Summer 2015

The Effect of Posterior Foot Positions on Electromyography, Joint Kinetics and Energetics During a Sit-to-Stand in Young and Older Adults

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THE EFFECT OF POSTERIOR FOOT POSITIONS ON ELECTROMYOGRAPHY, JOINT KINETICS AND ENERGETICS DURING A SIT-TO-STAND IN YOUNG AND OLDER ADULTS

by

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B.S. December 2012, Old Dominion University

A Thesis Submitted to the Faculty of
Old Dominion University in Partial Fulfillment of the
Requirements for the Degree of

MASTER OF SCIENCE in EDUCATION

EXERCISE SCIENCE AND WELLNESS

OLD DOMINION UNIVERSITY
August, 2015

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ABSTRACT

THE EFFECT OF POSTERIOR FOOT POSITIONS ON ELECTROMYOGRAPHY, JOINT KINETICS AND ENERGETICS DURING A SIT-TO-STAND IN YOUNG AND OLDER ADULTS

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The sit-to-stand (STS) movement, defined as standing up from a chair to an upright posture, is a common task performed daily. The inability to accomplish this task may lead to dependence, institutionalization and even death in older adults. A common strategy change for the STS is positioning the feet posteriorly by increasing flexion at the knee joint. This reduces the displacement of the center of mass to the base of support while increasing the joint moment of force at the knee while decreasing it at the hip. The aims of this study were to: 1) examine the joint kinetics, joint energetics and muscle activations of a STS task between young and older adults, and 2) determine the effects of posteriorly placing the feet on joint kinetics, joint energetics and muscle activations in STS.

Twenty participants were recruited for this study. The samples consisted of 10 young participants (age: 22.3±2.06 yr; mass: 70.1±11.7 kg; height: 1.71±0.06 m), and 10 older participants (age: 72.7±5.96 yr; mass: 82.2±13.84 kg; height: 1.73±0.07 m). Both groups were recreationally active, free of lower extremity injury in the past six months and had no history surgery to the lower extremity. Participants were asked to complete five STS with a natural foot position and then with their feet placed 0.10m posterior to the initial position. Three-dimensional marker coordinate data were collected synchronously with three-dimensional force data and surface electromyography (EMG) of 8 lower extremity muscles.

Results indicate that older adults have statistically higher integrated EMG activity in the soleus, rectus femoris and hamstrings muscles along with a statistically higher ankle joint angular impulse. Shifting the feet back produced statistically different values for trunk flexion, time of the rising phase, lower body joint work and hip joint angular impulse. The feet back condition also significantly decreased the integrated EMG magnitude for the tibailis Anterior, vastus lateralis and the hamstring muscles. This study reveals the benefits and limitations to a feet back condition for the STS. Placing the feet back will reduce the time of the rising phase while consequently reducing the magnitude of integrated EMG of the lower body muscles and decreasing the hip joint angular impulse. However, ankle joint angular impulse was increased in older adults with a foot back position. Specific attention should be provided to the individual when applying the feet back condition.

This thesis is dedicated to improving the quality of life for those with limitations in activities of daily living. This is the first step of my mission to improve functional ability of daily tasks throughout the human lifespan.

ACKNOWLEDGEMENTS

Special thanks to the Faculty and graduate students at Old Dominion University along with the staff at Prime Plus who gave their time to make this a truly extraordinary experience.

Dr. Joshua Weinhandl

Dr. Stacie Ringleb

Dr. Matthew Hoch

Dr. Kimberly Baskette

Zachary Sievert

Cody Fontenot

Dr. Jessica Mutchler

Dr. Michael Samaan

William Perez

Rachel Simmons

Chantel Randolph

Peyton Gega

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NOMENCLATURE

BF	Biceps Femoris
COM	Center of Mass
EMG	Surface Electromyography
GAS	Gastrocnemius
GRF	Ground reaction force
Hz	hertz
iEMG	integrated Surface Electromyography
MH	Medial Hamstring
REC	Rectus Femoris
STS	Sit-to-stand
SOL	Soleus
TA	Tibailis Anterior
VGRF	Vertical Ground reaction force
VL	Vastus Lateralis
VM	Vastus Medialis

Chapter 1: Development of the Problem

Background and Rationale

The sit-to-stand (STS) movement (Gross et al., 1998), defined as standing up from a chair to an upright posture, is performed daily. A survey of Dutch adults over the age of 55 reported that 25% of men and 37.4% of women have difficulties with this task. This same survey also indicated severe disability related to muscular weakness, neurological impairments and joint osteoarthritis in 5% of men and 7.8% of women (Odding, 1994). The majority of STS research in older adults mainly examines an age range of 65-85 years (Cheng et al., 2014; Dehail et al., 2007; Gross et al., 1998; Lindemann et al., 2003; Papa and Cappozzo, 2000). The inability to accomplish this task may lead to dependence, institutionalization and even death in older adults (Janssen et al., 2002).

As adults age, there is an expected declined in muscle strength and power (Aniansson et al., 1978). Muscle weakness was simulated by having young participants wear a weight coat during STS (Meijer et al., 2009). The added weight demonstrated limitations in sagittal plane joint energetics and kinetics, with the largest difference at the knee joint (Meijer et al., 2009). The reduced ability in older adults to accomplish physically demanding tasks such as stair climbing and lifting objects is related to muscle weakness (Miszko et al., 2003). As the inability to stand impacts the quality of life and ability to continue independent living, investigating how aging affects STS performance is imperative.

STS necessitates the harmonized movement of linked body segments to effectively displace the body's center of mass (COM) in a horizontal then vertical direction while preserving balance over a small base of support (Tully et al., 2005). The displacement of the COM in STS can be separated into three distinct phases: preparation, rising and stabilization. The preparation phase is the start of the movement before rising. When contact is lost with the seat, vertical ground reaction force (VGRF) is at its max and the rising phase starts and continues throughout full hip and knee extension (Lindemann et al., 2003). The extension of the hips and knees in this phase represents the range of motion and force of the muscles that act as prime movers. Three dimensional motion capture and force plate data during STS can provide valuable insight to this task through the comparison of energetics and kinetics between older and younger adults. Understanding the relationship between these phases and force may reveal how young adults differ from older adults for STS.

Normalized maximum VGRF was employed in able bodied young and older adults along with older adults that are at risk of falling. There is a reduction of maximum VGRF with age and an even larger reduction in older individuals who are at greater risk for falls (Cheng et al., 2014). The ability to understand the changes in VGRF production may be able to identify age related losses in force production in the STS. Measuring VGRF is useful to determine the force generated by the body as a whole; however, employing kinetics at each lower body joint will provide more details on the distribution of force at each lower body joint. Research by Hughes et al. (1996) examined muscle weakness in STS by comparing joint moments of force at the ankle knee and hip in young and older adults. It was reported that knee joint extensors were more than 50% lower in older than in young adults, demonstrating the strength differences between young and older adults (Hughes et al., 1996). Similar to the VGRF, peak moments of force only represent a discreet time point of the entire STS.

Employing energetics in the form of joint work and kinetics as joint angular impulse can reveal cumulative kinetic and energetic effort required to accomplish this task and its distribution at each joint. Differences in ankle, knee and hip angular impulse and work were investigated between a loaded and unloaded condition by having younger adults wear a weight coat during a STS (Meijer et al., 2009). This loaded condition served the purpose to demonstrate muscle weakness which eventually resulted in higher joint work and angular impulse in the knee joint. However, this weakness was the result of a weight coat and not of age related weakness (Meijer et al., 2009). Therefore, differences in lower extremity joint angular impulse and work between young and older adults needs to be investigated. However, this will only evaluate the force production between young and older adults.

The initial configuration of the seated posture should be considered when evaluating the STS motion. Foot positioning affects movement strategies during STS (Janssen et al., 2002). A larger distance of the feet from the chair negatively affects the ease of STS due to the greater challenge in moving the COM horizontally. The ability to simply shift the feet closer allowing less horizontal displacement of the COM can greatly improve STS performance in all people. Specifically when the feet were positioned forward; there were significant increases in movement duration, displacement and velocity of the trunk. Onsets of electromyography (EMG) in STS correlated with peak torque production about the hip, knee and ankle joints depending on the foot position (Hughes et al., 1996; Kawagoe et al., 2000; Khemlani et al., 1999). A foot back position used previously that quantified kinetics and EMG had the participants shift their feet back 0.10 m (Khemlani et al., 1999). The ability to quantify EMG activity throughout the entire STS rather than focusing on single maximal/peak events may reveal functional performance of muscles between young and older adults.

EMG activity was investigated in healthy older women during the sit-to-walk movement (Dehail et al., 2007). The muscles recorded for EMG activity were the tibailis anterior (TA), soleus (SOL), peroneus longus, medial gastrocnemius, biceps femoris (BF), vastus medialis (VM), gluteus maximus, and erector spinae at the level of the L3 vertebra. The maximal activation for TA and erector spinae was observed during the rising phase just after seat off. The activation for the muscles at the hip and the knee were well coordinated as described by a statistical correlation. The gluteus maximus and the SOL were the last muscles to activate at the end of the rising phase during standing and are considered as postural muscles involved with standing stability (Dehail et al., 2007). The early activation of the TA and peroneus longus can be viewed as an anticipated postural adjustment with a purpose of stabilizing the foot before the body moves forward. It is suggested that the lack of initial activation of the TA could exemplify degradation of the STS. It is well known that quadriceps weakness and central activation deficits are related to declines in functional performance for the STS in frail and older adults (Fukagawa et al., 1995). With these limiting factors in mind, muscle activation patterns and movement strategies may be altered during the STS, thus improving the performance of the STS in older adults.

Statement of the Problem

The question remains whether the distribution of joint kinetics, joint energetics and muscle activations will be different in younger adults compared to older adults performing STS. Furthermore, it remains unknown if a posterior foot position will increase the distribution of joint kinetics, joint energetics and muscle activations to the knee joint in younger adults compared to older adults.

Statement of Purpose

The aims of this study are to: 1) examine the joint kinetics, joint energetics and muscle activations of a STS task between young and older adults, and 2) determine the effects of posteriorly placing the feet on joint kinetics, joint energetics and muscle activations in STS.

