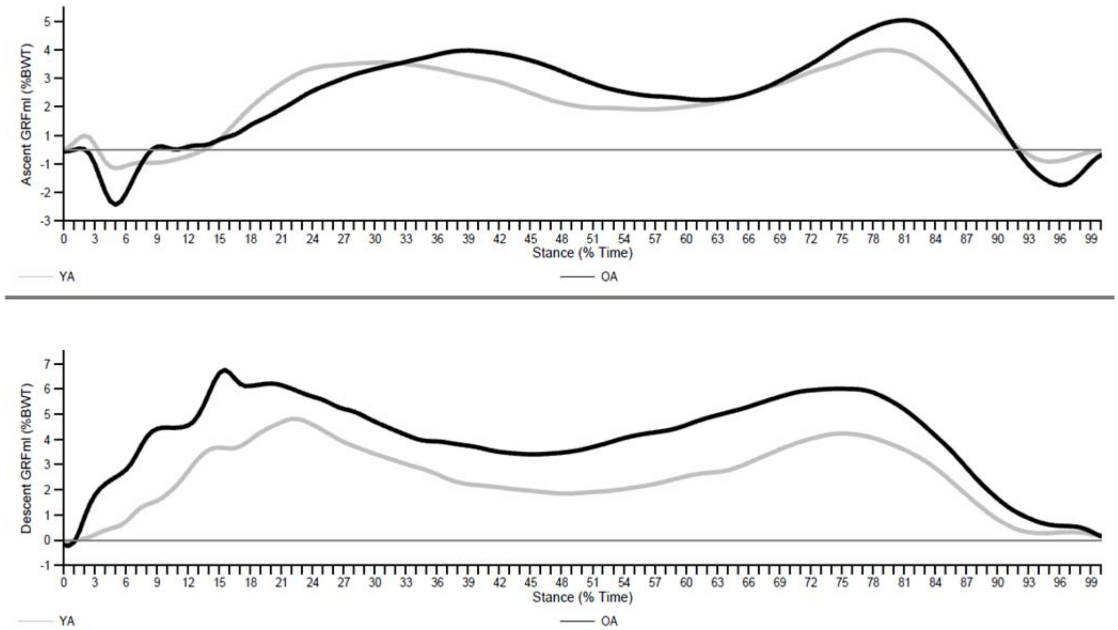

Figure 4.3 Group average profiles during Stair Ascent (top) and Descent (bottom) for the GRFml.

Free Vertical Moment

FVM during stair ascent was biphasic, switching mid stance, while the FVM during stair descent was of a singular negative phase (Figure 4.4). A significant main effect for ascent/descent existed in FVM Max, Min and Impulse ($p \leq 0.026$) with no significant main effects for groups ($p \geq 0.072$) or any interaction ($p \geq 0.088$) (Table 4.2). When examining subject variability, a significant main effect existed for ascent/descent in FVM impulse ($p \leq 0.007$) with no significant main effects for groups ($p \geq 0.225$) or any interaction ($p \geq 0.123$) (Table 4.3).

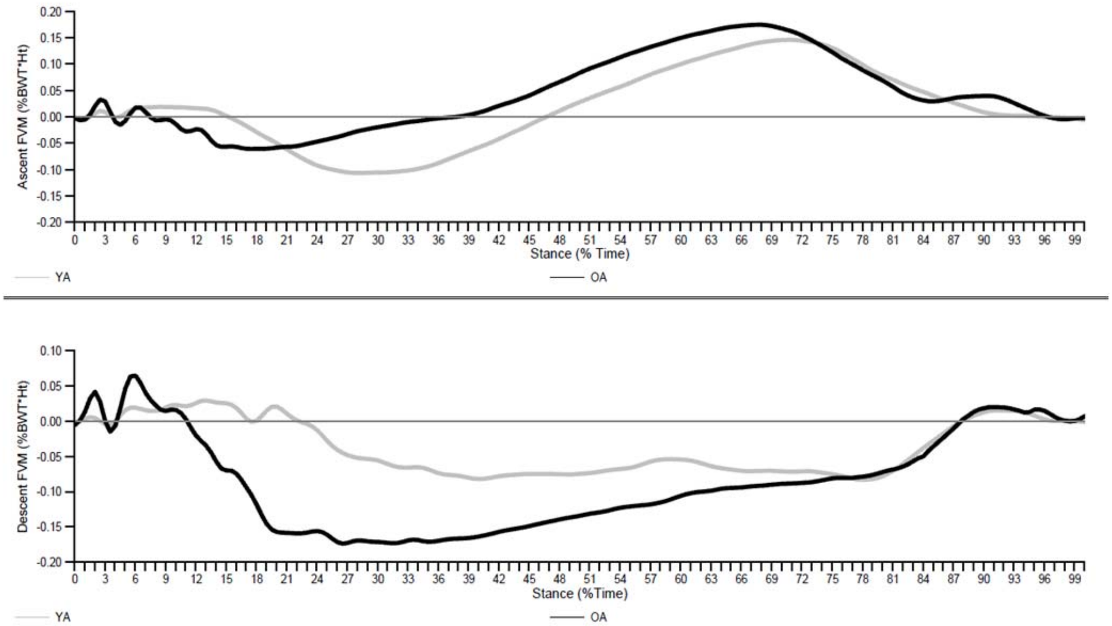

Figure 4.4 Group average profiles during Stair Ascent (top) and Descent (bottom) in the FVM.

Table 4.2: Ground Reaction Force & Free Vertical Moment Results

	Ascent				Descent				
	YA		OA		YA		OA		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Time									
Stance (s)	1.03	(0.08)	1.05	(0.08)	0.96	(0.09)	0.98	(0.12)	a
Load (s)	0.30	(0.04)	0.34	(0.06)	0.20	(0.05)	0.15	(0.03)	a,d
Mid (s)	0.56	(0.03)	0.48	(0.07)	0.45	(0.06)	0.46	(0.06)	a,c
Unload (s)	0.17	(0.02)	0.22	(0.05)	0.31	(0.08)	0.37	(0.15)	a
GRFv									
Fz2 (%BWT)	107	(9)	97	(5)	132	(12)	147	(14)	a,c,d
Fz3 (%BWT)	72	(9)	79	(7)	78	(6)	81	(8)	
Fz4 (%BWT)	112	(6)	108	(8)	98	(6)	95	(4)	a
Loading Slope (%BW/s)	424	(89)	389	(88)	794	(243)	1005	(228)	a
Loading 1 Slope (%BW/s)	318	(73)	505	(295)	679	(223)	927	(238)	a,b
Loading 2 Slope (%BW/s)	643	(115)	406	(117)	1093	(492)	1347	(558)	a,c
Unloading Slope (%BW/s)	-864	(106)	-701	(105)	-472	(92)	-405	(123)	a,b
Unloading 1 Slope (%BW/s)	-1135	(208)	-782	(130)	-648	(141)	-538	(181)	a,b,c
Unloading 2 Slope (%BW/s)	-712	(127)	-643	(84)	-394	(104)	-338	(96)	a,b
GRFap									
Stance Max (%BW)	6.1	(1.6)	5.3	(0.9)	14.0	(2.3)	16.2	(3.1)	a
Stance Min (%BW)	-8.2	(1.5)	-6.9	(1.8)	-12.5	(2.6)	-10.9	(1.6)	a
Stance Avg (%BW)	-0.3	(0.3)	-0.1	(0.3)	-0.3	(0.7)	-0.1	(0.3)	a
GRFml									
Loading Max (%BW)	3.5	(2.5)	5.0	(0.8)	6.7	(4.7)	9.7	(3.7)	a
Unloading Max (%BW)	3.9	(2.5)	5.8	(1.3)	5.8	(3.2)	8.0	(2.2)	a
Stance Avg (%BW)	1.1	(2.5)	2.0	(0.9)	2.5	(3.3)	4.1	(2.6)	a
FVM									
Stance Max (%BW*Ht)	0.22	(0.08)	0.28	(0.07)	0.22	(0.07)	0.36	(0.09)	a
Stance Min (%BW*Ht)	-0.18	(0.11)	-0.20	(0.10)	-0.15	(0.07)	-0.18	(0.10)	a
Stance Avg (%BW*Ht)	0.01	(0.06)	0.04	(0.04)	0.04	(0.06)	0.08	(0.05)	
Stance Impulse (%BW*Ht)	0.01	(0.06)	0.05	(0.05)	0.04	(0.06)	0.09	(0.06)	a

a=main effect for ascent/descent, b=main effect for group,
c=interaction with group difference in ascent, d=interaction with group difference in descent

Table 4.3: Ground Reaction Force & Free Vertical Moment Coefficient of Variance (%)

	Ascent				Descent				
	YA		OA		YA		OA		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Time									
Stance	4.4	(1.4)	5.1	(1.1)	4.7	(1.6)	7.7	(3.7)	a,b
Load	8.3	(4.3)	10.7	(4.8)	11.7	(6.1)	11.8	(4.3)	
Mid	6.7	(2.6)	11.4	(8.1)	15.9	(8.5)	13.8	(4.8)	a
Unload	11.3	(6.3)	12.4	(10.7)	24.6	(16.7)	17.8	(7.8)	a
GRFv									
Fz2	2.7	(1.0)	2.3	(1.0)	4.6	(1.1)	6.6	(3.4)	a
Fz3	3.6	(1.3)	3.9	(1.9)	5.7	(3.8)	5.4	(3.3)	a
Fz4	4.4	(1.5)	5.3	(2.2)	4.0	(3.2)	3.6	(1.9)	
Loading Slope	10.9	(5.1)	11.8	(3.5)	9.8	(3.8)	13.1	(4.2)	
Loading 1 Slope	13.9	(5.5)	35.7	(20.4)	10.9	(2.8)	14.2	(5.2)	a,b,c
Loading 2 Slope	11.5	(5.1)	14.8	(5.4)	20.3	(10.2)	36.2	(16.8)	a,b,d
Unloading Slope	8.7	(3.9)	11.9	(8.1)	14.5	(8.6)	14.0	(7.8)	
Unloading 1 Slope	8.9	(4.9)	16.4	(13.0)	17.2	(12.2)	26.2	(17.5)	b
Unloading 2 Slope	10.7	(3.2)	9.3	(3.6)	14.1	(6.2)	13.9	(7.8)	a
GRFap									
Stance Max	18.2	(10.5)	21.3	(12.3)	14.1	(6.6)	13.8	(7.7)	
Stance Min	20.9	(7.6)	17.1	(9.1)	11.4	(7.5)	15.0	(5.2)	
Stance Avg	840.7	(1539.4)	180.3	(125.8)	178.3	(126.1)	261.0	(173.3)	
GRFml									
Loading Max	131.6	(305.9)	42.6	(45.5)	66.5	(153.2)	129.4	(317.4)	
Unloading Max	29.0	(29.8)	30.4	(29.8)	46.7	(71.6)	34.5	(49.4)	a
Stance Avg	42.6	(49.9)	123.4	(180.7)	17.5	(4.4)	45.4	(55.2)	
FVM									
Stance Max	26.3	(16.1)	27.5	(17.3)	26.3	(13.8)	27.7	(14.8)	
Stance Min	37.6	(21.9)	54.4	(19.8)	40.6	(11.5)	38.1	(16.7)	
Stance Avg	151.3	(323.0)	111.9	(72.2)	41.2	(24.2)	53.1	(64.8)	
Stance Impulse	36.2	(16.7)	47.0	(13.2)	25.0	(17.1)	27.8	(16.9)	a

a=main effect for ascent/descent, b=main effect for group,
c=interaction with group difference in ascent, d=interaction with group difference in descent

Center of Pressure Displacements

***Loading Phase*.** A significant main effect for ascent/descent existed in all displacement CoPap Loading phase parameters ($p \leq 0.001$) with additional significant main effects for age groups in SD and PL ($p \leq 0.034$) with no significant interaction ($p \geq 0.10$) (Table 4.4). When examining subject variability, a significant main effect existed for ascent/descent in M-M, SD, and PL ($p \leq 0.001$) with no significant main effect between groups ($p \geq 0.218$) or any interaction ($p \geq 0.055$) (Table 4.5).

A significant main effect for ascent/descent in CoPml Loading phase for displacement parameters existed in all four variables ($p \leq 0.002$) with significant main effects between groups for each as well ($p \leq 0.001$) and no significant interactions ($p \geq 0.084$) (Table 4.4). When examining subject variability, a significant main effect for ascent/descent existed for SD ($p \leq 0.023$) with no significant effects between groups ($p \geq 0.181$) or any interaction ($p \geq 0.055$) (Table 4.5).

***Mid Phase*.** No significant main effect for ascent/descent existed in CoPap Mid phase displacement parameters ($p \geq 0.186$) or in the main effects for groups ($p \geq 0.431$) or any interaction ($p \geq 0.970$) (Table 4.4). When examining subject variability, no significant main effect for ascent/descent existed ($p \geq 0.22$) or between groups ($p \geq 0.382$) or any interaction ($p \geq 0.609$) (Table 4.5).