Research Hypotheses

- There will be significantly lower magnitude of EMG activity in older adults compared to young in all of the muscles.
- There will be significant decreases in hip and ankle joint angular impulse and work and significant increases in the knee joint for the older adults compared to the younger adults.
- There will be a significantly increased knee angular impulse and work with a significant decrease in hip and ankle joint angular impulse and work as the feet are positioned posteriorly.
- With a posterior foot placement, there will be significantly lower magnitude of EMG activity in the biceps Femoris, medial hamstring, soleus, gastrocnemius and

Tibailis anterior while there will be a significantly larger magnitude of EMG activity in the rectus Femoris, Vastus Lateralis, Vastus Medialis.

Independent variables

- Age group (18-35 years for young adults and 65-85 years for older adults)
- Foot Position (Self-selected and 0.10 meters posterior)

Dependent variables

- Kinetic Variables:
  - Rising phase sagittal plane angular impulse of the dominant limb hip, knee and ankle joints
- Energetic Variables:
  - Rising phase sagittal plane work of the dominant limb hip, knee and ankle joints
- Integrated Electromyography (iEMG) of the dominant limb:
  - Tibailis Anterior (TA)
  - Soleus (SOL)
  - Gastrocnemius (GAS)
  - Rectus Femoris (REC)
  - Vastus Lateralis (VL)
  - Vastus Medialis (VM)
  - Biceps Femoris (BF)
  - Medial Hamstring (MH)
  - Time of rising phase
  - Trunk sagittal plane rising phase range of motion

Limitations of the Study

- The STS will be performed in a laboratory setting, not the subject's preferred environment.
- The high-speed cameras will only identify reflective markers placed on the surface of the lower body and trunk segments.
- Wireless EMG electrodes are only placed on surface muscles.

Delimitations of the Study

- Young participant's age range will be 18-35 years.
- Older participant's age range will be 65-85 years.
- Anyone who has difficulty standing will be excluded.
- Any participant who has had lower extremity surgery will be excluded.
- Any participant who has had a lower extremity injury within 6 months will be excluded.
- Older participants who have fallen in the last year.

Assumptions of the Study

- The high-speed cameras and two force plates will be accurately calibrated for each participant throughout the experiments.
- The muscles are properly palpated and prepped to record EMG signals.
- Force plates, will only provide a measurement of ground reaction force under the feet.
- The reflective markers on the surface of the lower body and trunk will represent bone segments.
- Inverse dynamics, link segment model assumptions:

- That all body segments are a perfectly rigid body.
- The center of mass location is a fixed point for each segment.
- Each joint is a frictionless pin joint.
- The moment of inertia remains constant during the movement.

Significance of the Study

Many studies have been performed comparing young and older adults in STS with a particular emphasis on single maximal events by measuring peak torques, powers and muscle activations (Cheng et al., 2014; Gross et al., 1998; Meijer et al., 2009; Rodrigues-de-Paula Goulart and Valls-Solé, 1999). Furthermore, while the effects of a posterior foot placement are understood on STS peak torques and muscle activations (Hughes et al., 1996; Kawagoe et al., 2000; Khemlani et al., 1999), there are no studies known to date comparing posterior foot placements on STS performance in young and older adults. Additionally, few studies have evaluated the total effort in the rising phase by measuring joint angular impulse, joint work, and iEMG in the STS (Davidson et al., 2013; Meijer et al., 2009). The results of this study can help rehabilitation specialists determine appropriate foot positioning strategies to assuage the STS for those with muscle weakness at particular joints.

Operational Definition of Terms

- Joint moment of force, henceforth referred as joint moment, represents the rotational effect of an eccentric force on an object.
- Joint angular impulse represents the effect of a joint moment of force on a system over a specified period of time. It is defined as the integral of the joint moment of force with respect to time.

- Integrated electromyography represents the total accumulated myoelectric activity over a specified period of time. It is defined as the integral of the EMG signal with respect to time.
- Rising phase of the STS begins at peak ground reaction force and ends when ground reaction force returns to the subject's body weight.
- Joint work represents the amount of energy generated or absorbed by a joint moment of force acting through a distance. It is defined as the integral of the joint power curve with respect to time.

Chapter 2: Review of the Literature

Introduction

The STS motion is essential to maintain independent living and a sound quality of life. Many people experience movement difficulties with this task, especially older adults (Alexander et al., 1991; Schultz et al., 1992). There are a variety of mechanisms that contribute to age related losses of STS performance (Janssen et al., 2002).The ability to analyze biomechanical and EMG characteristics of a limited or disabled population can reveal performance limiting factors. Common factors that have been researched in STS include lower body kinetics, energetics, and muscle activity (Cheng et al., 2014; Gross et al., 1998; Lindemann et al., 2003; Meijer et al., 2009; Papa and Cappozzo, 2000; Tully et al., 2005).

A stronger population produces larger values for COM power and joint moments in a STS (Cheng et al., 2014; Gross et al., 1998), but these factors are discrete and specific to a certain time point. To properly describe the kinetics and energetics of the lower body joints for the entire rising phase, understanding joint work and angular impulse in the STS is essential. Work is the integral of the time power curve, which can provide information regarding the total energy about a specific joint. Few studies have investigated the role of individual joint energetics on STS performance (Meijer et al., 2009; Schofield et al., 2013) and no research has been conducted on the elderly. Another important factor to examine between young and older adults is the differences in joint loading throughout the entire STS, which is best described by joint angular impulse. When joint kinetics are unavailable, integrating GRF (iGRF) throughout the STS has proven to be effective in identifying performance limitations in participants with knee osteoarthritis (Davidson et al., 2013) which produces similar muscle weakness seen in older adults (Dillon et al., 2006). Joint kinetics and energetics presents useful insight in STS performance; however, this does not represent the activity of specific muscles involvement. The ability to measure EMG reveals activity of the muscles that produce their respective joint motion.

The majority of the research revealed information on peak activations of lower extremity muscles during the STS (Dehail et al., 2007; Kawagoe et al., 2000; Khemlani et al., 1999; Lindemann et al., 2003; Rodrigues-de-Paula Goulart and Valls-Solé, 1999; Roebroeck et al., 1994). The ability to use iEMG by integrating the linear envelope of the signal is useful for understanding the cumulative muscle activations for the entire STS versus peak activations at certain time points. Research by Davidson et al. (2013) used iEMG in STS to identify quadriceps activations deficits in participants with knee osteoarthritis, similar to weakness in older adults (Fukagawa et al., 1995). Measuring kinetics and EMG in older adults may identify decreased STS performance. However this does not necessarily account for a change in movement strategy to compensate for difficulties in performing STS, which affects the magnitude lower body kinetics, energetics and EMG (Dehail et al., 2007; Papa and Cappozzo, 2000).

One of the most common strategy shifts is placing the feet posterior by increasing knee flexion angle (Janssen et al., 2002). There has been a considerable amount of research regarding posterior foot placements and STS performance (Hughes et al., 1996; Kawagoe et al., 2000; Khemlani et al., 1999); however, there is no research to date that examines joint work and angular impulse along with measures for iEMG. The purpose of this literature review is to examine previous research comparing lower body kinetics, energetics and muscle activations between young and older adults. This review also examines strategy changes including foot positions and its influence on the factors outlined above.

Biomechanical Related Determinants

Understanding the motion and forces that contribute to STS is essential for analysis of several biomechanical factors that are used to evaluate the characteristics of a limited or disabled population and to measure the outcomes of therapy (Guralnik and Winograd, 1994). Common factors that have been researched extensively include kinetic and energetic analysis (Cheng et al., 2014; Davidson et al., 2013; Gross et al., 1998; Lindemann et al., 2003; Meijer et al., 2009; Papa and Cappozzo, 2000).These are the main topics of discussion for this section, which will explore their impact on how STS performance is quantified.

When analyzing the STS, it is important to consider which part of this motion is of particular interest. To clarify the events in STS, this movement can be described by three distinct phases (Lindemann et al. 2003). The preparation phase consists of forward acceleration of the trunk until the hip reaches maximum flexion and the buttocks lose contact with the seat. At the end of this phase, VGRF reaches its peak and the rising phase begins and continues until the VGRF decreases to the participant's body weight. There is a concurrent extension of the hip and knee joints to the standing position. Lastly, the stabilization phase is when postural control is reached when VGRF oscillates within 2.5% of the participant's body weight (Lindemann et al., 2003). The description of these phases applies to all populations to better understand the required forces to produce this movement. However, these phases were defined by healthy young adults and it is clear that aging will lead to a loss of muscle strength and limited STS performance (Riley et al., 1991).

In order to gauge STS performance between young and older adults, research by Papa and Cappozzo (2000) investigated sagittal plane COM motion in able bodied young and older adults by measuring VGRF. The older group adapted a movement strategy in the preparation phase which brought their COM closer to the base of support by flexing the trunk closer to the thighs. The resulting flexion increased velocity gaining a higher momentum to advance to the rising phase. In the rising phase, the older adults maintained their COM over the base of support by keeping the trunk flexed as the COM elevated (Papa and Cappozzo, 2000). Even though analyzing VGRF and COM location provides valuable insight on diffrence in the motion of the STS, the abilty to analyze energetics may reveal useful information on the muscular requirements to perform this task (Cheng et al., 2014; Davidson et al., 2013; Gross et al., 1998; Lindemann et al., 2003; Meijer et al., 2009).

This was the purpose of Cheng et al. (2014) in which lower limb muscle power was investigated. Participants were grouped into those aged between 20 and 30 years, above 65 years without a history of falling (nonfallers) and above 65 with a history of falls in the past 12 months (fallers). A force plate was used to measure VGRF and STS duration. Maximal power was calculated by multiplying the VGRF and the vertical upward velocity of the COM and normalized to each subject's body weight. The results of this study indicate a significantly higher maximal power for the young (9.05± 3.66 W/kg), followed by non-fallers (5.50± 2.02 W/kg) and fallers (3.66 ±1.45 W/kg). There was a significant difference in the max VGRF in young (138.79± 24.20 N/BW) and

(117.51±8.57 N/BW) in old non-fallers. There was a significant difference between the non-fallers and the fallers in the duration of the STS movement (2.74 ± 0.87 s) and (4.27 ± 2.56 s) respectively (Cheng et al., 2014). The maximal VGRF provides us with useful information on the functional performance in the STS; however, this discrete measure only represents a small fraction of the total force required to produce the entire STS.

Understanding the cumulative kinetic forces throughout the motion of the STS represents the loading of the lower body. This topic has been researched by Davidson et al. (2013) in participants undergoing total knee arthroplasties who have reduced muscle strength and power similar to older adults. Compared to a healthy control, the participants performed the five times STS test in late stage osteoarthritis and one month post-surgery. The main results from this study showed significant decrease in hamstring and quadriceps isometric strength in the preoperative group compared to the healthy. Specifically, strength was lower in both muscle groups for the surgical limb. For iGRF, loading was higher in the eccentric and concentric phases in the nonsurgical limb than the surgical limb (Davidson et al., 2013). This increased loading shows how weak population compensates the force distribution to perform a successful STS; however, measuring iGRF only represents the lower limb as a whole with no specific insight to distribution of force to the ankle, knee and hip joints.

There must be adequate force and velocity generated by the muscles of the ankle, knee and hip joints to displace the COM forward and up in STS (Bean et al., 2002). The concept of having a minimum amount of muscle strength is commonly represented with minimum required moment of force at each lower body joint (Gross et al., 1998). A common method to calculate kinetic and energetic variables at each joint is with Newton-

Euler inverse dynamics (Bresler and Frankel, 1950). This approach has shown to be an effective predictor of joint forces that is non invasive to the research participant in STS (Meijer et al., 2009; Tully et al., 2005). This application requires a combination of measured GRF and kinematic variables. Segment lengths can be derived from distances between the relevant markers placed on the surface of specified joints. Based on the segment lengths and a regression model from Dempster (1955), estimations can be made for segment masses, moments of inertia and the locations of the centers of masses. The inverse dynamics approach will yield all the internal forces acting about the joint. This includes the moment of force attributable to joint friction, ligaments, muscles, and structural constraint.

A good example of the appliction of inverse dynamincs is demonstrated by the work of Gross et al. (1998) in which muscle strength was evaluated in the STS comparing healthy young and elderly women. What seperates this research from Papa and Cappozzo (2000) is the addition of measured forces and kinematics of the lower body segments that make the ankle, knee and hip joints. Inverse dynamics was applied to determine the moments of force at hip, knee and ankle joints and how the magnitude of each joint contributes to STS. Understanding the force distribution amongst these joints can identify performance limiting factors of STS in older adults compared to young adults (Gross et al., 1998). The results from Gross et al. (1998) revealed ankle dorsiflexion, hip and knee extension moments of force peaked near the time that participants lost contact with the chair at the beginning of the rising phase. Although not statistically significant, the peak moment of force in the hip was higher in older adults while the young adults had higher moments of force in the knee and ankle (Gross et al., 1998). Research by Hughes et al. (1996) examined muscle weakness in STS by comparing joint moments of force in young and older adults. It was reported that knee joint extensors were more than 50% lower in older than in young adults, demonstrating the strength differences between young and older adults (Hughes et al., 1996). Similar to the VGRF, peak moments of force only represent a discreet time point of the entire STS. Analyzing joint work and angular impulse with secondary calculations from inverse dynamics may reveal the joints total energetic requirements and loading charactersitics throughout the entire STS.

The application of joint work and angular impulse is described by the work of Meijer et al. (2009) comparing young healthy participants STS performance with and without a weight coat to simulate muscle weakness. Inverse dynamics was applied to calculate sagittal plane moments of force at each lower extremity joint and secondary calculations were made for peak power, total work and angular impulse at each joint. With the addition of the weight coat that represents 45% of the participants body weight, values for peak joint powers and moments of force showed significant increases at the knee joint and subsequent decreases at the ankle and hip joints. There was a shift of energetic dominance and loading with increases at the knee joint and corresponding decreases in the hip and ankle joints. The increases in energy and loading at the knee joint emphasizes the dominance of the muscles responsible for knee extension (Meijer et al., 2009). Joint kinetic and energetic values are useful in identifying how much effort is needed to successfully perform STS, but this only reflects general muscle groups that produce the action about the respective joint. Understanding the activity patterns of specific muscles may reveal how joint kinetics and energetics is distributed throughout the lower limb.

Electromyography Related Determinants

Quantifying muscle activity in STS by recording EMG with surface electrodes for individual lower extremity muscles is a highly researched topic (Davidson et al., 2013; Rodrigues-de-Paula & Valls-Sole, 1999; Roebroeck et al., 1994). When kinematic datum is synchronized with EMG signals, these signals provide insight to timing of muscles activation, the force/EMG signal relationship, and fatigue index (De Luca et al., 1997). This section will explore previous research with EMG in STS performance. The main topics discussed will elaborate on the relationships of joint motion and moments of force with muscle activity patterns and iEMG in young and older adults.

Work by Roebroeck et al. (1994) investigated sagittal plane kinematics, GRF, and muscle activity of nine leg muscles in young healthy adults. In order to gauge the amplitude of muscle activation, EMG signals for the STS were normalized to the EMG signal of maximal voluntary isometric contractions (MVIC) that were performed before the STS trials. In the preparation phase the hip, knee and ankle moments of force rose, and increased extensively in the rising phase (Roebroeck et al., 1994). The TA was active before seat-off, but the largest increase of EMG activity at seat-off was found for the knee extensors (VM and REC) reaching 50-80% of the MVIC. The maximal extension moments of force of all 3 joints occurred just after seat-off at 40% of the STS cycle. The EMG activity in this phase displayed a modest increase of activity for gluteus maximus and hamstrings, followed by a continuous level of activity for these muscles for the rest of the STS cycle. There was a subsequent increase in EMG activity for the quadriceps, but the biarticular REC was much lower compared to VL and VF. In the stabilization phases, hip and knee moments of force decreased back to zero whereas the ankle moment of force remained active near its peak. During this phase, there was a decrease in activity for the hip extensors. The SOL and GAS muscles showed increased EMG activity from the start of the STS and remained active through the stabilization phase. This experiment revealed the magnitude and onset of the joint moments of force for the hip, knee and ankle follow a similar pattern to the recorded activations of the muscles that contribute to the action (Roebroeck et al., 1994). However, this is the results of young healthy adults and it is known that older adults experience muscle weakness that affects STS performance (Papa and Cappozzo, 1999; Riley et al., 1991).

To fully understand the differences EMG activity in young and older adults in STS, research by Dehail et al. (2007) investigated EMG activity in healthy older women in sit-to-walk movement. Compared with the STS phase descriptions, in rising phase for sit-to-walk, the trunk stays flexed longer at seat off in order to increase the forward body position for the transfer of gait initiation. The muscles recorded for EMG activity were the TA, SOL, peroneus longus, medial GAS, BF, VM, gluteus maximus, and erector spinae at the level of the L3 vertebra. The maximal activation for TA and erector spinae was observed during the rising phase just after seat off. In 62.5% of the cases, the TA was the first muscle activated compared with the peroneus longus. The activation for the muscles at the hip and the knee were well coordinated as described by a statistical correlation. In the beginning of the STS before seat off, the BF activation was correlated with the gluteus medius ($r = 0.74$; $p < 0.001$) and the VM ($r = 0.65$; $p < 0.001$). The gluteus maximus and the SOL were the last muscles to activate at the end of the rising phase during standing. The SOL and Gluteus Maximus are considered as postural muscles involved with standing stability maintaining the center of pressure within the base of the support (Dehail et al., 2007).The early activation of the TA and peroneus longus can be viewed as an anticipated postural adjustment with a purpose of stabilizing the foot before the body moves forward. It is suggested that the lack of initial activation of the TA could exemplify degradation of the STS. This study demonstrates that older adults compensate for muscle weakness by alter movement strategies and peak muscle activation patterns in STS (Dehail et al., 2007). However, peak activations only represent a discrete time point in the entire STS motion. Applying iEMG will characterize the total accumulated myoelectric activity across the STS phases.

Understanding iEMG in STS was the goal of Davidson et al. (2013) in participants with knee osteoarthritis, that demonstrate lower extremity muscle weakness similar to older adults (Brown et al., 1995). Research by Davidson et al. (2013) examined GRF and EMG in the five times STS test between healthy participants, participants with late stage knee osteoarthritis and 1 month post-surgery. For this study, bilateral surface EMG recorded the activity of the VL and BF. The five repetitions were calculated as an average then separated into sit to stand and stand to sit phases. Muscle activity was quantified for each phase by iEMG. The iEMG characterizes a distribution of recruitment across the phase, not necessarily the level of recruitment. The results of this study show the healthy participants exhibiting a higher iEMG in the VL over the BF. Before surgery, participants with knee osteoarthritis displayed higher iEMG values in the nonsurgical limb than the weak limb. For the weak limb, the BF displayed higher values than the VL while the nonsurgical limb had about the same level of iEMG activity. Post-surgery, the non-surgical limb had similar values between the muscle groups but was slightly higher in the BF. The weak limb decreased the overall iEMG activity in both muscles but the BF remained higher than the VL. This study directly compares the results iEMG activity in a strong and weak limb (Davidson et al., 2013) and is useful for understanding what may happen when comparing iEMG in young and older adults. Understanding EMG may be useful in identifying age related differences in the activity of specific muscles. However older adults are known to change movement strategy to compensate for difficulties in performing STS, which affects the magnitude lower body kinetics, energetics and EMG (Dehail et al., 2007; Papa and Cappozzo, 1999). Understanding the changes that occur with altered movement strategy in the distribution of muscle activity, joint kinetics and energetics may reveal new insight on how we perform the STS.

Strategy Related Determinants by Foot Positioning

It is shown that employing a strategy change will ease STS in a weaker population including older adults (Dehail et al., 2007; Papa and Cappozzo, 1999). The goal for a strategy change is to limit sagittal plane knee and hip range of motion which reduces the knee and hip extension moments of force while increasing the distribution of muscle activity is between quadriceps and hamstrings (Benedetti et al., 2003). A review article by Janssen et al. (2002) examined performance related determinants to STS. The main topics highlighted in this review were related to the participants, chair properties and movement strategy. The most popular movement strategy topics include speed of the movement, foot positioning, and trunk position (Janssen et al., 2002). Differences in trunk movement in STS have been identified in previous sections of this literature review (Dehail et al., 2007; Papa and Cappozzo, 1999); however, understanding the influence of foot positions can have a substantial impact on STS performance. This section will emphasize the effects of seat height and foot placement on joint kinetics and EMG in young adults (Kawagoe et al., 2000; Khemlani et al., 1999) and older adults (Hughes et al., 1996).