A significant main effect for ascent/descent existed in displacement for CoPml Mid phase PL ($p=0.018$) with significant main effects between groups for PL ($p=0.013$) and a significant interaction between groups and ascent for PL ($p=0.010$) (Table 4.4). When examining subject variability, there was no significant main effects between ascent/descent ($p \geq 0.098$) or between groups ($p \geq 0.053$) or interactions ($p \geq 0.084$) (Table 4.5).

***Unloading Phase*.** A significant main effect existed between groups in CoPap Unloading phase for all displacement parameters ($p \leq 0.006$) with no significant main effect for ascent/descent ($p \geq 0.217$) and no significant interaction ($p \geq 0.234$) (Table 4.4). When examining subject variability, no significant main effects existed for ascent/decent ($p \geq 0.224$) or between groups ($p \geq 0.053$) or interaction ($p \geq 0.570$)

(Table 4.5).

A significant main effect for ascent/descent existed in CoPml Unloading phase for the displacement parameters M-M and PL ($p \leq 0.050$) with significant main effects between groups for M-M and PL displacement parameters ($p \leq 0.044$), but no significant interaction ($p \geq 0.283$) (Table 4.4). When examining subject variability, no significant main effects existed for ascent/descent ($p \geq 0.457$) or between groups ($p \geq 0.371$,) or interactions ($p \geq 0.311$) (Table 4.5).

***Stance Phase*.** A significant main effect for ascent/descent existed in CoPap for the entire Stance phase for the displacement parameter E-S and PL ($p \leq 0.001$) with significant main effects between groups for all displacement parameters ($p \leq 0.045$) but no significant interactions ($p \geq 0.076$) (Table 4.4). When examining subject variability, a significant main effect for ascent/descent existed for SD ($p \leq 0.025$) with no significant main effects between groups ($p \geq 0.280$) or interaction ($p \geq 0.060$) (Table 4.5).

A significant main effect for ascent/descent existed in CoPml Stance phase for displacement parameters M-M, E-S and PL ($p \leq 0.004$) with significant main effects between groups for M-M, SD and PL ($p \leq 0.036$) but no significant interaction ($p \geq 0.088$) (Table 4.4). When examining subject variability, no significant main effects existed for ascent/descent ($p \geq 0.437$) or between groups ($p \geq 0.241$) or in the interactions ($p \geq 0.350$) (Table 4.5).

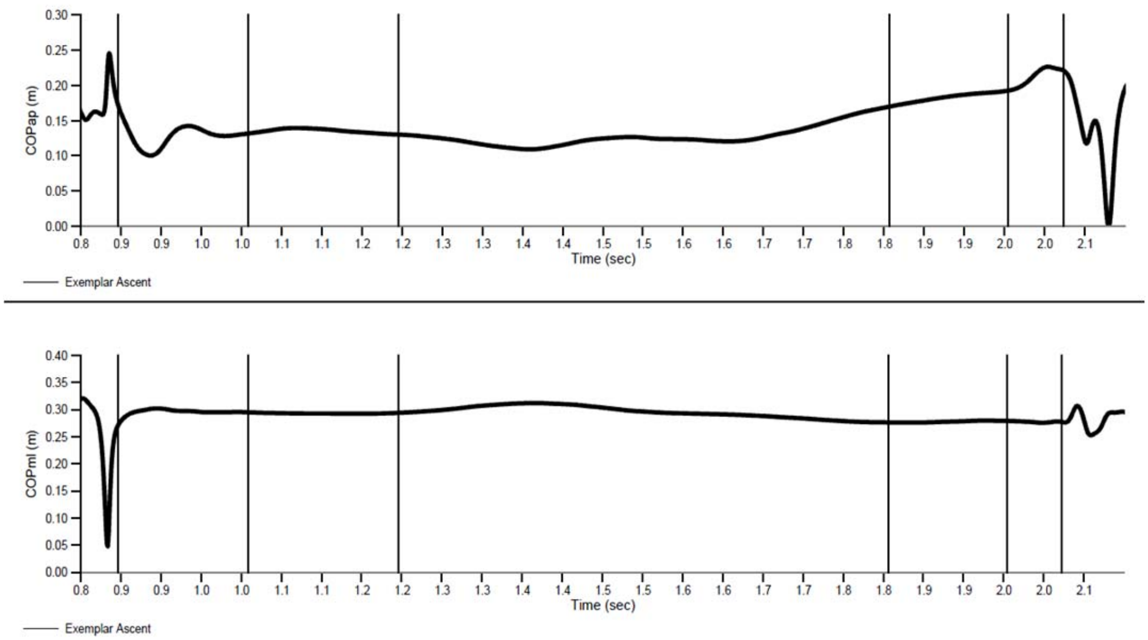

Figure 4.5 Exemplar subject profile of the CoPap (top) and CoPml (bottom) displacements. First and last vertical lines represent the entire stance phase. COP data were analyzed data from ~ 1.0s to 2.0s (second half of loading through first half of unloading) to ignore the highly variable ground contact and toe off regions. Middle two vertical lines mark the mid phase of stance.

Table 4.4: Center of Pressure Displacement Results*

	Ascent				Descent				
	YA		OA		YA		OA		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
COPap Loading									
M-M	8.6	(4.1)	9.6	(4.3)	16.9	(7.2)	22.1	(3.0)	a E-
S	-5.9	(6.8)	-3.4	(6.2)	16.5	(7.1)	21.7	(2.9)	a
SD	2.5	(1.2)	2.6	(1.1)	5.2	(2.5)	7.4	(1.2)	a,b
PL	9.5	(3.7)	13.7	(6.3)	18.0	(7.5)	22.8	(3.3)	a,b
COPml Loading									
M-M	7.5	(2.5)	15.2	(5.6)	16.9	(5.7)	29.3	(10.1)	a,b
E-S	1.1	(6.6)	7.6	(6.3)	-7.9	(15.7)	-26.4	(14.8)	a,b
SD	2.2	(0.8)	4.3	(1.6)	4.9	(1.7)	9.3	(3.4)	a,b
PL	9.0	(2.4)	22.5	(11.6)	18.8	(6.2)	31.8	(9.7)	a,b
COPap Mid									
M-M	24.3	(9.2)	24.1	(5.2)	22.1	(6.1)	23.8	(8.5)	
E-S	20.4	(7.9)	15.4	(11.5)	-13.6	(11.6)	-13.8	(5.9)	
SD	6.6	(2.9)	6.6	(1.3)	6.9	(2.1)	6.9	(2.7)	
PL	31.9	(10.7)	31.7	(8.8)	30.9	(8.2)	37.9	(13.3)	
COPml Mid									
M-M	21.8	(5.8)	22.5	(6.9)	18.2	(5.3)	27.2	(11.5)	
E-S	-9.0	(16.8)	-12.7	(11.4)	-1.1	(11.2)	8.8	(10.2)	
SD	6.4	(1.8)	6.6	(1.4)	5.6	(1.8)	8.1	(3.3)	
PL	30.6	(5.7)	32.7	(8.3)	29.9	(8.4)	47.5	(14.8)	a,b,d
COPap Unloading									
M-M	8.4	(3.2)	20.1	(9.8)	14.5	(9.4)	19.9	(8.0)	b
E-S	8.4	(3.2)	20.1	(9.8)	-14.4	(9.4)	-19.8	(7.6)	b
SD	2.4	(0.9)	6.2	(3.7)	4.1	(2.9)	5.8	(2.6)	b
PL	8.5	(3.2)	20.2	(9.9)	14.9	(9.5)	20.4	(8.5)	b
COPml Unloading									
M-M	6.3	(3.8)	12.4	(11.4)	11.4	(5.2)	17.4	(9.7)	a,b
E-S	-3.6	(6.2)	-5.1	(15.5)	1.1	(9.0)	8.5	(14.0)	
SD	1.9	(1.2)	3.9	(4.4)	3.4	(1.5)	5.5	(3.6)	
PL	7.1	(3.6)	13.9	(11.3)	14.6	(7.7)	21.5	(11.4)	a,b
COPap Stance									
M-M	33.5	(7.8)	44.1	(5.1)	34.4	(7.4)	41.9	(7.1)	b
E-S	22.9	(10.6)	32.1	(6.2)	-11.4	(5.0)	-11.8	(3.7)	a,b
SD	9.7	(2.8)	12.2	(1.7)	10.0	(2.8)	12.4	(2.6)	b
PL	49.8	(9.9)	65.4	(9.7)	63.7	(16.1)	80.9	(13.4)	a,b
COPml Stance									
M-M	26.4	(8.0)	32.4	(11.8)	30.3	(6.3)	42.8	(14.1)	a,b
E-S	-11.5	(16.3)	-10.2	(13.2)	-7.9	(17.7)	-9.0	(10.6)	a
SD	8.1	(2.5)	9.8	(3.7)	8.1	(2.1)	12.9	(5.5)	b
PL	46.5	(8.2)	69.0	(18.0)	63.1	(17.9)	100.5	(25.1)	a,b

*units for COPap are %foot length and COPml are %foot width
a=main effect for ascent/descent, b=main effect for group,
c=interaction with group difference in ascent, d=interaction with group difference in descent

Table 4.5: Center of Pressure Displacement Coefficients of Variance (%)

	Ascent				Descent				
	YA		OA		YA		OA		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
COPap Loading									
M-M	50.2	(21.2)	46.1	(20.4)	24.5	(15.2)	22.2	(10.7)	a
E-S	85.1	(46.1)	387.7	(736.8)	26.8	(17.0)	22.3	(10.7)	
SD	55.7	(20.9)	49.0	(17.1)	28.6	(12.7)	25.3	(12.3)	a
PL	41.7	(22.7)	43.0	(10.6)	24.4	(14.2)	21.6	(10.5)	a
COPml Loading									
M-M	48.9	(23.4)	39.3	(19.8)	33.1	(14.8)	26.8	(12.4)	
E-S	84.7	(37.9)	205.1	(132.2)	75.7	(120.1)	72.6	(135.6)	
SD	55.7	(23.2)	46.2	(24.2)	34.1	(15.5)	31.7	(15.6)	a
PL	39.6	(18.8)	38.4	(20.4)	31.7	(12.9)	22.9	(10.1)	
COPap Mid									
M-M	34.4	(41.2)	32.6	(10.7)	36.3	(17.5)	31.1	(12.8)	
E-S	47.1	(42.4)	80.5	(109.7)	109.3	(124.4)	198.6	(404.4)	
SD	40.9	(45.6)	34.1	(11.0)	38.5	(10.5)	34.0	(15.0)	
PL	25.3	(27.9)	26.3	(9.7)	31.6	(15.0)	27.9	(9.6)	
COPml Mid									
M-M	36.2	(16.7)	47.0	(13.2)	44.3	(12.1)	32.8	(13.7)	
E-S	62.1	(52.3)	152.7	(141.3)	131.8	(114.0)	302.6	(347.7)	
SD	40.8	(17.8)	47.0	(12.1)	47.4	(12.6)	37.0	(9.5)	
PL	28.2	(14.8)	39.9	(19.8)	36.5	(18.7)	27.2	(5.3)	
COPap Unloading									
M-M	35.5	(18.1)	38.7	(25.1)	47.2	(36.3)	42.5	(25.7)	
E-S	36.0	(18.2)	38.7	(25.1)	46.9	(35.0)	42.4	(25.8)	
SD	6.6	(2.9)	6.6	(1.3)	6.9	(2.1)	6.9	(2.7)	
PL	35.0	(18.1)	38.5	(24.9)	46.4	(37.0)	42.2	(24.4)	
COPml Unloading									
M-M	47.4	(17.1)	43.3	(29.2)	45.4	(25.1)	52.6	(38.8)	
E-S	219.3	(292.4)	105.1	(128.3)	130.2	(77.3)	140.9	(128.9)	
SD	53.6	(19.3)	43.1	(30.8)	51.3	(27.1)	58.7	(38.6)	
PL	40.8	(15.4)	39.9	(21.1)	39.6	(29.9)	49.6	(33.5)	
COPap Stance									
M-M	18.8	(16.5)	14.4	(5.4)	22.6	(15.7)	16.9	(8.6)	
E-S	35.0	(20.6)	18.6	(7.6)	92.3	(222.9)	22.8	(10.9)	
SD	25.4	(30.6)	18.4	(6.2)	33.5	(23.6)	26.0	(10.6)	a
PL	17.6	(16.9)	17.2	(5.7)	22.8	(14.2)	12.9	(6.3)	
COPml Stance									
M-M	27.4	(10.5)	30.6	(13.7)	27.8	(15.8)	28.4	(11.8)	
E-S	85.1	(93.2)	138.2	(98.3)	44.5	(19.4)	571.5	(1532.8)	
SD	31.9	(12.0)	35.4	(14.7)	32.8	(15.4)	36.6	(15.4)	
PL	21.8	(11.6)	25.1	(13.7)	22.6	(11.0)	20.0	(12.4)	

a=main effect for ascent/descent, b=main effect for group,
c=interaction with group difference in ascent, d=interaction with group difference in descent

Center of Pressure Velocities

***Loading Phase*.** A significant main effect for ascent/descent existed in all CoPap Loading phase velocity parameters ($p \leq 0.016$) with significant main effects for age groups in all velocity parameters ($p \leq 0.047$) (Table 4.6). Significant interactions existed between groups and descent for Vmin, Vavg and VSD velocity parameters ($p \leq 0.031$). When examining subject variability, a significant main effect existed for ascent/descent in Vmax and Vmin ($p \leq 0.004$) with no significant effect between groups ($p \geq 0.19$) or any interaction ($p \geq 0.220$) (Table 4.7).

There were significant main effects for ascent/descent in CoPml Loading phase for velocity parameters Vmax, Vavg and VSD ($p \leq 0.005$) with significant main effects between groups for Vmax, Vavg and VSD ($p \leq 0.001$) and significant interactions between groups and ascent/descent for Vmax, Vavg and VSD ($p \leq 0.005$) (Table 4.6). When examining subject variability, no significant main effects existed for ascent/descent ($p \geq 0.054$) or between groups ($p \geq 0.247$), now was there any interaction ($p \geq 0.100$) (Table 4.7).

***Mid Phase*.** A significant main effect for ascent/descent velocity existed in CoPap Mid phase for Vmin and VSD ($p \leq 0.007$) with no significant main effects between groups ($p \geq 0.156$) and no significant interaction ($p \geq 0.097$) (Table 4.6). When examining subject variability a significant main effect existed for ascent/descent for Vmin ($p = 0.050$) with no significant main effect for groups ($p \geq 0.193$) or interaction ($p \geq 0.100$) (Table 4.7).

A significant main effect for ascent/descent existed in velocity for CoPml Mid phase Vmax, Vmin, Vavg and VSD ($p \leq 0.049$) with significant main effects between groups for Vmax, Vmin, VSD ($p \leq 0.016$) (Table 4.6). There were significant interactions between descent and groups in Vmax and Vmin ($p \leq 0.005$) (Table 4.6). When examining subject variability, a significant main effect between groups existed in Vavg ($p \leq 0.034$) with no significant effect for ascent/descent ($p \geq 0.073$) and no significant interaction ($p \geq 0.084$) (Table 4.7).

Unloading Phase. A significant main effect existed for ascent/descent in CoPap Unloading phase for the velocity parameters Vmin and Vavg ($p \leq 0.002$) with a significant main effect between groups for Vavg ($p=0.009$). A significant interaction existed between groups and descent for Vavg ($p=0.044$) (Table 4.6). When examining subject variability, a significant main effect existed for ascent/descent for VSD ($p=0.013$) with no significant main effects between groups ($p \geq 0.363$) or interaction ($p \geq 0.278$) (Table 4.7).

A significant main effect for ascent/descent existed in CoPml Unloading phase for the velocity parameter Vmax ($p=0.003$), with significant main effects between groups for VSD ($p=0.043$) and no significant interaction ($p \geq 0.103$) (Table 4.6). When examining subject variability, no significant main effects existed for ascent/descent ($p \geq 0.237$) or between groups ($p>0.264$), or interactions ($p \geq 0.474$) (Table 4.7).

Stance Phase. A significant main effect for ascent/descent existed in CoPap for the entire Stance phase for velocity parameters Vmin, Vavg and Vavg ($p \leq 0.001$) with significant main effects between groups for Vmax, Vmin and VSD ($p \leq 0.006$) and a significant interaction between groups and descent for Vmin ($p \leq 0.017$) (Table 4.6). When examining subject variability, a significant main effect for ascent/descent existed for Vmin ($p \leq 0.008$) with no significant main effects between groups ($p \leq 0.264$) or interaction ($p \geq 0.406$) (Table 4.7).

A significant main effect for ascent/descent existed in CoPml Stance phase for velocity parameters Vmax, Vavg and VSD ($p \leq 0.002$) with significant main effects between groups for Vmax, Vmin and VSD ($p \leq 0.019$). A significant interaction existed for both ascent and descent between groups for Vmax and VSD ($p \leq 0.022$) (Table 4.6). When examining subject variability, no significant main effects existed for ascent/descent ($p \geq 0.317$) and between groups ($p \geq 0.187$) or interaction ($p \geq 0.406$) (Table 4.7).

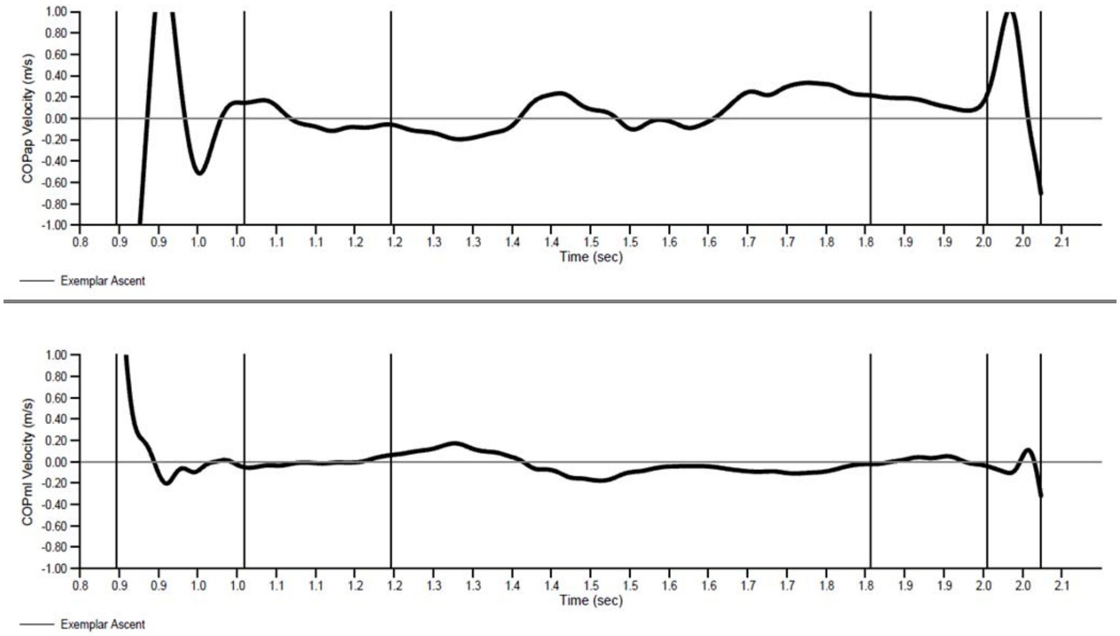

Figure 4.6 Exemplar subject profile the CoPap (top) and CoPml (bottom) velocities. First and last vertical lines represent the entire stance phase. COP data were analyzed data from ~ 1.0s to 2.0s (second half of loading through first half of unloading) to ignore the highly variable ground contact and toe off regions. Middle two vertical lines mark the mid phase of stance.

Table 4.6: Center of Pressure Velocity Results*

	Ascent				Descent				
	YA		OA		YA		OA		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
COPap Loading									
Vmax	33.0	(43.3)	135.4	(128.2)	445.5	(220.0)	716.9	(156.6)	a,b
Vmin	-152.8	(80.2)	-159.8	(56.6)	-15.2	(49.4)	-7.1	(58.1)	a,b,d
Vavg	-39.6	(47.3)	-14.8	(23.7)	154.4	(70.9)	314.0	(103.8)	a,b,d
VSD	50.9	(11.2)	64.3	(31.9)	149.1	(80.9)	247.8	(67.0)	a,b,d
COPml Loading									
Vmax	104.1	(77.3)	275.1	(128.6)	132.5	(207.5)	80.4	(146.7)	a,b,d
Vmin	-71.6	(48.2)	-160.4	(167.2)	-321.5	(240.6)	-911.0	(389.7)	
Vavg	6.1	(45.2)	33.0	(31.8)	-72.2	(145.0)	-396.5	(229.4)	a,b,d
VSD	49.8	(16.7)	102.1	(51.4)	139.3	(83.1)	330.7	(104.5)	a,b,c,d
COPap Mid									
Vmax	187.0	(59.8)	188.5	(73.2)	150.2	(57.6)	193.0	(89.1)	
Vmin	-57.8	(19.6)	-65.0	(28.5)	-164.5	(56.2)	-233.4	(51.7)	a
Vavg	35.9	(12.9)	33.6	(26.0)	-26.2	(29.4)	-28.7	(12.5)	
VSD	65.8	(21.8)	71.1	(25.2)	75.6	(16.1)	95.5	(28.1)	a
COPml Mid									
Vmax	89.7	(36.4)	102.4	(40.0)	153.7	(56.4)	308.0	(105.8)	a,b,d
Vmin	-138.5	(56.3)	-179.6	(66.8)	-176.4	(84.3)	-325.2	(104.1)	a,b,d
Vavg	-16.0	(30.2)	-26.7	(26.0)	-5.8	(26.7)	17.8	(19.3)	a
VSD	58.0	(14.0)	76.6	(18.9)	79.4	(33.4)	123.5	(45.0)	a,b
COPap Unloading									
Vmax	153.5	(87.9)	225.4	(82.5)	-8.2	(26.3)	-26.7	(33.4)	
Vmin	37.7	(21.2)	54.7	(26.6)	-167.6	(77.4)	-157.1	(23.1)	a
Vavg	77.0	(31.2)	124.6	(32.3)	-66.5	(20.6)	-85.4	(28.3)	a,b,c
VSD	35.4	(28.6)	56.0	(29.4)	42.7	(23.0)	37.1	(8.2)	
COPml Unloading									
Vmax	27.8	(60.0)	71.0	(59.8)	125.4	(108.3)	172.0	(111.9)	a
Vmin	-115.3	(73.5)	-161.4	(128.2)	-101.4	(47.9)	-113.9	(91.4)	
Vavg	-37.9	(65.3)	-20.2	(74.9)	-1.0	(42.7)	28.5	(50.0)	
VSD	43.5	(21.9)	72.9	(34.5)	62.6	(31.9)	79.8	(27.4)	b
COPap Stance									
Vmax	218.4	(78.4)	303.1	(104.3)	463.5	(206.8)	718.7	(153.1)	b
Vmin	-162.1	(68.4)	-167.8	(49.5)	-218.1	(54.8)	-247.5	(50.6)	a,b,d
Vavg	27.1	(12.1)	36.9	(6.6)	-15.0	(7.0)	-15.3	(4.2)	a
VSD	76.7	(15.5)	94.1	(20.6)	116.5	(30.1)	156.2	(26.1)	a,b
COPml Stance									
Vmax	145.6	(64.4)	290.6	(115.0)	261.7	(153.6)	364.3	(147.7)	a,b,c,d
Vmin	-170.5	(45.3)	-289.9	(162.5)	-359.2	(234.5)	-926.5	(362.6)	b
Vavg	-13.6	(19.4)	-12.4	(15.7)	-10.0	(23.4)	-11.2	(13.9)	a
VSD	67.4	(16.5)	105.2	(30.6)	110.0	(38.9)	202.6	(71.1)	a,b,c,d

*units for COPap are %foot length/s and COPml are %foot width/s

a=main effect for ascent/descent, b=main effect for group,

c=interaction with group difference in ascent, d=interaction with group difference in descent

Table 4.7: Center of Pressure Velocity Coefficients of Variance (%)