To demonstrate the effect of seat height on joint moments and peak muscle activations, research by Kawagoe et al. (2000) analyzed the effects of foot placement with varying chair height in young participants. The chair heights were organized in 30, 40 and 50 centimeters. The lower the seat height produced an anterior foot placement while the higher height produced a posterior position. Muscle activity was recorded for the TA, GAS, REC and the hamstrings. With the feet in an anterior position, the overall activity increased in the TA, REC and hamstrings with the peaks occurring simultaneously after seat off in the beginning of the rising phase and remained highly active until standing. With the anterior position, the knee joint produced a flexion moment and then became an extension moment when the hips reached its maximal extension moment. When the feet were placed posteriorly, the activity of the TA increased more prior to seat off with a decrease in the rising phase. The REC experienced its maximal activation at the same time but the activity decreased more dramatically in the rising phase. The hamstrings reached its maximal activation later in the rising phase with a posterior foot placement. The GAS was relatively unaffected by foot position and saw its peak activation after standing in the stabilization phase. With a posterior placement, both the knee and the hip experienced their peak extension moment at about the same time. Compared to the anterior placement, there was dramatic reduction in the magnitude of the hip joint extension moment and a subsequent increase in the knee joint extension moment. This study revealed how seat height and corresponding foot placement distributes joint moments and peak muscle activations in a young healthy population (Kawagoe et al., 2000). However, it is unclear how changes seat height and foot placement affects performance in older adults.

A similar experimental design by Hughes et al. (1996) evaluated STS in functionally impaired older adults compared to young healthy adults. Functional impairment for the STS was defined as the inability to rise from a 0.33 meter high chair. The knee joint moment was calculated at three chair heights; 0.58 meters, knee height and the lowest chair height the impaired older adults could achieve or 0.33 meters for the young group. For both groups, decreases in chair height increased the required knee joint moment. The young participants produced a significantly greater knee extension moment for the knee height chair versus the older adults. However, the maximum required knee joint moments did not have a significant difference between the two groups in the lowest chair height. This study shows that young adults produce higher knee joint torques at a normalized seat height compared to the older adults, suggesting increased muscular strength (Hughes et al., 1996). This was the result of changing seat height and it is unclear if these changes in height resulted in anterior or posterior foot placements.

To fully understand the effect of a posterior foot placement independent of changes in seat height, research by Khemlani et al. (1999) investigated joint torques and EMG variables in which participants placed their feet forward and then shifting their feet posterior by 0.10 meters by flexing the knee joint without changing the seat height. Nine male participants had EMG signals recorded on the right leg for the REC, VL, BF, TA, GAS and SOL. Simultaneously, GRF and a 2 dimensional video recoded sagittal plane kinetics and kinematics. When the feet were forward, the movement duration and the rising phase were significantly longer than when the feet were placed posterior. Peak extensor moments of force at the hip and ankle occurred just after seat off in the early portion of the rising phase. For the peak ankle moment of force, the amplitude was significantly greater when the feet were posterior. The peak hip moment of force was similar through both conditions. The knee moment was inconsistent in the study and was not used for statistical analysis. However, the knee joint moments were predominantly flexor throughout the STS with a posterior foot placement in 93% of the trials and 42% of the trials with an anterior placement. Positioning the feet posterior significantly increased the amplitude of the peak support moment of force.

The temporal relationship with EMG onsets revealed interesting information on foot position. In the preparation phase before contact was lost from the seat, the TA was the first to activate followed by the RF, BF and VL. The GAS and SOL were activated in the rising phases after the thighs left the seat. Between foot placements, the TA activated significantly earlier when the feet were positioned posterior compared with the forward position. When the feet were placed forward the GAS and SOL activated significantly earlier than in the back position. The relationship between kinetics and EMG activity are the result of a longer time to reach the peak extensor force when the feet were positioned forward. When the feet were forward, the BF, GAS and SOL activated significantly earlier than the time at which peak moment occurred (Khemlani et al., 1999). It can be inferred from the findings from Kawagoe et al. (2000) and Khemlani et al. (1999) that populations with limited lower limb muscle force production should consider adopting a feet back strategy for STS. However, this section only displayed peaks for joint moments muscle activations which occurred at certain time points in the STS. It is unclear how the total energetic, kinetic and muscle activations will respond to a posterior foot placement in STS.

Conclusion

There is a growing interest in movement studies for the biomechanics and EMG of STS between young and older adults. Standing up is a considerable challenge to specific populations, including individuals with osteoarthritis, neurological impairments, and in the elderly (Khemlani et al., 1999). There are several biomechanical variables of STS that need to be examined and clarified in order to fully understand performance limiting factors in STS. When comparing young and older adults in STS, older adult's exhibit reduced muscle strength resulting in decreased peak moments of force and a longer movement time (Hughes et al., 1996). A longer movement time was demonstrated by the work by Meijer et al. (2009) in which muscle weakness was simulated by adding a weight coat to participants which increased joint work and angular impulse at the knee. Even though increased body weight does not necessarily occur with aging, similar strength deficits are apparent in older adults which may produce similar joint loading characteristics. Evaluating iEMG along with joint work and angular impulse in persons with limitations in STS can reveal the distribution total muscle activity along with joint loading and the energy produced at the joint (Davidson et al., 2013; Meijer et al.,2009).

The results from Khemlani et al. (1999) suggest placing the feet posterior before the movement of STS reduces the time of the movement and the hip joint moment of force. There is a discrepancy of the initial configuration of the knee joint in STS with lowest reported knee joint angle of 90-95 degrees was employed by (Cheng et al., 2014;

Dehail et al., 1999 & Kerr et al., 1997; Yu et al., 1999) while others used knee angles of 100-110 (Roebroeck et al., 1994; Tully et al., 2005; Worsley et al., 2011). Knowing that shifting the feet back a small distance of 0.10 meters will increase knee joint moments and alter the lower body muscle activation patterns (Khemlani et al., 1999), comparing a natural self-selected foot positions with a posterior foot position should be investigated. Therefore, in order to optimize STS performance in the elderly by maximizing work of the muscles and minimizing joint loading, research on multiple posterior foot placements on STS is necessary.

Chapter 3: Methods

Subjects

Twenty participants were recruited for this study. The samples consisted of 10 young participants (age: 22.3±2.06 yr; mass: 70.1±11.7 kg; height: 1.71±0.06 m), and 10 older participants (age: 72.7±5.96 yr; mass: 82.2±13.84 kg; height: 1.73±0.07 m). Both groups were recreationally active, free of lower extremity injury in the past six months and had no history surgery to the lower extremity. Recreationally active was defined as participating in exercise three times a week for at least 30 minutes. The participants did not have recent lower extremity pain. All participants were capable of standing up and walking without assistance. The research was approved by the Institutional Review Board of Old Dominion University (Appendix A).

To ensure all participants met inclusionary criteria a series of questionnaires were completed. The first was the Disablement in the Physically Active scale (Vela & Denegar, 2010) (Appendix B). Each participant also recorded their physical activity on the (Godin Leisure time Exercise questionnaire) (Godin & Shepard, 1997) which includes the average intensity of exercise along with the frequency of physical activity (Appendix C). The Western Ontario and McMaster Universities Osteoarthritis index (WOMAC) was used to determine severity of disability from osteoarthritis with maximum scores of 6.5 for pain, 4.0 for stiffness and 25.0 for physical function (Hughes et al., 2004) (Appendix D). The Lower Extremity Functional Scale (LEFS) (Binkley et al., 1999) was also collected to determine severity of disability from osteoarthritis (Appendix E). Finally, a falls risk assessment questionnaire was collected for the older participants with a maximum allowable score of 3 out of 13 possible points based off the inclusionary criteria for an exercise study for older adults (Rubenstein et al., 2011) (Appendix F).

Experimental Protocol

All testing was performed in the Neuromechanics Lab (ODU, Norfolk, VA, USA). Prior to the screening questionnaires and experimental protocol, written informed consent was obtained for each participant. Participants were required to restrain from performing lower extremity tasks outside normal daily activities, in particular exercises that elicit lower extremities soreness for 48 hours prior to data collection. To ensure this, the participants were informed to be rested for 48 hours and asked not to physically overexert themselves in this time period. Each participant wore spandex shorts that reveal the knee joint and a thin shirt that was tucked into the spandex shorts. All participants were given a pair of lab shoes in their preferred size (Air Max Glide, Nike, Beaverton, OR, USA). The participants then had their standing height and weight measured.

Eight EMG surface electrodes were placed on the participants' muscle bellies of the dominant limb (Figure 1). Before placement of the electrodes, hair was removed from the skin at the electrode placement site. The electrode placement site was abraded to remove epithelial cells and then cleaned with alcohol. These muscles included for electrode placement were; TA, SOL, medial GAS, REC, VL, VM, BF and MH. The electrodes were placed on the belly of each muscle based on the guidelines proposed by Criswell (2010).

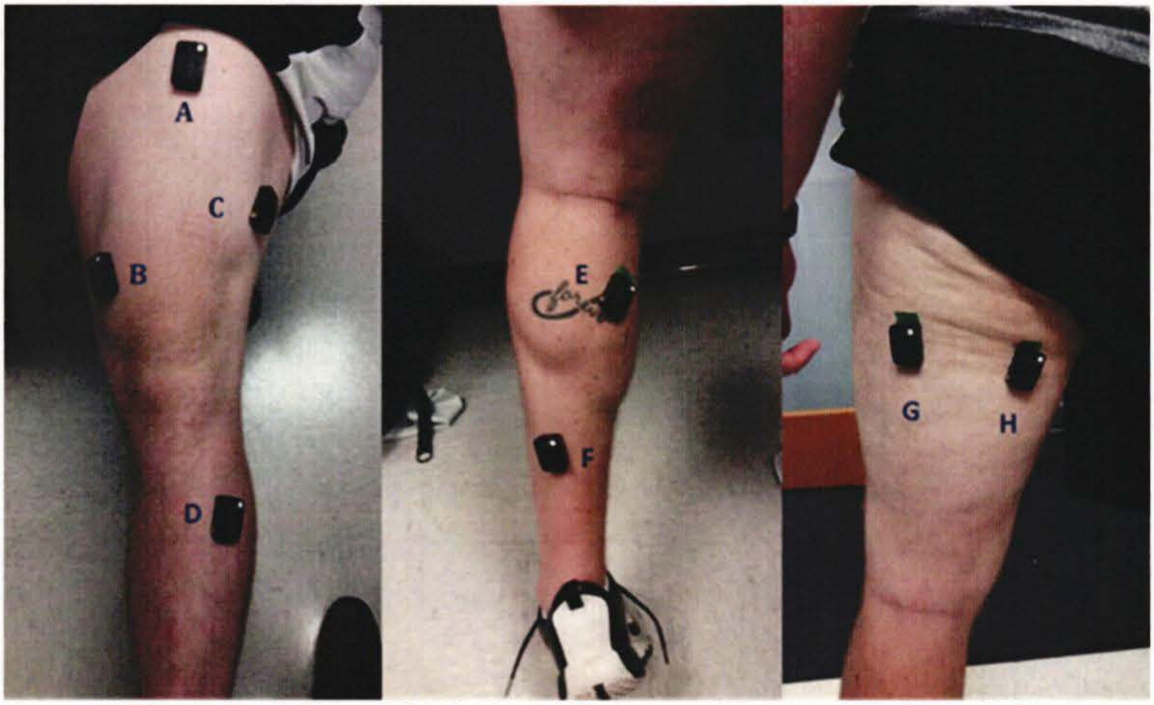

Figure 1. Electrode placement for the lower body muscles. A= rectus femoris, B= vastus medialis, C= vastus lateralis, D= tibailis anterior, E= medial gastrocnemius, F= soleus, G= biceps femoris, H= medial hamstring