	Ascent				Descent				
	YA		OA		YA		OA Mean		
		SD	Mean	SD	Mean	SD	Mean	SD	
COPap Loading									
Vmax	122.3	(84.7)	102.0	(62.8)	31.8	(22.9)	32.7	(14.4)	a
Vmin	58.5	(36.3)	52.1	(22.3)	207.6	(193.9)	304.2	(304.1)	a
Vavg	82.2	(46.3)	314.6	(556.8)	32.6	(16.3)	44.5	(12.9)	
VSD	39.2	(22.6)	48.2	(13.7)	40.5	(28.4)	33.7	(17.2)	
COPml Loading									
Vmax	154.2	(288.1)	54.7	(21.2)	262.8	(300.2)	494.1	(682.7)	
Vmin	81.8	(103.3)	80.4	(47.5)	49.9	(36.8)	48.1	(52.6)	
Vavg	82.5	(34.1)	347.0	(508.2)	60.4	(65.9)	1947.6	(5696.4)	
VSD	34.9	(10.3)	44.8	(17.0)	34.5	(16.3)	29.8	(11.2)	
COPap Mid									
Vmax	40.5	(37.0)	32.1	(9.2)	60.0	(42.8)	33.5	(14.1)	
Vmin	45.6	(26.1)	54.4	(29.1)	34.0	(15.5)	37.2	(13.9)	a
Vavg	47.7	(40.6)	78.5	(96.5)	136.7	(193.5)	119.3	(160.7)	
VSD	36.7	(31.9)	28.7	(6.0)	40.4	(34.2)	25.3	(7.7)	
COPml Mid									
Vmax	48.3	(29.8)	55.9	(20.2)	40.6	(24.2)	41.8	(15.6)	
Vmin	39.3	(16.3)	43.4	(10.6)	48.7	(36.7)	28.3	(11.5)	
Vavg	65.4	(53.7)	157.9	(135.1)	130.0	(104.1)	308.7	(315.2)	b
VSD	27.4	(11.5)	38.0	(13.9)	39.2	(28.3)	22.7	(7.1)	
COPap Unloading									
Vmax	34.4	(19.1)	37.3	(16.1)	178.4	(114.8)	3624.8	(9743.4)	
Vmin	2512.9	(7852.1)	57.1	(77.8)	25.0	(17.1)	27.8	(16.9)	
Vavg	28.3	(14.8)	27.2	(10.8)	36.0	(14.8)	31.0	(17.8)	
VSD	53.3	(27.7)	64.8	(29.5)	37.4	(23.8)	37.9	(21.8)	a
COPml Unloading									
Vmax	87.2	(54.4)	181.8	(349.9)	3756.7	(11575.5)	49.7	(26.0)	
Vmin	85.8	(138.7)	117.6	(124.2)	46.0	(33.1)	66.0	(86.9)	
Vavg	271.3	(394.1)	107.1	(131.0)	149.3	(117.5)	142.8	(125.3)	
VSD	35.8	(14.9)	47.4	(22.8)	39.2	(23.6)	40.3	(27.9)	
COPap Stance									
Vmax	26.3	(9.7)	33.2	(15.2)	32.4	(19.6)	31.8	(13.4)	
Vmin	41.5	(20.1)	42.8	(12.9)	30.5	(12.6)	27.2	(15.0)	a
Vavg	33.5	(18.6)	18.9	(8.8)	83.8	(191.2)	22.8	(10.8)	
VSD	19.4	(6.4)	22.1	(7.6)	24.2	(17.7)	17.9	(8.0)	
COPml Stance									
Vmax	36.4	(22.1)	43.9	(20.3)	33.4	(14.9)	32.4	(13.4)	
Vmin	28.8	(10.1)	40.1	(16.1)	34.2	(20.2)	38.4	(26.4)	
Vavg	87.5	(95.2)	126.9	(81.2)	44.8	(20.6)	250.7	(584.4)	
VSD	23.5	(10.1)	24.5	(14.8)	28.3	(13.4)	22.5	(7.2)	

a=main effect for ascent/descent, b=main effect for group,
c=interaction with group difference in ascent, d=interaction with group difference in descent

CHAPTER 5

DISCUSSION

Considering that force control tends to be impaired to a greater extent during eccentric contractions more so than concentric contractions [7], it was hypothesized that differences in average values and variability of measures would exist between stair descent and stair ascent. It was further hypothesized that older adults would exhibit differences in average values and variability of measures compared to young adults, which are suggestive of reduced functional capacities related to quantities such as strength and control of force. Both hypotheses were supported by our results. However, differences between ascent and descent were more prevalent than those between age groups.

Stair Ascent versus Descent

Ground Reaction Force (GRF). As described in the literature the GRFv profile was “M” shaped during both ascent and descent [70]. Also consistent with previous reports Fz2 was higher during descent and Fz4 was higher during ascent [15]. There was no difference in Fz3 between ascent and descent suggesting the minimum peak during mid-stance phase is similar between ascent and descent, which is consistent with previous findings [1]. With these differences in relative peak values there were also differences in total stance time and the time of each phase, even though step cadence was the same during ascent and descent. Stance was longer in ascent along with the load and mid phases, though the unload phase was longer during descent. This is consistent with previous reports for the stance phase [62], however the present study is the first to analyze the time duration of load, mid and unload phases at a controlled speed. The relative time spent in each phase is most likely related to how quickly loads are generated within the load and unload phases. The loading slopes were greater during descent than ascent and the unloading slopes greater during ascent which is consistent with previous reports [1]. This had the effect of lengthening the “region of risk” in the early support phase, during which a fall may occur [15].

The COVs for the phase times and GRFv parameters were typically higher during descent than ascent, which was also representative of the previous research [132]. These differences can be explained by the difficulty in the inability to control eccentric contraction, which occur during stair descent. In contrast, during stair ascent, in which the muscle is shortening, or concentric contractions, it is easier to control the force. Stair descent has been described as a controlled fall [5], which may lend to inherent trial to trial differences.

As described in the literature the GRFap profiles show two peaks, a posterior directed (braking) peak followed by an anteriorly directed (propulsive) peak [120]. The GRFml profile is “M” shaped and directed medially throughout which is consistent with the literature [120]. In the GRFap the maximum during propulsion and minimum during braking were both larger during descent, consistent with a previous report [15]. The maximums during GRFml were both larger during ascent which is also consistent with a past study [15]. There were few differences in the COV between ascent and descent in the GRFap and GRFml. However the COV of the unloading GRFml max were much larger during descent. The present study is the first to analyze these values. These high COV values during descent suggest that it may be an unreliable parameter along with the stance averages of the GRFap and other GRFmlparameters. These COV values are most likely high because their parameter values are quite low. So any small change from trial to trial will be a large percent change in the COV. If they are deemed necessary for analysis more than the current five trials should be collected to ensure as representative a value as possible is available.

Free Vertical Moment (FVM). The FVM profile showed a positively directed propulsive and braking peak during ascent and a negatively directed minimum peak during descent. The maximum FVM was higher in descent with the negative minimum larger during ascent. However, the average was similar between the two, near zero. When the average was combined with time to calculate impulse, the FVM impulse was slightly greater during descent. The FVM has not been described during stair ascent or descent previously, only during step-downs [20]. The descent profile in the present study was similar to that found in a previous study examining FVM during step-downs [20]. The differences between ascent and descent could be explained by the rotational torque used to help push a person up in the 2nd half of the stance during ascent versus a “controlled fall” during the entire stance during stair descent.

Center of Pressure (CoP). The CoPap displacement profile exhibited an initial negative peak during touchdown which gradually rose positively with a small peak during mid-stance followed by a rise throughout the remaining stance. While the profile of the CoPap during stair descent and ascent has

not been described in the literature, it is consistent with a gait that begins with a braking phase followed by propulsion. The CoPap Stance exhibited a significantly greater E-S displacement during ascent compared to descent but a significantly greater PL displacement during descent compared to ascent.

The present study is the first to examine the CoPap displacement during stair ascent versus descent. The greater displacement during descent supports the notion of the difficulty to control the lengthening of the muscle during the “controlled fall” of the body during eccentric contractions that occur during stair descent. Interestingly, there were no significant differences in any of the displacement variables for the Mid and Unloading phases. However, during the Loading phase, all of the displacement variables were found to be significantly greater during descent. Thus, the Loading phase could justifiably explain the CoPap displacement for the entire Stance. The PL being greater during descent could be explainedby the difficulty to control the force of the foot during eccentric loading of the muscle, resulting in a greater PL displacement.

The CoPml displacement profile exhibited a relatively level value with a slight peak during the mid-stance. While the profile of the CoPml during ascent and descent has not been described in the literature, the majority of the movement during mid-stance is consistent with a gait that has single-limb stance at this time where the person's center of mass would "fall" medially without the opposing support of the contralateral limb. The CoPml Stance exhibited a significantly greater E-S displacement during ascent, compared to a significantly greater M-M and PL displacement during descent. All of the displacement variables were significantly greater during descent for the Loading phase. During the Mid phase, PL displacement was significantly greater during descent, while during the Unloading phase both the M-M and PL displacements were significantly greater during descent compared to ascent. Thus, the greater displacement during descent could be explained by the difficulty of eccentric contraction during the Loading phase contributing to the greater displacement in the overall Stance phase.

There were relatively few differences between ascent and descent in the COV for the COP displacement parameters. The few that existed were predominantly during the Loading phase in the A/P direction with greater values during ascent. This is contrary to expectations, since the descent tends to be harder to control than the ascent. However, the greater variable during the initial stance phase in ascent may be related to placing the foot in an elevated position. Sometimes the foot is placed solidlyon the step while in others it is on the edge of the step. During descent it is most likely easier to place the foot solidly on the step. Overall, the COVs are relatively high for all COP displacement parameters. Based on these results it appears that more than five trials may be needed to confidently explore them. The present findings are the first to examine the CoP displacement parameters comparing ascent to descent.

The CoPap velocity profile exhibited an "M" shape with the first maximum during the Mid phase while the second peak appeared at the end of the Stance, consistent with the changes that occur in its displacement profile. The CoPap velocities during the Stance phase are consistent with the CoPap displacement results where there is more movement posteriorly during descent and anteriorly during descent. However, unlike the displacement profiles where differences between ascent and descent are predominantly confined to the loading phase, they exist in all phases for velocity. This is most likely due to the higher sensitivity to change of velocity compared to displacement. The motion of the COP more posteriorly during descent, opposite the direction of motion of the body, may further compound the difficulty to control force during descent above and beyond those caused by the increased magnitude of force and its eccentric nature.

The CoPml velocity profile exhibited a slight maximum peak followed by a minimum peak during the mid-phase, but was otherwise relatively low and stable. Again, this is consistent with the changes in COP displacement. Unlike the CoPap velocities, however, the differences were highly contained to the loading and mid phases of stance. The majority of the velocity variables were greater during descent compared to ascent which could be explained by the difficulty to control force during eccentric contractions as well as the notion that a person is "falling" more when descending. With less control there is likely to be faster motion medially with the center of mass inside the base of support. This is further supported by the fact that it is occurring during the mid-phase where only one foot is on the ground. It appears that in the frontal plane the body is acting like an inverted pendulum at this time, falling towards the contralateral side. There were no differences in COV averages between ascent and descent in the CoPml velocities. As with the CoPap velocities, COVs are relatively high, suggesting that more trials may be needed to confidently analyze these parameters. The present study is the first to analyze CoP velocity during ascent and descent.

Young Adults (YA) versus Old Adults (OA)

While all variables were normalized by bodyweight or anthropometric dimensions as appropriate for comparison of two separate groups, there were no differences between the groups except for age. Therefore, considering that cadence was strictly controlled for using the audio pulsed metronome, differences between groups should be highly indicative of differences that exist as a result of healthy aging.

Time. While there were no differences in the total stance times between groups in either ascent or descent, the YA spent more time in mid-stance during ascent and more time in the loading phase during descent than the OA. On their own, whether these differences are due to a reduction in control of postural and balance or to the fact that OA generally choose a safer strategy remains unclear. While statistically different between groups these differences were relatively small (<0.10 s) and did not translate to differences in any other phase of ascent or descent. Relative to variability of phase times, the OA exhibited slightly greater total stance time COV, but still within good standards of repeatability. This has not been described in previous research.

Ground Reaction Force (GRF). Differences between groups in max/min GRFv values were limited, though the YA had a higher Fz2 during ascent and the OA had a higher Fz2 during descent. This is consistent with findings in previous studies [15]. There were no differences in COV between groups in the max/min GRFv. This has not been described in previous studies. While there were no differences in loading slope between groups, the OA had a higher initial Loading Slope in both ascent and descent

while the YA had a higher Loading Slope 2 during ascent which may reflect a lack of control at touchdown when compared with the young or an increase in joint stiffness [112]. Unloading Slopes were considerably greater in the YA. These findings confirm what others have found when GRF parameters were examined during stair negotiation [1, 16]. The age effects that have been found here can be interpreted as a less vigorous push off in the older adults. This generally more conservative profile is similar to that found in other contexts (such as step clearance) in which OA adopt more cautious strategies [27]. It has been previously shown that the Loading and Unloading slopes in the OA indicate a lower intensity in the weight transfer [136] to match the step cycle frequency of the audio metronome in the present study.

While there were no differences in the COVs of the Loading and Unloading Slopes between groups, there tended to be higher COVs within the Loading and Unloading 1 and 2 slopes for the OA. This might show reduced control of force between YA and OA. There were also interactions between groups during these slopes that suggest OA exhibit exceptionally higher variability during Loading Slope 1 during ascent and Loading Slope 2 during descent. Breaking the Loading and Unloading Slopes into two distinct regions is novel. From these results it is clear that they should be examined further. The higher variability in the OA may be useful diagnostically when testing for loss of control and increased risk for falls. There were no differences between YA and OA in any of the GRFap, GRFml or FVM in either mean value or of COV. These findings are consistent with previous findings [1, 20].

Center of Pressure (CoP) Displacements. During the present study we found that all of the CoPap displacement variables and most of the CoPml displacement variables during the Stance phase were significantly greater for the OA compared to the YA. This contradicts previous findings [90], in which CoPap and CoPml displacements were significantly greater for YA compared to OA. However, speed was not controlled for in the previous study and the YA most likely walked faster than the OA explaining the difference. As with the differences between ascent and descent, many of the COP displacement differences between groups occurred during the Loading phase, with values greater for OA compared to YA. There were also several age-related differences during the Unloading phase with displacement variables greater for the OA compared to YA. Thus, the Unloading phase could justifiably be contributed to differences observed in the Stance phase.

There were no significant differences related to age in COV for CoP displacement values in any phase. This is the first study to examine the variability of CoP displacement throughout a Stance phase, analyzing the Loading, Mid and Unloading phases. It could be the case that there were in fact age-related differences in COV but the standard deviations was too high. If this is the case, more trials examining COV for CoP displacements might be needed to determine if this is the case.

Center of Pressure (CoP) Velocity. The present study found that during the Stance phase the CoPap and CoPml velocity variables of Vmax, Vmin and VSD were significantly greater for OA compared to YA. These findings contradict previous findings [90] in which YA exhibited significantly greater mean CoPap velocities compared to OA. However, just as with the displacement data, speed was not controlled for in the previous study nor was stair ascent examined. The most likely faster pace of the YA in that study contributed to their faster COP velocities. Differences in velocities during Loading phase appears to contribute heavily to the greater velocities in the total stance of the OA compared to YA in both CoPap and CoPml velocity parameters. However, additional differences during the mid-phase also contribute to the differences in the CoPml velocities.

There were few COV values related to COP velocity that were significantly different across age groups for any of the phases. However, there may in fact be differences across age groups, but the standard deviations are too high. More trials would be needed to then examine this further. The present study is the first to analyze age-related COV differences in CoP velocity.

Limitations

There were several limitations to the current study. The current study has a relatively low sample size of older and younger adults. A further investigation with a larger sample size is warranted. Furthermore, the older subjects in this study had a high functional level; thus, the study population may not be representative of community dwelling older adults who have mobility or cognitive deficits. The interpretation of results is also limited by the data collected. As such differences in kinematics, joint kinetics, and muscle activity could not be examined to help fully understand the observed differences in force platform variables. Further, since we did see some significant effects in COV values, more trials could be warranted. Doubling the number of trials might be necessary to further explore the trial to trial variability observed. Although the speed was controlled, it was only a single speed that may be too challenging for some of the subjects. Therefore, the results are limited to a single cadence.

Future Work

Stair negotiation is a critically important component of successful independent living. Often it is one of the most difficult tasks faced in any given day. Our findings suggest that older adults have an impaired ability to control force compared to younger subjects during both stair ascent and descent. Measures of mean values and of trial-to-trial variability confirm this. These results reinforce the need to improve force control with the natural aging process and serve to identify factors that may limit stair negotiation. Thus, future studies examining stair negotiation with aging at a consistent speed is warranted. These future studies should exhibit longitudinal training and/or rehab studies to see how force control could be improved. Further, as the present study has shown, it is imperative to control for speed as we found differences from previous studies when speed was not controlled. Further, it would be important to increase the subject size to gather a better representation of community dwelling older adults who have mobility or cognitive defects. Moreover, more trials per subject and the addition of additional speeds might be warranted.

Practical Applications for Training

Based upon our findings we advise that maintaining older adult's mobility and function should be a top priority to reduce the incidence of falls during stair negotiation as an individual's overall balance, strength and ability to produce power are necessary to successfully complete this difficult task [60]. Based upon our results the ability to control force, especially during the loading and mid-stance phases during stair descent, should be considered important to reduce the risk of falls in the older adult population. Not only is the ability to generate strength and power important for this task, but also the ability to react quickly [61].

Since force production is highest during eccentric contractions when activated muscle is lengthened by an external load, and eccentric contractions are responsible for successful stair descent, negative work exercises are recommended. Because high-force production is the stimulus for increasing muscle size and strength, the elevated forces produced during eccentric contractions could be the most powerful stimulus to promote muscle growth and strength [48]. A further benefit of eccentric contractions is that the energy required (i.e. oxygen consumption) to produce negative work is trivial relative to magnitude (i.e., same force production) of positive work, displacing an external load [49]. This low energy requirement results in a perception of much “less effort” to those participating in this exercise. Therefore, the “high force, low cost” abilities of eccentric contractions are thought to be ideally suited to older adults participating in resistance exercises [51].

It is well known that muscle atrophy occurs with aging with a decline in muscle power output, eccentric force and level of physical function. Past studies have shown that eccentric exercises increase muscle strength and power [7, 62, 63]. Observations in Tai Chi movements show that they involve continuous knee flexion and extension motion during the weight-bearing phase of the movement [64]. Thus, a movement may require a relatively long duration of eccentric activation of leg muscles over a large range of motion. Past studies have hypothesized that long-term Tai Chi practice may improve eccentric strength of leg muscles, enhancing an individual’s postural stability [64]. However, Tai Chi is slow-movement oriented, which may not aid in developing the dynamic skills necessary for stair negotiation. Further, flexibility of the lower extremities should be stressed as it plays an important contributor to successful stair negotiation [42]. Finally, strength exercises requiring eccentric force control should be emphasized. Strength exercises such as the slow controlled lowering phase during a leg press exercise are recommended to improve force control.

Results of the present study found that older adults displayed decreased stability particularly in the frontal plane during the mid-stance phase. This is important as this was the phase of the stance in which a single leg controlled the force. In controlling this force, balance exercises on a single leg should be recommended for the older population. Exercises focusing on standing on one foot while moving the opposite leg in a slow controlled manner forwards and backwards, mimicking the stair negotiation task, to build up endurance and balance could possibly reduce the incidence of falls during stair descent in the elderly. Further, our results found that the loading phase is the segment with a great region of risk for the older adult population as shown by the differences from young adults in loading slopes as well as the displacement and velocity of the CoPap. This demonstrates the importance of being able to act quickly to complete this task. Thus, light plyometrics such as hopping with both legs and/or on a single leg, slightly forward would be advised to improve eccentric control. These exercises would improve strength and power during sudden loading, which might aid in improving force and the ability to react during the loading phase to reduce the incidence of falls during stair descent/ascent.

It should be stated that older adults should not overdo eccentric exercises since they may not be able to recover as quickly [53]. A recent study has suggested that older adults may have reduced abilities to rebuild proteins after eccentric exercise [53]. This fact puts caution to eccentric training, suggesting it should be used in moderation. Finally, during this exercise, extreme caution and safety should be considered when performing these dynamic exercises.

Importance

The findings of this study provide data for GRF, FVM and CoP parameters across age groups during stair ascent and descent and may help to identify risk factors associated with stair-based accidents or falling. While some of these conditions and variables have been examined before, many of the results are novel, especially comparisons that controlled speed in ascent and descent and between groups. For rehabilitation professionals, it is important to appreciate the nature and extent of adaptations of stair negotiation as a natural progression with aging in order to be able to identify unique altercations in force control patterns due to the superimposition of physical impairments. It is reasonable to expect that in clinical population, the capacity to redistribute the generation of the forces required to accomplish the task of stair negotiation may be compromised. However, knowledge about what the normative requirements are can provide a useful starting point upon which to base targeted interventions. The present findings of different GRF, FVM and CoP data between young and old age need to be considered during rehabilitation and/or clinical settings at old age.

Our results were able to potentially help better explain force control during stair negotiation. The significant differences and greater variability during descent highlights the difficulty to control force during eccentric contraction. Further, by dividing the Loading and Unloading slopes down to first half and second half phases we were able to show that OA exhibited exceptionally higher variability during the first half of the Loading Slope during ascent and during the second half of the Loading Slope during descent. The higher variability in the OA may be useful diagnostically when testing for loss of control and risked risk for falls.

The present study was the first to examine FVM across age groups during stair negotiation. Our findings in the differences and greater variability during stair descent further support the claim that FVM is more affected during eccentric contractions. Further, even though no significant differences were found across age groups, the present study is the first to examine this parameter. Thus, FVM appears not to be affected during stair negotiation as part of the natural aging process when speed is controlled.

Displacement of CoP provides an indication of dynamic stability during gait. The greater displacement and velocity during descent compared to ascent supports the notion of the difficulty to negotiation stair descent and further supports the difficulty to control eccentric contraction. Further, the greater CoP displacement exhibited in older adults is a lack of postural control during stair negotiation, perhaps explained by lack of neuromuscular control that is part of the natural aging process. The velocity of CoP provides valuable information about how individuals modulate gait when negotiating stairs. A greater CoP velocity may present a challenge to the maintenance of balance.   The velocity of CoP shown by the older adults was significantly greater than that shown by the younger adults. When we controlled for speed and examined ascent and descent we found older adults negotiated stairs at a higher velocity. Further, by breaking the Stance of ascent/descent into distinct phases we were able to show that the greatest displacement of these variables occurred during the Loading phase. This can prove to be useful for rehabilitation professionals to help identify altercations in force control patterns.

The ability to negotiation stairs is important in being able to live independently. Because OA negotiate stairs differently from YA it is important to separate the distinct biomechanical patters of stair ascent/descent. This is especially true when finding therapies to specifically target the ability to ascend/descend stairs. This thesis is the first step towards giving rehabilitation professionals a complete assessment of how different patterns of how force control during stair negotiation during stair ascending/descending is affected as part of the natural aging process.

REFERENCES

1. Larsen, A.H., et al., *Comparison of ground reaction forces and antagonist muscle coactivation during stair walking with ageing.* J Electromyogr Kinesiol, 2008. **18**(4): p. 568-80.
2. Nevitt, M.C., S.R. Cummings, and E.S. Hudes, *Risk factors for injurious falls: a prospective study.* J Gerontol, 1991. **46**(5): p. M164-70.
3. Campbell, A.J., et al., *Falls in old age: a study of frequency and related clinical factors.* Age Ageing, 1981. **10**(4): p. 264-70.
4. Campbell, A.J., M.J. Borrie, and G.F. Spears, *Risk factors for falls in a community-based prospective study of people 70 years and older.* J Gerontol, 1989. **44**(4): p. M112-7.
5. Svanstrom, L., *Falls on stairs: an epidemiological accident study.* Scand J Soc Med, 1974. **2**(3): p. 113-20.
6. Startzell, J.K., et al., *Stair negotiation in older people: a review.* J Am Geriatr Soc, 2000. **48**(5): p. 567-80.
7. Hortobagyi, T., et al., *Old adults perform activities of daily living near their maximal capabilities.* J Gerontol A Biol Sci Med Sci, 2003. **58**(5): p. M453-60.
8. Goncalves, L.C., et al., *Flexibility, functional autonomy and quality of life (QoL) in elderly yoga practitioners.* Arch Gerontol Geriatr, 2011. **53**(2): p. 158-62.
9. Goble, D.J., et al., *Proprioceptive sensibility in the elderly: degeneration, functional consequences and plastic-adaptive processes.* Neurosci Biobehav Rev, 2009. **33**(3): p. 271-8.
10. Laidlaw, D.H., M. Bilodeau, and R.M. Enoka, *Steadiness is reduced and motor unit discharge is more variable in old adults.* Muscle Nerve, 2000. **23**(4): p. 600-12.
11. Connelly, D.M., et al., *Motor unit firing rates and contractile properties in tibialis anterior of young and old men.* J Appl Physiol, 1999. **87**(2): p. 843-52.
12. Andriacchi, T.P., et al., *A study of lower-limb mechanics during stair-climbing.* J Bone Joint Surg Am, 1980. **62**(5): p. 749-57.
13. McFadyen, B.J. and D.A. Winter, *An integrated biomechanical analysis of normal stair ascent and descent.* J Biomech, 1988. **21**(9): p. 733-44.
14. Golriz, S., et al., *The validity of a portable clinical force plate in assessment of static postural control: concurrent validity study.* Chiropr Man Therap, 2012. **20**(1): p. 15.
15. Hamel, K.A., et al., *A comparison of foot/ground interaction during stair negotiation and level walking in young and older women.* Ergonomics, 2005. **48**(8): p. 1047-56.
16. Christina, K.A. and P.R. Cavanagh, *Ground reaction forces and frictional demands during stair descent: effects of age and illumination.* Gait Posture, 2002. **15**(2): p. 153-8.
17. Karamanidis, K. and A. Arampatzis, *Evidence of mechanical load redistribution at the knee joint in the elderly when ascending stairs and ramps.* Ann Biomed Eng, 2009. **37**(3): p. 467-76.
18. Lafond, D., et al., *Intrasession reliability of center of pressure measures of postural steadiness in healthy elderly people.* Arch Phys Med Rehabil, 2004. **85**(6): p. 896-901.
19. Bleuse, S., et al., *Effect of age on anticipatory postural adjustments in unilateral arm movement.* Gait Posture, 2006. **24**(2): p. 203-10.
20. Buckley, J.G., S.F. Jones, and L. Johnson, *Age-differences in the free vertical moment during step descent.* Clin Biomech (Bristol, Avon), 2010. **25**(2): p. 147-53.
21. Alexander, N.B., A.B. Schultz, and D.N. Warwick, *Rising from a chair: effects of age and functional ability on performance biomechanics.* J Gerontol, 1991. **46**(3): p. M91-8.
22. Luukinen, H., et al., *Fracture risk associated with a fall according to type of fall among the elderly.* Osteoporos Int, 2000. **11**(7): p. 631-4.
23. Stevens, J.A., et al., *The costs of fatal and non-fatal falls among older adults.* Inj Prev, 2006.