The TA had an electrode placed perpendicular to the muscle fibers lateral from the tibia and one third of the distance inferior from the knee. The medial GA had an electrode placed at an oblique angle perpendicular to the muscle fibers, medial to the midline. The SOL had an electrode placement perpendicular to the muscle fibers that are inferior and lateral to the belly of the GA. The REC had an electrode placed halfway between the anterior inferior ridge of the iliac spine and the patella perpendicular to the muscle fibers. The VL had an electrode placed at an oblique angle on the muscle belly perpendicular to the muscle fibers, lateral from the midline of the thigh. The VM had an electrode placed at an oblique angle on the muscle belly perpendicular to the muscle fibers above the knee cap, medial from the midline. The BF and MH had an electrode placed parallel on each muscle belly perpendicular to the muscle fibers half way between the gluteal fold and the posterior knee. After placement, each electrode was secured with pre wrap (Mueller Sports Medicine, Prairie du sac, WI) and then wrapped with athletic tape (Collins Sports Medicine, Raynham, MA). Two elastic Velcro straps (McDavid, Woodridge, IL, USA) were placed on the dominant limb at middle of the thigh and the middle of the shank. These straps provided extra compression against the electrodes on the RF, BF and MH. At this point the electrodes were secured and the participant then performed MVIC for each muscle.

The participant had 12 retro reflective markers placed on anatomical landmarks of the dominant limb which included: the first and fifth metatarsophalangeal joints, the lateral and medial malleoli, and the lateral and medial epicondyles of the femur. There were 4 additional retro reflective markers placed on the greater trochanters and the most superior aspect of the iliac crests. After the placement of the anatomical markers, the two elastic Velcro straps previously placed at middle of the thigh and the middle of the shank securely held tracking plates made from semi-rigid, molded Orthoplast (Johnson & Johnson, Raynham, MA, USA). Each plate had four retro reflective tracking markers. Tracking plates were also placed on the heel of the shoe and on an elastic Velcro strap around the waist on the posterior pelvis at the level of the posterior superior iliac spine. Participants stood on the laboratory force plates with one foot on each plate while maintaining an upright posture with their arms crossed over their chest. The participants held this position while a static calibration trial was recorded. After the calibration trial was completed the anatomical landmark markers were removed.

The participants sat comfortably on the edge of a box adjusted to their lower leg height (Hughes et al., 1996) with one foot on each force plate (Figure 2). The participant then stood from a seated position with the arms crossed over the shoulders and with a self-selected distance between the feet that was kept constant by marking initial foot position on the force plate with tape. The condition of foot positions follows: The first position was self-selected and the second position was 0.10 meters posterior to the self-selected position (Khemlani et al., 1999; Kwong et al., 2014). Each foot position was marked with a piece of tape to ensure consistency in foot placement throughout the trials and the feet remained flat on the force plate for the entire STS. The participant performed five trials of each condition, standing up at a self-selected speed.

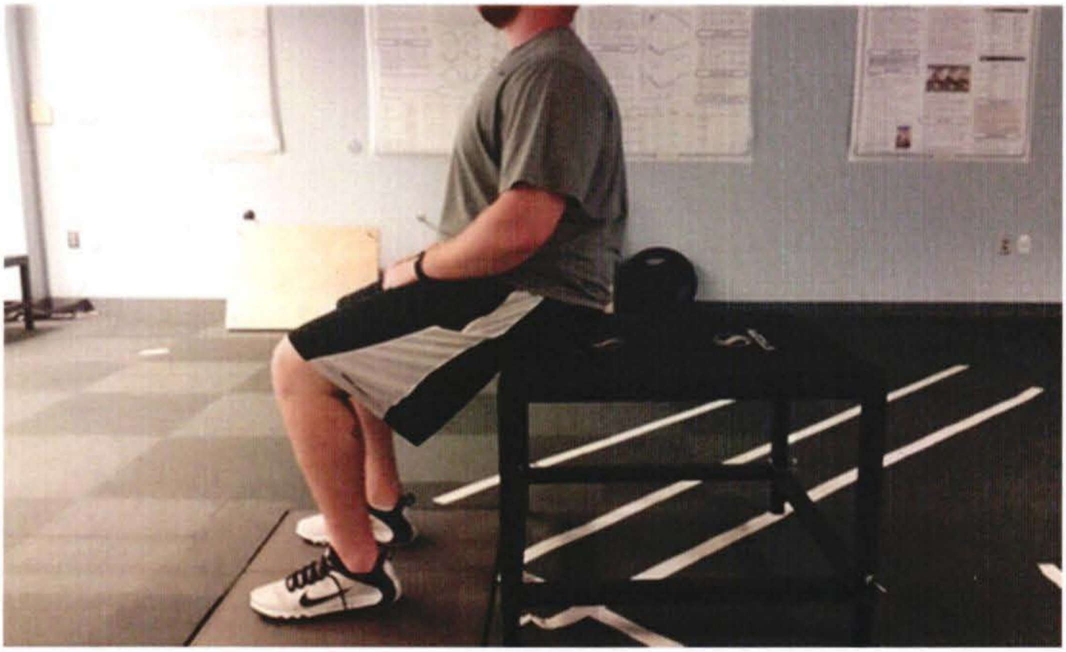

Figure 2. Height adjustable seat to the participant's knee joint.

Instrumentation

Three-dimensional plane marker coordinate data was collected at 200 Hz with an 8-camera motion analysis system (Vicon, Oxford, UK). Force data was collected at 2000 Hz with two Bertec force plates (Bertec Co., Columbus, OH). Data for EMG was collected at 2000 Hz with a Trigno 16 channel wireless surface electrode electromagnetic system (Delsys Inc., Boston, MA). Marker coordinate, GRF, and EMG data was captured synchronously using the Vicon Nexus software package.

Data Analysis

A five segment kinematic model made of trunk, pelvis, thigh, shank and foot was created from the standing calibration trial (Weinhandl and O'Connor, 2010). Marker and GRF data were reduced with Visual 3D (v5.00, C-Motion Inc., Rockville, MD) by applying a low-pass fourth order Butterworth zero lag filter with 12 Hz and 50 Hz cut off frequencies, respectively. Three-dimensional ankle, knee, hip angles were calculated using a joint coordinate system approach (Grood and Suntay, 1983) although only data in the sagittal plane was extracted for analysis. The ankle joint center was defined as the midpoint between the lateral and medial malleoli. The knee joint center was defined as the midpoint between the lateral and medial markers on the condyles of the femur (Grood and Suntay, 1983).The hip joint center was defined as 25% of the distance from the ipsilateral to the contralateral greater trochanter (Weinhandl and O'Connor, 2010). Trunk flexion range was calculated by finding the difference in segment angle from the beginning and the end of the rising phase.

The x-axis (flexion and extension) was defined as the vector between the markers placed on each iliac crest for the pelvis, medial to lateral epicondyles for the thigh and medial to lateral malleoli markers for the shank and foot. The frontal plane was defined as the plane of best fit between the iliac crest and greater trochanter markers for the pelvis, the x-axis and the hip joint center for the thigh, the epicondyles and malleoli markers for the shank and the malleoli markers and the metatarsal markers for the foot. The y-axis (adduction and abduction) for each segment was defined as the vector orthogonal to the frontal plane for each respective segment. The z-axis (internal and external rotation) for each segment was defined as the vector orthogonal to the x-and-y axes.

Three-dimensional joint moments were calculated using a Newton-Euler approach (Bresler and Frankel, 1950) with estimated lower limb segment parameters from Dempster (1955) and were reported in the distal segment reference frame. Visual 3D (v5.00, C-Motion Inc., Rockville, MD) was used to extract work and angular impulse for the ankle, knee and hip joints in the sagittal plane.

Matlab (MathWorks, Natick, MA) was used to post process EMG data. The raw EMG data was pre-amplified and high-pass filtered using fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 10Hz to remove movement artifact. The signal was full-wave rectified and then it was low pass filtered with a cutoff frequency of 5 Hz to create a linear envelope and normalized to the MVICs of the individual muscles. The linear envelope was then integrated over duration of the rising phase (Criswell, 2010). The rising phase (Figure 3.) was defined as when VGRF reaches its peak at seat off and continues until the VGRF decreases to the participant's body weight (Lindemann et al. 2003).

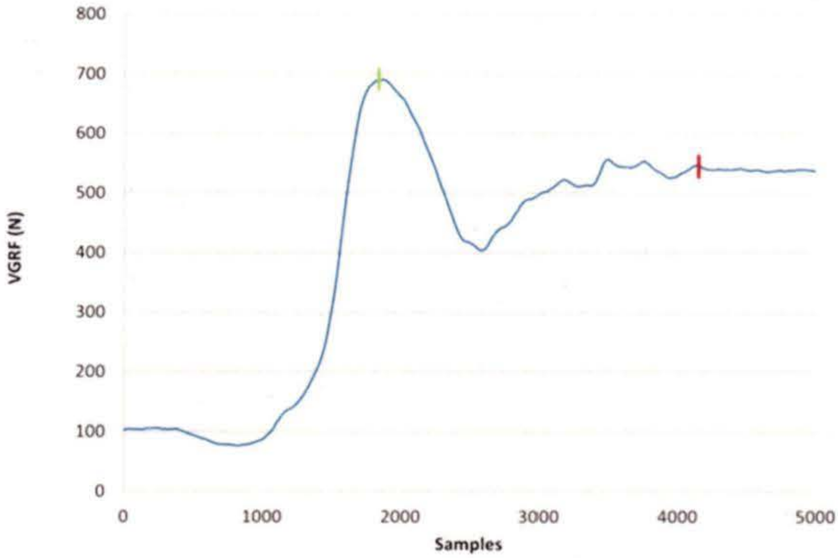

Figure 3. Vertical ground reaction (VGRF) force for a Sit-to-Stand trial. The green and red lines represent the beginning and of the rising phase, respectively.