12(5): p. 290-5.
24. Williamson, J.D. and L.P. Fried, *Characterization of older adults who attribute functional decrements to "old age".* J Am Geriatr Soc, 1996. **44**(12): p. 1429-34.
25. Tinetti, M.E., M. Speechley, and S.F. Ginter, *Risk factors for falls among elderly persons living in the community.* N Engl J Med, 1988. **319**(26): p. 1701-7.
26. Booth, F.W., S.H. Weeden, and B.S. Tseng, *Effect of aging on human skeletal muscle and motor function.* Med Sci Sports Exerc, 1994. **26**(5): p. 556-60.
27. Simoneau, G.G., et al., *The influence of visual factors on fall-related kinematic variables during stair descent by older women.* J Gerontol, 1991. **46**(6): p. M188-95.
28. Brechter, J.H. and C.M. Powers, *Patellofemoral joint stress during stair ascent and descent in persons with and without patellofemoral pain.* Gait Posture, 2002. **16**(2): p. 115-23.
29. Hasselkus, B.R. and G.M. Shambes, *Aging and postural sway in women.* J Gerontol, 1975. **30**(6): p. 661-7.
30. Era, P. and E. Heikkinen, *Postural sway during standing and unexpected disturbance of balance in random samples of men of different ages.* J Gerontol, 1985. **40**(3): p. 287-95.
31. Baloh, R.W., et al., *Comparison of static and dynamic posturography in young and older normal people.* J Am Geriatr Soc, 1994. **42**(4): p. 405-12.
32. Fernie, G.R., et al., *The relationship of postural sway in standing to the incidence of falls in geriatric subjects.* Age Ageing, 1982. **11**(1): p. 11-6.
33. Ageberg, E., et al., *Balance in single-limb stance in healthy subjects--reliability of testing procedure and the effect of short-duration sub-maximal cycling.* BMC Musculoskelet Disord, 2003. **4**: p. 14.
34. Laughton, C.A., et al., *Aging, muscle activity, and balance control: physiologic changes associated with balance impairment.* Gait Posture, 2003. **18**(2): p. 101-8.
35. Doherty, T.J., *Invited review: Aging and sarcopenia.* J Appl Physiol, 2003. **95**(4): p. 1717-27.
36. Wolfson, L., et al., *Strength is a major factor in balance, gait, and the occurrence of falls.* J Gerontol A Biol Sci Med Sci, 1995. **50 Spec No**: p. 64-7.
37. Janssen, I., *Influence of sarcopenia on the development of physical disability: the Cardiovascular Health Study.* J Am Geriatr Soc, 2006. **54**(1): p. 56-62.
38. Nakazawa, K., et al., *Differences in activation patterns in elbow flexor muscles during isometric, concentric and eccentric contractions.* Eur J Appl Physiol Occup Physiol, 1993. **66**(3): p. 214-20.
39. Spanjaard, M., et al., *Influence of step-height and body mass on gastrocnemius muscle fascicle behavior during stair ascent.* J Biomech, 2008. **41**(5): p. 937-44.
40. Theodorou, A.A., et al., *Stair descending exercise increases muscle strength in elderly males with chronic heart failure.* BMC Res Notes, 2013. **6**: p. 87.
41. Astrand, I., et al., *Reduction in maximal oxygen uptake with age.* J Appl Physiol, 1973. **35**(5): p. 649-54.
42. Astrand, P.O., *Physical performance as a function of age.* JAMA, 1968. **205**(11): p. 729-33.
43. Waters, R.L., et al., *Comparative cost of walking in young and old adults.* J Orthop Res, 1983. **1**(1): p. 73-6.
44. Zhang, J., et al., *Age-specific patterns of factors related to fatal motor vehicle traffic crashes: focus on young and elderly drivers.* Public Health, 1998. **112**(5): p. 289-95.
45. Donelan, J.M., R. Kram, and A.D. Kuo, *Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking.* J Exp Biol, 2002. **205**(Pt 23): p. 3717-27.
46. Doke, J., J.M. Donelan, and A.D. Kuo, *Mechanics and energetics of swinging the human leg.* J Exp Biol, 2005. **208**(Pt 3): p. 439-45.
47. Alexander, R.M., *Energy-saving mechanisms in walking and running.* J Exp Biol, 1991. **160**: p. 55-69.

48. Hunter, L.C., E.C. Hendrix, and J.C. Dean, *The cost of walking downhill: is the preferred gait energetically optimal?* J Biomech, 2010. **43**(10): p. 1910-5.
49. Himann, J.E., et al., *Age-related changes in speed of walking.* Med Sci Sports Exerc, 1988. **20**(2): p. 161-6.
50. Duncan, P.W., et al., *How do physiological components of balance affect mobility in elderly men?* Arch Phys Med Rehabil, 1993. **74**(12): p. 1343-9.
51. Trueblood, P.R. and L.Z. Rubenstein, *Assessment of instability and gait in elderly persons.* Compr Ther, 1991. **17**(8): p. 20-9.
52. Shiomi, T., *Effects of different patterns of stairclimbing on physiological cost and motor efficiency.* J Hum Ergol (Tokyo), 1994. **23**(2): p. 111-20.
53. Studenski, S., et al., *Predicting falls: the role of mobility and nonphysical factors.* J Am Geriatr Soc, 1994. **42**(3): p. 297-302.
54. Livingston, L.A., J.M. Stevenson, and S.J. Olney, *Stairclimbing kinematics on stairs of differing dimensions.* Arch Phys Med Rehabil, 1991. **72**(6): p. 398-402.
55. Riener, R., M. Rabuffetti, and C. Frigo, *Stair ascent and descent at different inclinations.* Gait Posture, 2002. **15**(1): p. 32-44.
56. Andriacchi, T.P., J.O. Galante, and R.W. Fermier, *The influence of total knee-replacement design on walking and stair-climbing.* J Bone Joint Surg Am, 1982. **64**(9): p. 1328-35.
57. Olney, *Joint structure and function: a comprehensive analysis*, in *Gait*, L.P.a.N. CC, Editor. 2005: Philadelphia.
58. Nadeau, S., B.J. McFadyen, and F. Malouin, *Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: what are the challenges compared to level walking?* Clin Biomech (Bristol, Avon), 2003. **18**(10): p. 950-9.
59. Protopapadaki, A., et al., *Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals.* Clin Biomech (Bristol, Avon), 2007. **22**(2): p. 203-10.
60. Mian, O.S., et al., *Kinematics of stair descent in young and older adults and the impact of exercise training.* Gait Posture, 2007. **25**(1): p. 9-17.
61. Benedetti, M.G., et al., *How do the elderly negotiate a step? A biomechanical assessment.* Clin Biomech (Bristol, Avon), 2007. **22**(5): p. 567-73.
62. Stacoff, A., et al., *Ground reaction forces on stairs: effects of stair inclination and age.* Gait Posture, 2005. **21**(1): p. 24-38.
63. Dixon, P.C. and D.J. Pearsall, *Gait dynamics on a cross-slope walking surface.* J Appl Biomech, 2010. **26**(1): p. 17-25.
64. Fujiyama, T., Tyler, Nick, *Pedestrian Speeds on Stairs - An Initial Step for a Stimulation Model.* Newcastle upon Tyne, 2004.
65. Costigan, P.A., K.J. Deluzio, and U.P. Wyss, *Knee and hip kinetics during normal stair climbing.* Gait Posture, 2002. **16**(1): p. 31-7.
66. Spanjaard, M., et al., *Gastrocnemius muscle fascicle behavior during stair negotiation in humans.* J Appl Physiol, 2007. **102**(4): p. 1618-23.
67. Patla, A.E., et al., *Visual control of locomotion: strategies for changing direction and for going over obstacles.* J Exp Psychol Hum Percept Perform, 1991. **17**(3): p. 603-34.
68. Woollacott, M.H. and P.F. Tang, *Balance control during walking in the older adult: research and its implications.* Phys Ther, 1997. **77**(6): p. 646-60.
69. Inglin, B. and M. Woollacott, *Age-related changes in anticipatory postural adjustments associated with arm movements.* J Gerontol, 1988. **43**(4): p. M105-13.
70. Winter, D.A., *Foot trajectory in human gait: a precise and multifactorial motor control task.* Phys Ther, 1992. **72**(1): p. 45-53; discussion 54-6.

71. Christou, E.A. and L.G. Carlton, *Age and contraction type influence motor output variability in rapid discrete tasks.* J Appl Physiol, 2002. **93**(2): p. 489-98.
72. Grabiner, M.D. and T.M. Owings, *EMG differences between concentric and eccentric maximum voluntary contractions are evident prior to movement onset.* Exp Brain Res, 2002. **145**(4): p. 505-11.
73. Sharma, L. and Y.C. Pai, *Impaired proprioception and osteoarthritis.* Curr Opin Rheumatol, 1997. **9**(3): p. 253-8.
74. Enoka, R.M., et al., *Mechanisms that contribute to differences in motor performance between young and old adults.* J Electromyogr Kinesiol, 2003. **13**(1): p. 1-12.
75. Cole, K.J., *Grasp force control in older adults.* J Mot Behav, 1991. **23**(4): p. 251-8.
76. Hortobagyi, T., et al., *Effects of standard and eccentric overload strength training in young women.* Med Sci Sports Exerc, 2001. **33**(7): p. 1206-12.
77. Vandervoort, A.A., J.F. Kramer, and E.R. Wharram, *Eccentric knee strength of elderly females.* J Gerontol, 1990. **45**(4): p. B125-8.
78. Hortobagyi, T., et al., *The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength.* J Gerontol A Biol Sci Med Sci, 1995. **50**(6): p. B399-406.
79. Carolan, B. and E. Cafarelli, *Adaptations in coactivation after isometric resistance training.* J Appl Physiol, 1992. **73**(3): p. 911-7.
80. Allen, D.G., et al., *Calcium and the damage pathways in muscular dystrophy.* Can J Physiol Pharmacol, 2010. **88**(2): p. 83-91.
81. Yeung, E.W., et al., *Effects of stretch-activated channel blockers on [Ca2+]i and muscle damage in the mdx mouse.* The Journal of physiology, 2005. **562**(Pt 2): p. 367-80.
82. Burnett, R.A., D.H. Laidlaw, and R.M. Enoka, *Coactivation of the antagonist muscle does not covary with steadiness in old adults.* J Appl Physiol, 2000. **89**(1): p. 61-71.
83. Beal, M.F., *Mitochondria take center stage in aging and neurodegeneration.* Ann Neurol, 2005. **58**(4): p. 495-505.
84. Graves, A.E., K.W. Kornatz, and R.M. Enoka, *Older adults use a unique strategy to lift inertial loads with the elbow flexor muscles.* J Neurophysiol, 2000. **83**(4): p. 2030-9.
85. Frontera, W.R., et al., *Aging of skeletal muscle: a 12-yr longitudinal study.* J Appl Physiol, 2000. **88**(4): p. 1321-6.
86. Lexell, J. and C.C. Taylor, *Variability in muscle fibre areas in whole human quadriceps muscle: effects of increasing age.* J Anat, 1991. **174**: p. 239-49.
87. Lexell, J., C.C. Taylor, and M. Sjostrom, *What is the cause of the ageing atrophy? Total number, size and proportion of different fiber types studied in whole vastus lateralis muscle from 15- to 83-year-old men.* J Neurol Sci, 1988. **84**(2-3): p. 275-94.
88. Kent-Braun, J.A., A.V. Ng, and K. Young, *Skeletal muscle contractile and noncontractile components in young and older women and men.* J Appl Physiol, 2000. **88**(2): p. 662-8.
89. Kent-Braun, J.A. and A.V. Ng, *Specific strength and voluntary muscle activation in young and elderly women and men.* J Appl Physiol, 1999. **87**(1): p. 22-9.
90. Kim, H.D., *A Comparison of the Center of Pressure during Stair Descent in Young and Healthy Elderly Adults.* Journal of Physical Therapy Science, 2009. **21**(2): p. 129-134.
91. Nagai, K., et al., *Effect of the muscle coactivation during quiet standing on dynamic postural control in older adults.* Arch Gerontol Geriatr, 2013. **56**(1): p. 129-33.
92. Zanou, N., et al., *Essential role of TRPV2 ion channel in the sensitivity of dystrophic muscle to eccentric contractions.* FEBS Lett, 2009. **583**(22): p. 3600-4.