Statistical Analysis

Descriptive statistics were calculated to determine normality, kurtosis and skewness. Separate 2 x 2 repeated measures ANOVAs examined differences in age (young, old) and foot position (anterior, posterior) for each dependent variable. In the presence of a significant interaction, post-hoc independent and paired t-tests were used for further statistical evaluation. The alpha level for all statistical analyses was set at $p \leq 0.05$. All statistical analyses were conducted using SPSS (v21.0, SPSS Inc., Armonk, NY).

Chapter 4: Results

Questionnaire Results

Twenty individuals were recruited to participate in this study (young group n=10; old group n=10) with an equal amount of men and women in each group. Not all samples were included for statistical analyses; the BF electrode did not record for one young male and the MH did not record for one older male. The TA, SOL, GAS, REC, VL, VM were not usable for a second older male participant; therefore, those muscles were excluded for that participant. The falls risk assessment for the older adults had an average score of 1.19 ± 0.9 out of a maximum allowable score of 3. The WOMAC assessment for the older adults had an average score of 4.9 ± 3.78 out of a maximum allowable score of 35.5. Although not used for exclusionary criteria, the LEFS assessment for the older adults had an average score of 13.6 ± 12.11 out of a maximum score of 80. For both young and older adults, the score for health history was 16.6 ± 0.96 for young and 27.2 ± 8.89 out of a maximum score of 80. For the physical activity level questionnaire, a minimum allowable score of 10 indicating 30 minutes of physical activity a week was collected with an average score of 153.5 ± 131.4 for young and 42.5 ± 35.07 older adults.

Trunk Flexion and Rising Time

For the range of trunk flexion (Figure 4) in the rising phase, there was not a significant interaction (p=0.94) (Table 1). However, there was a significant effect of foot position (p<0.01) as the posterior foot position resulted in a 5.71 degree reduction in trunk flexion when compared to the naturally selected foot position. There was not a significant effect for age (p=0.46). The rising time experienced significant results from age but not foot position. For the rising time of STS in rising phase, there was not a significant interaction (p=0.216). However, there was a significant effect of foot position (p=0.037) as the posterior foot position resulted in a 0.06 second reduction in time when compared to the naturally selected foot position. There was a significant effect for age (p=0.012) as the young group resulted in a 0.31 second reduction in time when compared to the old group.

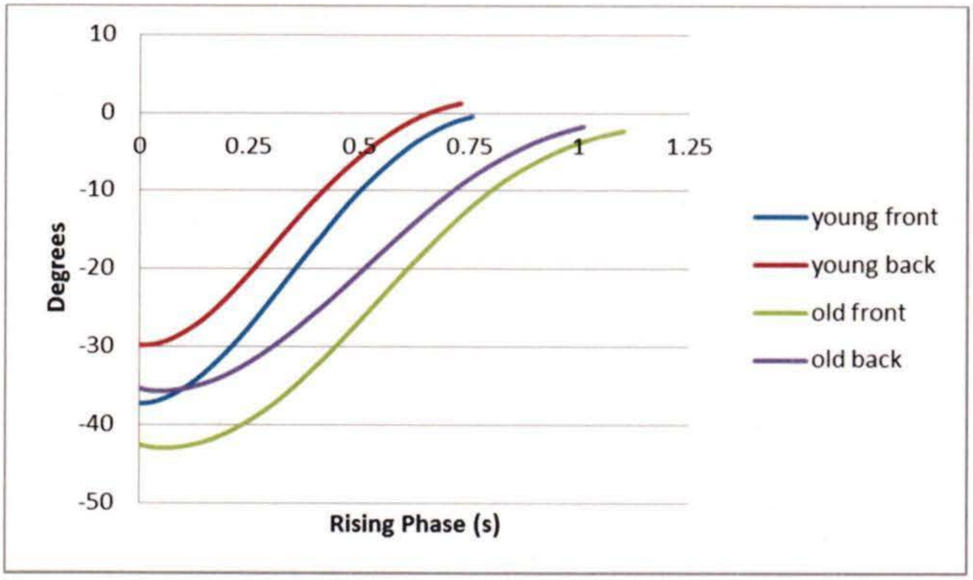

Figure 4. Average trunk flexion angle (degrees) in the rising phase

Table 1. Group means and standard deviations and interactions for trunk flexion, time, for energetic and kinetic dependent variables for the rising phase.

	Young		Old	
	Front	Back	Front	Back
Trunk flexion * (Degrees)	36.9(11.3)	31.2(11.2)	40.4 (9.4)	34.8(9.9)
Time*† (Seconds)	0.756(0.092)	0.730(0.128)	1.101(0.326)	1.009(0.363)
Joint angular impulse (Nms/kg)				
Ankle †	-0.086(0.033)	-0.102(0.037)	-0.167(0.078)	-0.168(0.104)
Knee	0.259(0.066)	0.249(0.060)	0.309(0.133)	0.295(0.152)
Hip *	-0.211(0.066)	-0.158(0.063)	-0.262(0.094)	-0.205(0.085)
Joint Work (J/kg)				
Ankle *	0.033(0.017)	0.045(0.023)	0.042(0.013)	0.053(0.024)
Knee *	0.479(0.127)	0.509(0.126)	0.407(0.169)	0.450(0.207)
Hip *	0.419(0.132)	0.323(0.133)	0.393(0.133)	0.303(0.086)

† Significant main effect for age ($p<0.05$)
*Significant main effect for foot position ($p<0.05$)
ǂ Significant age x foot position interaction ($p<0.05$)

Joint Work

For ankle work (Figure 5), there was not a significant interaction (p=0.901). However, there was a significant effect of foot position (p=0.012) as the posterior foot position resulted in a 0.01 J/kg increase in work when compared to the naturally selected foot position. On the other hand, there was not a significant effect for age (p=0.306).

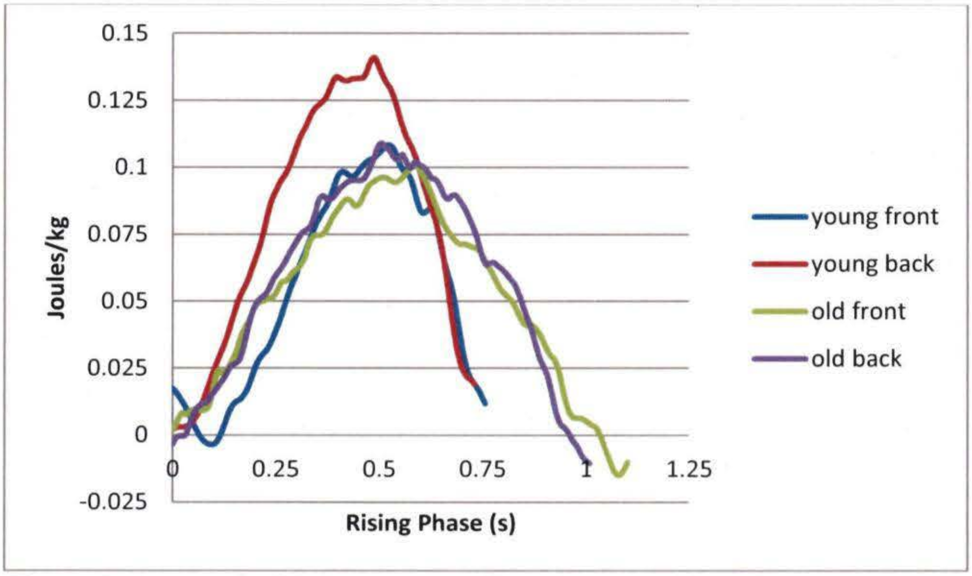

Figure 5. Average ankle joint power (W/kg) in the rising phase. Area under the power-time curve indicates joint work (J/kg).

Similar results from the ankle were seen for knee work (Figure 6). Although there was not a significant interaction (p=0.628), there was a significant effect of foot position (p=0.034) as the posterior foot position resulted in a 0.04 J/kg increase in work when compared to the naturally selected foot position. There was not a significant effect for age (p=0.364).

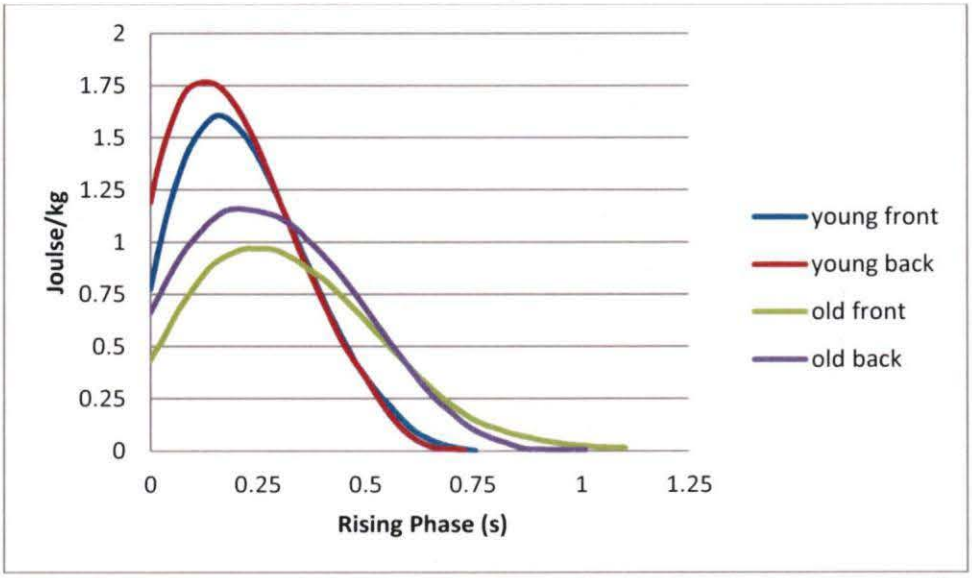

Figure 6. Average knee joint power (W/kg) in the rising phase.
Area under the power time curve indicates joint work (J/kg).

Hip work (Figure 7) experienced similar results to the work of the ankle and knee joints. There was not a significant interaction (p=0.838); however, there was a significant effect of foot position (p<0.01) as the posterior foot position resulted in a 0.09 J/kg decrease in work when compared to the naturally selected foot position. There was not a significant effect for age (p=0.364).