93. Yeung, E.W., S.I. Head, and D.G. Allen, *Gadolinium reduces short-term stretch-induced muscle damage in isolated mdx mouse muscle fibres.* The Journal of physiology, 2003. **552**(Pt 2): p. 449-58.
94. Deconinck, N. and B. Dan, *Pathophysiology of duchenne muscular dystrophy: current hypotheses.* Pediatr Neurol, 2007. **36**(1): p. 1-7.
95. Kurebayashi, N. and Y. Ogawa, *Depletion of Ca2+ in the sarcoplasmic reticulum stimulates Ca2+ entry into mouse skeletal muscle fibres.* The Journal of physiology, 2001. **533**(Pt 1): p. 185-99.
96. Warren, G.L., et al., *What mechanisms contribute to the strength loss that occurs during and in the recovery from skeletal muscle injury?* J Orthop Sports Phys Ther, 2002. **32**(2): p. 58-64.
97. Enoka, R.M., *Eccentric contractions require unique activation strategies by the nervous system.* J Appl Physiol, 1996. **81**(6): p. 2339-46.
98. Doherty, T.J., A.A. Vandervoort, and W.F. Brown, *Effects of ageing on the motor unit: a brief review.* Can J Appl Physiol, 1993. **18**(4): p. 331-58.
99. Poulin, M.J., et al., *Eccentric and concentric torques of knee and elbow extension in young and older men.* Can J Sport Sci, 1992. **17**(1): p. 3-7.
100. Gefen, A., *Simulations of foot stability during gait characteristic of ankle dorsiflexor weakness in the elderly.* IEEE Trans Neural Syst Rehabil Eng, 2001. **9**(4): p. 333-7.
101. McGibbon, C.A. and D.E. Krebs, *Age-related changes in lower trunk coordination and energy transfer during gait.* J Neurophysiol, 2001. **85**(5): p. 1923-31.
102. Alfredson, H., et al., *Heavy-load eccentric calf muscle training for the treatment of chronic Achilles tendinosis.* Am J Sports Med, 1998. **26**(3): p. 360-6.
103. Lavender, A.P. and K. Nosaka, *Comparison between old and young men for changes in makers of muscle damage following voluntary eccentric exercise of the elbow flexors.* Appl Physiol Nutr Metab, 2006. **31**(3): p. 218-25.
104. Jarvinen, M., et al., *Histopathological findings in chronic tendon disorders.* Scand J Med Sci Sports, 1997. **7**(2): p. 86-95.
105. Herzog, W., et al., *Mysteries of muscle contraction.* J Appl Biomech, 2008. **24**(1): p. 1-13.
106.Lindstedt, S.L., P.C. LaStayo, and T.E. Reich, *When active muscles lengthen: properties and consequences of eccentric contractions.* News Physiol Sci, 2001. **16**: p. 256-61.
107. Maganaris, C.N., et al., *Biomechanics and pathophysiology of overuse tendon injuries: ideas on insertional tendinopathy.* Sports Med, 2004. **34**(14): p. 1005-17.
108. Bassey, E.J., et al., *Leg extensor power and functional performance in very old men and women.* Clin Sci (Lond), 1992. **82**(3): p. 321-7.
109. Bean, J., et al., *Weighted stair climbing in mobility-limited older people: a pilot study.* J Am Geriatr Soc, 2002. **50**(4): p. 663-70.
110. Sayers, S.P., et al., *Effect of leg muscle contraction velocity on functional performance in older men and women.* J Am Geriatr Soc, 2005. **53**(3): p. 467-71.
111. Bosco, C. and P.V. Komi, *Influence of aging on the mechanical behavior of leg extensor muscles.* Eur J Appl Physiol Occup Physiol, 1980. **45**(2-3): p. 209-19.
112. Hortobagyi, T. and P. DeVita, *Altered movement strategy increases lower extremity stiffness during stepping down in the aged.* J Gerontol A Biol Sci Med Sci, 1999. **54**(2): p. B63-70.
113. Hortobagyi, T. and P. DeVita, *Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging.* J Electromyogr Kinesiol, 2000. **10**(2): p. 117-26.
114. Cavagna, G.A., N.C. Heglund, and C.R. Taylor, *Mechanical work in terrestrial locomotion: two basic mechanisms for minimizing energy expenditure.* The American journal of physiology, 1977. **233**(5): p. R243-61.
115. Kaufman, K.R., et al., *Gait characteristics of patients with knee osteoarthritis.* J Biomech, 2001. **34**(7): p. 907-15.

116. Riener, R. and T. Edrich, *Identification of passive elastic joint moments in the lower extremities.* J Biomech, 1999. **32**(5): p. 539-44.
117. Nigg, B.M., et al., *Range of motion of the foot as a function of age.* Foot Ankle, 1992. **13**(6): p. 336-43.
118. Stussi, E., U. Noelpp, and B. Nold, *[Active switch for use with different gamma cameras in a data processing system].* Nuklearmedizin, 1980. **19**(2): p. 93-4.
119. Zachazewski, J.E., P.O. Riley, and D.E. Krebs, *Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects.* J Rehabil Res Dev, 1993. **30**(4): p. 412-22.
120. Reeves, N.D., M. Spanjaard, and A.A. Mohagheghi, *The demands of stair descent relative to maximum capacities in elderly and young adults.* Journal of Electromyography and Kinesiology, 2006(18): p. 218-227.
121. Townsend, M.A., et al., *Variability and biomechanics of synergy patterns of some lower-limb muscles during ascending and descending stairs and level walking.* Med Biol Eng Comput, 1978. **16**(6): p. 681-8.
122. Hahn, M.E. and L.S. Chou, *Age-related reduction in sagittal plane center of mass motion during obstacle crossing.* J Biomech, 2004. **37**(6): p. 837-44.
123. Martin, M., et al., *Gait initiation in community-dwelling adults with Parkinson disease: Comparison with older and younger adults without the disease.* Phys Ther, 2002. **82**(6): p. 566-577.
124. Chou, L.S., et al., *Medio-lateral motion of the center of mass during obstacle crossing distinguishes elderly individuals with imbalance.* Gait Posture, 2003. **18**(3): p. 125-33.
125. Polcyn, A.F., et al., *Age-related changes in the initiation of gait: Degradation of central mechanisms for momentum generation.* Arch Phys Med Rehabil, 1998. **79**(12): p. 1582-1589.
126. Shephard, R.J., M. Berridge, and W. Montelpare, *On the Generality of the Sit and Reach Test – an Analysis of Flexibility Data for an Aging Population.* Research Quarterly for Exercise and Sport, 1990. **61**(4): p. 326-330.
127. Cunningham, D.A., et al., *Determinants of self-selected walking pace across ages 19 to 66.* J Gerontol, 1982. **37**(5): p. 560-4.
128. Lee, H.J. and L.S. Chou, *Balance control during stair negotiation in older adults.* J Biomech, 2007. **40**(11): p. 2530-6.
129. Wang, Y., V.M. Zatsiorsky, and M.L. Latash, *Muscle synergies involved in preparation to a step made under the self-paced and reaction time instructions.* Clin Neurophysiol, 2006. **117**(1): p. 41-56.
130. Robert, T., et al., *Angular momentum synergies during walking.* Exp Brain Res, 2009. **197**(2): p. 185-97.
131. Kimmeskamp, S. and E.M. Hennig, *Heel to toe motion characteristics in Parkinson patients during free walking.* Clin Biomech (Bristol, Avon), 2001. **16**(9): p. 806-12.
132. Leitner, M., et al., *Test-retest reliability of vertical ground reaction forces during stair climbing in the elderly population.* Gait Posture, 2011. **34**(3): p. 421-5.
133. Stacoff, A., et al., *Ground reaction forces on stairs. Part II: knee implant patients versus normals.* Gait Posture, 2007. **26**(1): p. 48-58.
134. Wang, Y. and K. Watanabe, *Limb dominance related to the variability and symmetry of the vertical ground reaction force and center of pressure.* J Appl Biomech, 2012. **28**(4): p. 473-8.
135. Holden, J.P. and P.R. Cavanagh, *The free moment of ground reaction in distance running and its changes with pronation.* J Biomech, 1991. **24**(10): p. 887-97.
136. Nigg, B., Skeleryk, BN, *Gait characteristics of the elderly.* Clin Biomech (Bristol, Avon), 1988. **3**(2): p. 79-87.
137. Buchner, D.M., *Preserving mobility in older adults.* West J Med, 1997. **167**(4): p. 258-64.

138. Lan, C., et al., *Tai Chi Chuan to improve muscular strength and endurance in elderly individuals: a pilot study.* Arch Phys Med Rehabil, 2000. **81**(5): p. 604-7.
139. Kullman, E.L., et al., *Age attenuates leucine oxidation after eccentric exercise.* Int J Sports Med, 2013. **34**(8): p. 695-9.

APPENDIX

NOTICE OF APPROVAL FOR HUMAN RESEARCH

DATE:	March 23, 2012
TO:	Tracy, Brian, Health & Exercise Science Grossman, Amy, 1582 Dept Hlth & Exer Sci, Israel, Richard, Health & Exercise Science
FROM:	Barker, Janell, , CSU IRB 1
PROTOCOL TITLE:	Ankle Steadiness, Postural Control, and Physical Frailty
FUNDING SOURCE:	National Institute of Health : 79097
PROTOCOL NUMBER:	09-946H
APPROVAL PERIOD:	Approval Date: April 01, 2012

Expiration Date: March 31, 2013

The CSU Institutional Review Board (IRB) for the protection of human subjects has reviewed the protocol entitled: Ankle Steadiness, Postural Control, and Physical Frailty. The project has been approved for the procedures and subjects described in the protocol. This protocol must be reviewed for renewal on a yearly basis for as long as the research remains active. Should the protocol not be renewed before expiration, all activities must cease until the protocol has been re-reviewed.

If approval did not accompany a proposal when it was submitted to a sponsor, it is the PI's responsibility to provide the sponsor with the approval notice.

This approval is issued under Colorado State University's Federal Wide Assurance 00000647 with the Office for Human Research Protections (OHRP). If you have any questions regarding your obligations under CSU's Assurance, please do not hesitate to contact us.

Please direct any questions about the IRB's actions on this project to:

Janell Barker, Senior IRB Coordinator - (970) 491-1655 Janell.Barker@Colostate.edu
Evelyn Swiss, IRB Coordinator - (970) 491-1381 Evelyn.Swiss@Colostate.edu

Barker, Janell

[redacted]

Barker, Janell

Includes:
Approval is to recruit the remaining 100 young adults and 100 older adults with the approved recruitment and consent material. This reflects an approved amendment to increase approved participant numbers: add 100 young adults, and to add 89 to the current remaining balance of 11 older adults. The above-referenced project was approved by the Institutional Review Board with the condition that the approved consent form is signed by the subjects and each subject is given a copy of the form. NO changes may be made to this document without first obtaining the approval of the IRB. NOTE: This approval also reflects an approval to use the data already collected from all 52 young adults.

Colorado State University
Knowledge to Go Places

Approval Period:	April 01, 2012 through March 31, 2013
Review Type:	FULLBOARD
IRB Number:	00000202
Funding:	National Institute of Health : 79097

Consent to Participate in a Research Study

Colorado State University

TITLE OF STUDY: *Ankle Steadiness, Postural Control, and Physical Frailty*

PRINCIPAL INVESTIGATOR: *Brian L. Tracy, Ph.D. 491-2640*

CO-INVESTIGATOR: *Raoul Reiser II, Ph.D. 491-6958*

WHY AM I BEING INVITED TO TAKE PART IN THIS RESEARCH? *You are a man or woman between the ages of 18-30 or 65-95 years. You either 1) do not report major health problems, <u>or</u> 2) report problems with falling and/or frailty. Our research is looking at the effect of healthy and frail aging, and contributions to the control of muscle force.*

WHO IS DOING THE STUDY? *This research is being performed by Brian Tracy, Ph.D., and Raoul Reiser II, PhD of the Health and Exercise Science Department. Trained graduate students, undergraduate students, research associates, or research assistants are assisting with the research. These studies are paid for by the National Institutes of Health, a part of the US Government.*

WHAT IS THE PURPOSE OF THIS STUDY? *The way in which muscles are controlled by the brain and nerves may change in older people. The effect of vision, mental distraction, and/or vibratory stimuli feedback may be different in young, healthy elderly, and frail elderly, and may be different between muscles. The purpose of the research is to examine these changes and differences in hand, arm, and leg muscles.*

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST? *This whole research project will take place over a period of approximately two years. Your part in this study will take place over five to seven visits over a period of eight weeks.*

WHAT WILL I BE ASKED TO DO? *This consent form applies to a large research project. You are only being asked to participate in parts of the total project. Depending on the part of the research project that you are involved in, you will be asked to participate in some of the following procedures. Many potential procedures are described in the section below. However, the procedures that you will be asked to do <u>for this part of the study</u> have a check mark next to them. The check marks were put there by one of the researchers. A member of the research team will fully explain each checked procedure that applies to your participation.*

______ *You will be asked to answer some questions about your health and exercise to determineif you can participate in the study. 30 minutes)*

______*(your initials)*

______ *If you are in the 65-95 yr-old age group, you will be asked to undergo a brief physical exam by a physician. This test will occur in the Human Performance Clinical/Research Laboratory in the Department of Health and Exercise on the CSU campus. (~ 15 minutes)*

______ *(your initials)*

______ *The fat, muscle, and bone in your body will be measured using an x-ray device (dual-energy x ray absorptiometer) that will scan you from head to toe while you lie quietly on a special table for approximately 10 minutes. The amount of x-ray radiation you will receive is extremely low.*
______*(your initials)*

______ *You will be asked to lightly warm-up your arms and legs with light stretching, simple footwork and slow walking at a comfortable level. (~ 5 minutes)*
______ *(your initials)*

______ *You will be asked to complete brief mental tests of your ability to remember words and numbers on two separate occasions. (~ 20 minutes)*
______*(your initials)*

______ *You will perform a short physical performance test comprised of simple one-legged and two-legged balance tests with your eyes open or closed, rising from a chair five times, and walking a short distance. (~ 20 minutes)*
______*(your initials)*

______ You will be asked to ascend and descend a staircase at a pace comfortable to you. A handrail and research assistant will be within close reach at all times for assistance. (~ 2 minutes)
______ *(your initials)*

______You will undergo clinical examination of the sensory capacity using fine filaments and probes on the skin surface to measure sensory capacity.
______ (your initials)

______*While standing, you will complete two different stepping tests. You will be asked to step as rapidly as possible to the front, side, and rear. (~20 minutes)*
______*(your initials)*

______ *You will perform three reaction time tests with a computer and keyboard. You will respond to either a symbol on the computer screen or a brief sound.(~15 minutes)*
______*(your initials)*

______ *You will perform a mobility test. This will involve rising from a chair, walking 10 feet, turning around, walking back to the chair, and sitting down. This will be repeated three times. (~5 minutes)*
______ *(your initials)*

______ *You will stand next to a wall and reach your arm out as far as you can without moving your feet. This task will be attempted and measured three times. (~2 minutes)*
______*(your initials)*

______ *You will sit in a special chair and perform light and heavy muscle contractions with your hand, arm, thigh and/or ankle muscles while your leg, hips, and shoulders are comfortably secured. (1 – 2.5 hours)*
______*(your initials)*

______ *You will stand as still as possible for 15-60 seconds with your feet together and arms by your side. This will be performed several times in a row with several minutes rest between each trial. During some of the trials you will look forward at a point on a wall in front of you. During some of the trials you will have your eyes closed. During this test you will be standing on a device called a force plate that measures the forces that your feet apply on the surface. (~20 minutes)*
______*(your initials)*

______ *You will stand on the force plates and gently sway or lean forwards and backwards without falling while keeping your feet flat for 60-90 seconds. You will be spotted by a research assistant. (~20 minutes)*
______ *(your initials)*

______ *You will stand in place while keeping your feet flat for approximately a minute on the force plates while a small weight disrupts your stance gently. (~20 minutes)*
______*(your initials)*

______*While performing light and heavy muscle contractions or standing tasks, you may be asked to perform a slightly challenging counting drill out loud during the task. (1-2.5 hours)*
______*(your initials)*

_____ *Sticky electrodes will be placed on the skin over the muscles involved for some of the visits and will remain in place until the end of that visit. Natural oil in the skin will be removed with rubbing alcohol, and the skin will be gently roughened with a fine abrasive paste or cloth.*
______*(your initials)*

______ *An electrode made of hair-sized fine wires will be inserted into your hand, arm, thigh and/or ankle muscle using a small needle. The skin will be thoroughly disinfected, similar to when you get your blood drawn. The needle is sterilized and is the same as the ones used for blood drawing. Either the fine hair size wires or the needle will remain in your muscle for the duration of the visit and then will be removed. Usually there will only be one electrode insertion. However, it is possible that electrodes may need to be inserted 1-5 times in different locations in the muscle. (1-2.5 hours)*
______*(your initials)*

______ *A vibrating device will be placed against leg muscle/tendon for a time period of several seconds up to several minutes, causing a brief muscle contraction.*
______*(your initials)*

______ *An electrical stimulus will be delivered to a nerve or muscle in your leg or arm using a standard stimulator. This may cause a brief muscle contraction.*
______*(your initials)*

ARE THERE REASONS WHY I SHOULD NOT TAKE PART IN THIS STUDY? *If you are not 18-30 or 65-95 years of age, are pregnant, are a regular smoker, or have any diseases that would affect our measurements, we will not be able to include you in the research.*

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS? *(The procedures that apply to your proposed participation are checked)*

- *Health questionnaires* – *There are no known risks associated with answering health questions. All information is kept strictly confidential.*
  ___ *(your initials)*

- *Physical examination* – *There are no known risks associated with a physician- administered physical examination.*
  ___ *(your initials)*

- *Warm-up* – *There are no known risks associated with completing this preventative task. It will be completed at a level comfortable to the subject.*
  ___ *(your initials)*

- *Stair climb task* – *There is a slight risk of falling on the stairs during this test. There will be a research investigator near you for assistance and a handrail within reach at all times. Rest will be given to prevent tiredness.*
  ___ *(your initials)*

- *Brief mental Tests* – *There are no known risks associated with completing these tests. The information is confidential.*
  ___ *(your initials)*

- *Short physical performance test* – *There is a slight risk of falling and potential muscle strain during these tests. A research investigator will be spotting nearby at all times to prevent falls and rest will be given to prevent tiredness.*
  ___*(your initials)*

- *Sensory acuity exam* – *There is no risk associated with this task.*
  ___ *(your initials)*

- *Rapid stepping test* – *There is a slight risk of soreness or muscle strain with these procedures. A researcher will be nearby for safety. Rest will be given to prevent tiredness.*
  ___ *(your initials)*

- *Reaction time* – *There are no known risks associated with the computer reaction time tests.*
  ___ *(your initials)*

- *Mobility test* – *There is a slight risk of falling and injury as a result of rising from a chair and walking a short distance. A research investigator will be nearby to help. Rest will be given to prevent tiredness.*
  ___*(your initials)*

- *Standing reach test* – *There is a slight risk of falling or muscle strain from this test.You will be next to a wall to help keep balance. A research investigator will be next to you for safety.*
  ___ *(your initials)*

- *Muscle contractions* – *There is a slight risk of muscle strain and muscle soreness resulting from brief, light and strong muscle contractions with the hand, arm, thigh and/orankle. Soreness should not last more than two days or affect your normal function.*
  ___ *(your initials)*

- *Postural Standing* – *The risks associated with this balance test include loss of balance with the potential for falling. This risk is extremely low because you will have both feet on the ground and be closely surrounded by a padded handrail and a research assistant.*
  ___ *(your initials)*

- *Postural Sway* – *The risks associated with this balance test include loss of balance with the potential for falling. This risk is extremely low because you will have both feet on the ground and be closely surrounded by a padded handrail and a research assistant.*
  ___ *(your initials)*

- *Perturbed Standing* – *The risks associated with this balance test include loss of balance with the potential for falling. This risk is extremely low because you will have both feet on the ground and you will have a security rail, a research assistant near and a cord attached to the ceiling to prevent you from falling if you lose your balance.*
  ___ *(your initials)*

- *Counting drill* - *There is a minimal risk of feeling anxious while counting and performing muscle contractions or standing. Although, trials will be less than 30 seconds at a time and are not meant to be strenuous. The task will be terminated if you feel uncomfortable.*
  ___ *(your initials)*

- *Sticky electrodes* – *There is no known risk with the preparation or use of sticky electrodes on the surface of the skin.*
  ___ *(your initials)*

- *Fine-wire electrodes* – *There is a risk of discomfort from the needle, temporary soreness in that muscle, and a remote risk of infection. The equipment we use is sterile and only used once and then thrown away. We use special procedures to kill the germs on the skin. In cases where we keep the needle in the muscle during the test, it may cause slightly more discomfort.*
  __ *(your initials)*

- *Vibration of muscle or tendon* – *There is no known risk associated with vibration of your tendon or muscle. The sensation you will feel is similar to what you would feel from a home massage device. The muscle that is vibrated may experience a small involuntary contraction.*
  ___ *(your initials)*

- *Electrical stimulus of nerve or muscle* – *There is no known risk associated with electrical stimulation of nerves or muscle. The device is isolated from dangerous electrical voltages. You will experience a mild sensation of electrical shock in your arm or leg when we stimulate with low levels. When we stimulate with higher levels, you will likely experience abrief but uncomfortable sensation of electrical shock. The electrical stimuli will likely cause an involuntary muscle contraction.*
  ___ *(your initials)*

- *Body composition (DEXA) scan* – *the risks associated with the DEXA are very low.The radiation you will receive is less than 1/3000th of the Food and Drug Administration (FDA) limit for annual exposure. The FDA is a government organization responsible for medical safety. In other words, you could receive 3000 DEXA scans in a single year and still* *not meet the FDA limit for radiation exposure. In this study you will receive one scan. The more radiation you receive over the course of your life, the greater the risk of having cancerous tumors or of inducing changes in genes. The radiation in this study is not expected to greatly increase these risks, but the exact increase in such risks is not known. Women who are pregnant or could be pregnant should receive no unnecessary radiation and should not participate in this study.* _

______ *(your initials)*

- It is not possible to identify all potential risks in research procedures, but the researcher(s) have taken reasonable safeguards to minimize any known and potential, but unknown, risks.

ARE THERE ANY BENEFITS FROM TAKING PART IN THIS STUDY? *There are no direct benefits to you for participating in this study except the health information from the body composition assessment.*

DO I HAVE TO TAKE PART IN THE STUDY? *Your participation in this research is voluntary. If you decide to participate in the study, you may withdraw your consent and stop participating at any time without penalty or loss of benefits to which you are otherwise entitled.*

WHAT WILL IT COST ME TO PARTICIPATE? *There is no cost to you for participating except that associated with your transportation to our facilities.*

WHO WILL SEE THE INFORMATION THAT I GIVE? *We will keep private all research records that identify you, to the extent allowed by law. Your information will be combined with information from other people taking part in the study. When we write about the study to share it with other researchers, we will write about the combined information we have gathered. You will not be identified in these written materials. We may publish the results of this study; however,*

we will keep your name and other identifying information private. We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information, or what that information is. For example, your name will be kept separate from your research records and these two things will be stored in different places under lock and key. You should know, however, that there are some circumstances in which we may have to show your information to other people. For example, the law may require us to show your information to a court, the National Institutes of Health, or to the Human Research Committee at CSU.

CAN MY TAKING PART IN THE STUDY END EARLY? *Your participation in the study could end in the rare event of muscle strain, if you become pregnant, or if you miss an excessive number of appointments.*

WILL I RECEIVE ANY COMPENSATION FOR TAKING PART IN THIS STUDY? For experiments that involve fine wire electrodes, you will be paid $8/hr. Your identity/record of receiving compensation (NOT your data) may be made available to CSU officials for financial audits.

WHAT HAPPENS IF I AM INJURED BECAUSE OF THE RESEARCH? *Please be aware that for this study the University has made special arrangements to provide initial medical coverage for any injuries that are directly related to your participation in this research project. The research project will provide for the coverage of reasonable expenses for emergency medical care related to the treatment of research-related injuries, if necessary.*
LIABILITY:

Because Colorado State University is a publicly-funded, state institution, it may have only limited legal responsibility for injuries incurred as a result of participation in this study under a Colorado law known as the Colorado Governmental Immunity Act (Colorado Revised Statutes, Section 24-10-101, et seq.). In addition, under Colorado law, you must file any claims against the University <u>within 180 days after the date of the injury</u>. In light of these laws, you are encouraged to evaluate your own health and disability insurance to determine whether you are covered for any physical injuries or emotional distresses you might sustain by participating in this research, since it may be necessary for you to rely on your individual coverage for any such injuries. Some health care coverages will not cover research-related expenses. If you sustain injuries, which you believe was caused by Colorado State University or its employees, we advise you to consult an attorney. Questions concerning treatment of subjects' rights may be directed to Janell Barker, Human Research Administrator at 970-491-1655.

WHAT IF I HAVE QUESTIONS? *Before you decide whether to accept this invitation to take part in the study, please ask any questions that might come to mind now. Later, if you have questions about the study, you can contact the investigator, Brian Tracy, Ph.D., at (970)491-2640, or via email at tracybl@cahs.colostate.edu. If you would like to ask a medical doctor about your participation in the study, you may contact ___________________ at ______________. If you have any questions about your rights as a volunteer in this research, contact Janell Barker, Human Research Administrator at (970) 491-1655.*

This consent form was approved by the CSU Institutional Review Board for the protection of human subjects in research on April 1, 2012.

We will give you a copy of this consent form to take with you. Your signature acknowledges that you have read the information stated and willingly sign this consent form. Your signature also acknowledges that you have received, on the date signed, a copy of this document containing 6 pages.

____ _____________________

Signature of person agreeing to take part in the study Date

____

Printed name of person agreeing to take part in the study

____ _____________________

Name of person providing information to participant Date

____

Signature of Research Staff