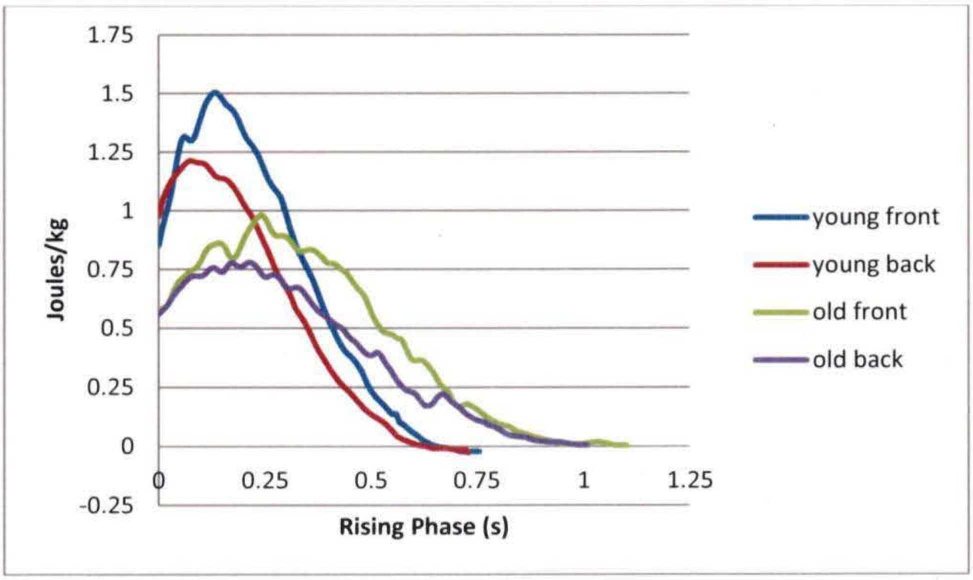

Figure 7. Average hip joint power (W/kg) in the rising phase. Area under the power-time curve indicates joint work (J/kg)

Joint Angular Impulse

For ankle angular impulse (Figure 8), there was not a significant interaction (p=0.48) and there was a not a significant effect of foot position (p=0.44). Conversely, there was a significant effect for age (p=0.023) as the young group resulted in a 0.07 Nms/kg reduction in angular impulse when compared to the old group.

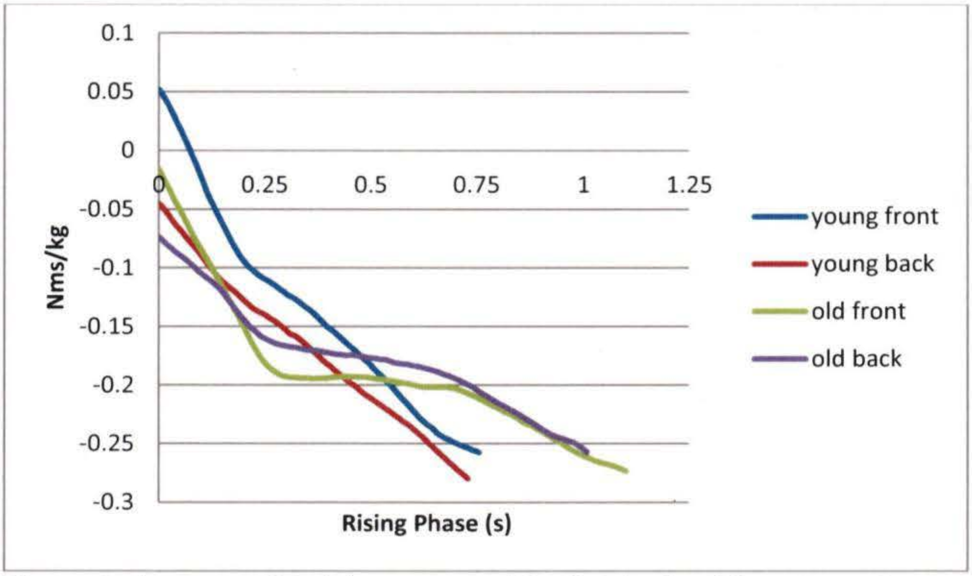

Figure 8. Average ankle joint moment (Nm/kg) in the rising phase. Area under the moment-time curve indicates joint angular impulse (Nms/kg).

For knee angular impulse (Figure 9) there was not a significant interaction (p=0.869). There was not a significant effect for the two foot positions (p=0.339) and there was no significant main effect for age (p=0.327).

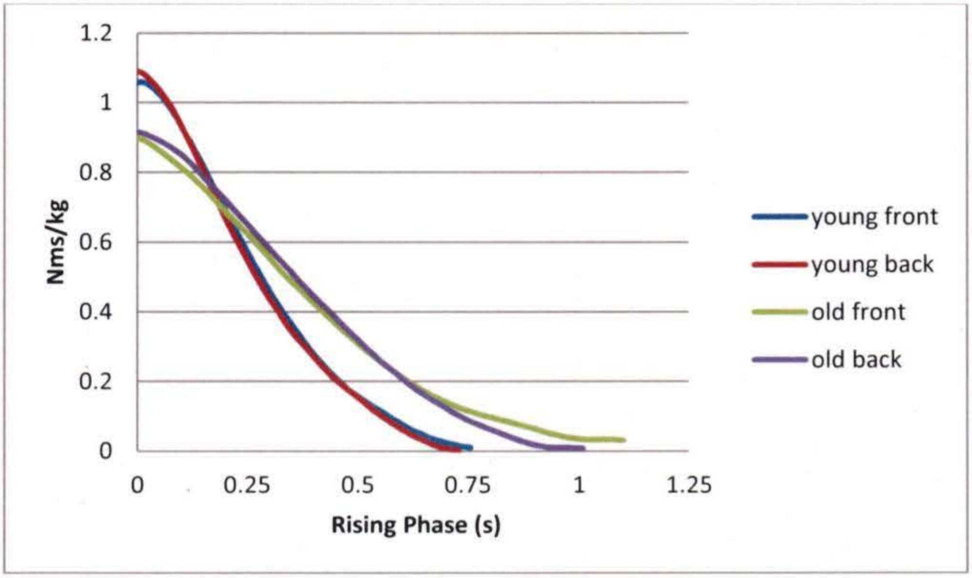

Figure 9. Average knee joint moment (Nm/kg) in the rising phase. Area under the moment-time curve indicates joint angular impulse (Nms/kg).

Unlike the knee joint angular impulse, the hip joint angular impulse (Figure 10) experienced significant results. There was not a significant interaction (p=0.867), although there was a significant effect of foot position (p<0.01) as the back foot position resulted in a 0.06 Nms/kg decrease in angular impulse when compared to the naturally selected foot position. However, there was not a significant effect for age (p=0.667).

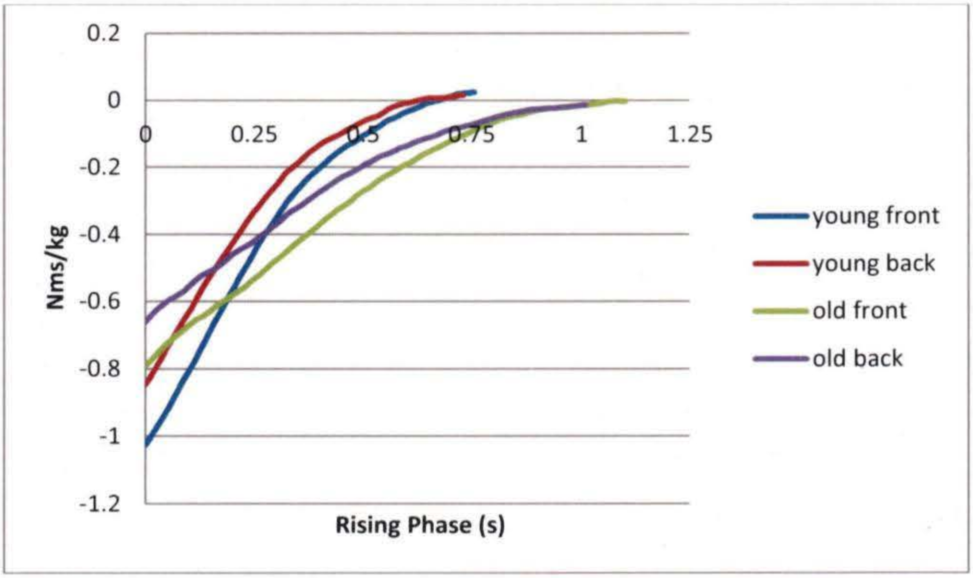

Figure 10. Average hip joint moment (Nm/kg) in the rising phase. Area under the moment-time curve indicates joint angular impulse (Nms/kg).

Electromyography

For the TA iEMG there was not a significant interaction (p=0.147) (Table 2). However, there was a significant effect of foot position (p=0.013) as the posterior foot position resulted in a 2% MVIC decrease in activity when compared to the naturally selected foot position. Unlike the results for foot position, there was not a significant effect for age (p=0.083). The SOL iEMG experienced opposite results from the TA iEMG. There was not a significant interaction (p=0.23) and there was not a significant effect of foot position (p=0.773). However, there was a significant effect for age (p=0.02) with the younger adults resulting in 21% MVIC less than the older adults. Lastly the GAS iEMG did not have a significant interaction (p=0.116), with no significant effect of foot position (p=0.131) and no significant effect for age (p=0.088).

For the REC iEMG there was not a significant interaction (p=0.351) and there was not a significant effect of foot position (p=0.249). However, there was a significant effect for age (p=0.001) with the younger adults resulting in 17 % MVIC less than the older adults. The VL iEMG shows opposite results from the REC iEMG. There was not a significant interaction (p=0.967); however, there was a significant effect of foot position (p=0.037) as the posterior foot position resulted in a 3% MVIC decrease when compared to the naturally selected foot position. There was not a significant effect for age (p=0.082). Lastly for the VM iEMG there was not a significant interaction (p=0.224) and there was not a significant effect of foot position (p=0.101) or a significant effect for age (p=0.055).

There was a significant interaction (p=0.015) for the MH iEMG. Post hoc paired t-test for young adults revealed an 8.5 %MVIC decrease from the naturally selected foot position to the posterior foot position (p=0.01). There was also a significant interaction for the BF iEMG (p<0.01). Post hoc paired t-test for older adults revealed a 5% MVIC decrease from the naturally selected foot position to the posterior foot position (p=0.002). A separate independent t-test for the naturally selected foot position resulted in a 13% MVIC increase from the young adults to older adults (p=0.001). Finally, the posterior foot position resulted in an 11% MVIC increase from the young adults to older adults (p=0.001).

Table 2. Group means standard deviations and significant interactions between young and older adults between natural foot position and foot back for iEMG dependent variables for the rising phase.

	Young		Old	
	Front	Back	Front	Back
(%MVIC)				
TA *	6.22(0.031)	5.33(0.043)	14.9(0.138)	11.9(0.113)
SOL †	12.2(0.117)	14.4(0.148)	35.5(0.241)	34.1(0.214)
GAS	5.38(0.064)	5.43(0.0574)	12.4(0.1003)	10.03(0.061)
REC †	9.38(0.0589)	9.18(0.0636)	27.6(0.107)	25.8(0.136)
VL *	36.4(0.218)	33.2(0.203)	54.9(0.207)	51.5(0.248)
VM	33.0(0.199)	32.3(0.209)	62.1(0.342)	57.4(0.38)
MH ǂ	5.08(0.0265)	4.24(0.023)	25.6(0.242)	20.6(0.203)
BF ǂ	7.98(0.041)	6.43(0.039)	21.5(0.092)	17.2(0.076)

† Significant main effect for age ($p<0.05$)
*Significant main effect for foot position ($p<0.05$)
ǂ Significant age x foot condition interaction ($p<0.05$)

Chapter 5: Discussion and Conclusions

Summary of Results

The aims of this study were to: 1) examine the joint kinetics, joint energetics and muscle activations of a STS task between young and older adults, and 2) determine the effects of posteriorly placing the feet on joint kinetics, joint energetics and muscle activations in STS. It was hypothesized that there would be significantly lower magnitude of iEMG activity and significant decreases in the magnitude of the hip and ankle joint angular impulse and work along with significant increases in work and angular impulse for the knee joint in older adults compared to the younger adults. It was also hypothesized that there would be a significantly increased knee angular impulse and work with a significant decrease in hip and ankle joint angular impulse and work as the feet are positioned posteriorly. Finally, it was hypothesized that with a posterior foot placement, there would be significantly lower magnitude of iEMG activity in the BF, MH, SOL, GAS and TA while there will be a significantly larger magnitude of EMG activity in the REC, VL, and VM.

After examining the effects of a self- selected foot position and a posterior foot placement in young and older adults, changes were observed in iEMG, joint kinetics and energetics with age and foot position. Specifically, there was an increase in iEMG for the REC, SOL, MH and BF in older adults versus younger adults; therefore we partially confirmed the research hypothesis for decreased magnitude of iEMG in all the muscles for older adults versus younger adults. The iEMG results may be influenced by the nature of the signal processing; specifically, scaling the iEMG to the participants MVIC as it is possible that maximum effort was not truly recorded. It is also important to consider that the first foot position was self-selected and all participants moved their feet posterior 0.10m. With this in mind, it is likely that each participant will have a unique initial angular configuration in there lower body joints and trunk. The variance in the joint and segment angles between foot positions for each participant may be a potential limitation in the final outcome of the results.

The topic of variance in foot position may be useful to understand the results of the ankle joint angular impulse. This was the only impulse parameter that was significantly higher for angular impulse when compared to younger adults. It is possible that the older adults may have a more posterior foot placement compared to the younger adults, resulting in this increase. None of the joint work variables were significantly different between age groups; therefore, the research hypothesis is partially supported for predicting the results of aging on the magnitude and distribution of joint kinetics and energetics.

For changes in joint kinetics and energetics with a posterior foot position there were statistically significant increases in work for the knee and ankle joint and statistically significant decreases in hip joint work. The hip joint had a statistically significant decrease in joint angular impulse with a posterior foot placement. Therefore the research hypothesis for statistically significant increases in knee joint work and angular impulse and statistically significant decreases in the hip and ankle joints is partially supported.

For changes in iEMG with a posterior foot position, the TA, VL, MH and BF were the only muscles that experienced significant decreases in iEMG from the resulting posterior foot placement. There were no muscles that experienced a significant increase in iEMG with a posterior foot placement. Therefore the research hypothesis for a significantly lower magnitude of iEMG activity in the BF, MH, SOL, GAS and TA and a statistically significantly larger magnitude of iEMG activity in the REC, VL, and VM is partially supported.

However, a posterior foot position resulted in increased amplitude in the SOL and GAS for the young adults while the older adults produced a decrease in SOL and GAS magnitude with a posterior foot placement. Although this was not significant, it may be related to the variance in the joint angles in the initial configuration with foot position.

The results for iEMG magnitude between young and older adults may be explained by the nature of the signal processing by integrating the linear envelope over the time of the rising phase. Since these integrated values are based off time curves of the respective dependent variables, the time to complete the STS may be useful in understanding the differences in iEMG amplitude in young and older adults. The older adults had a significantly higher time in the rising phase compared to the younger adults. With more time in the rising phase, there is more opportunity to accumulate muscle activations which would result in a higher magnitude of iEMG.

Discussion

Electromyography and Aging

In present study, the muscles that had statistically significantly lower iEMG magnitude in the older adults were the SOL, REC, MH and BF. This is evidence to suggest that there is a meaningful decrease in amplitude of these muscles with age. Previous research examined mean peak amplitudes of young and older adults in the STS that were normalized to the participants MVIC of the TA, SOL, GAS, RF, BF and Gluteus Maximus (Gross et al., 1998). Young adults had less movement time in the STS compared to the older adults in the present study and in the previous study (Gross et al., 1998). The peak magnitudes were greater in the older adults for all the muscles, but the RF was the only muscle to be statistically significantly different from the young adults (Gross et al., 1998). Although the magnitude was greater with advanced age, these were only peak values and do not account for the accumulative activation magnitude throughout the STS.

When iEMG was investigated in participants with knee extensor weakness from osteoarthritis, there was higher iEMG values in the stronger limb than the weakened limb (Davidson et al., 2013). It was expected that the weakness associated with aging would provide similar results to the weakness from osteoarthritis; however, this is not directly comparable. In the present study, only the dominant limb was investigated and there were pre-screening procedures to exclude participants with a disabling amount of osteoarthritis or other handicaps. A potential explanation for the current findings is the older adults did have lower physical activity scores from the questionaire compared to the young adult participants which may suggest that the younger adults are stonger than the older adults. Measuring iEMG may be useful in understaning the activation requiremnts for the STS, knowing the increased strength in younger adults requires less activation. Previous research addresses isometric knee extensor strength and peak activations normalized to the subjects MVIC in young and older adults. There was a clear relationship between knee extensor strength and peak RF activation showing a 45% decrease in maximal quadriceps strength and a 53% decrease in peak RF activation (Gross et al., 1998).

With age related weakness in mind, another factor to consider is the nature of the STS which requires the arms to be crossed over the chest. This arms crossed procedure is common for STS research; however, this is not a typical way to stand in a natural setting. It is possible that this arms crossed posistion was more physically strenouous for the older adults, requiring a larger magnitide of iEMG to complete the task. It is also possible that the nature of the MVIC influenced the results between age groups. Knowing that the iEMG is scaled to the MVIC, the participants may not have been able to produce a true MVIC. Future work should consider measuring isometric strength along with the MVIC of the knee flexors and extensors to help clarify the meaning of the iEMG values. Secondly, a qualitative measure for difficulty of the STS with the arms crossed should be considered.

Joint Angular Impulse and Aging

In the current study, there were changes in joint angular impulse at each joint but the ankle was the only to be statistically significant. The increase of magnitude for joint angular impulse from young to older adults may also be explained by the nature of the signal processing by integrating the joint moments over the time in the rising phase. Since there was more time in the rising phase for older adults, there is more opportunity to accumulate joint angular impulse. It can be inferred that the older adults in this study are weaker than the younger and are not able to produce the same magnitude in the peak joint torques. With this in mind, the older adults produce a submaximal torque which requires a longer time in the rising phase which ultimately results in the increased joint angular impulse. The results from the current study show that older adults have a significantly higher ankle joint impulse which may be further explained by the nature of the shank iEMG. The SOL and GAS showed increased EMG activity from the start of the STS and remained active through the stabilization phase (Roebroeck et al., 1994). The present study revealed a statistically significantly higher activation of the SOL in older adults, which may suggest that the increased activation may influence the amplitude of joint angular impulse and work in the ankle for older adults.

Previous research investigating integrated GRF shows that those with lower limb weakness from knee osteoarthritis display higher values of integrated GRF in the unaffected limb versus the weak limb (Davidson et al., 2013). When these values were compared to a healthy participant, there is a larger overall value for integrated GRF for participants with osteoarthritis. It is also important to note that the healthy participants had higher scores for isometric knee extensor strength and were able to complete a five times STS more quickly than the participants with knee osteoarthritis (Davidson et al., 2013). Although the variables are different from the current study, this information may suggest that muscle weakness and a slow rise in the STS contributes to increased impulse. It can be inferred from Figures 8, 9 and 10 that younger adults produce larger peak torques in the lower body joints resulting from increased strength. With a larger peak joint torque production, this will require less time to complete the rising phase which ultimately results in a lower joint angular impulse. More importantly, joint angular impulse may reveal more useful information when comparing limbs. This would indicate if one side is has performance deficits resulting in an over dominant side. With this in mind, future work employing joint angular impulse should compare both limbs.

Joint Work and Aging

Although not significant, the joint work showed an increase in the ankle for the older adults but there was a higher magnitude of joint work in the knee and hip joints for the young adults. The ability for younger adults to produce higher values for knee and hip joint work suggest that more energy is available from the knee and hip extensors muscles. Conversely, older adults seem to have a reliance of the ankle joint to produce work more than the younger adults. This change in magnitude and distribution of joint work reveals what changes we can expect in energy production with aging.

Previous research examined muscle power in the STS comparing young adults to healthy older adults to older adults at risk for falling (Cheng et al., 2014). Maximal power was calculated by multiplying the VGRF and the vertical upward velocity of the COM and normalized to each subject's body weight. The results show that the time to complete the STS was shorter for young adults and that the young adults were able to produce more power (Cheng et al., 2014). Knowing that joint work was integrated from the joint power time curve during the rising phase, it is possible that muscle weakness and a slow rise in the STS contribute to decreased magnitude in peak joint power. With this in mind, the older adults produce a submaximal power which requires a longer time in the rising phase which ultimately results in a decrease in joint work. The ability to increase strength in the knee extensors will increase the amount of knee joint peak power production. Since the quadriceps is the prime movers for the STS, finding techniques to improve knee joint work should decrease the risk of falling and improve quality of life for ADLs.

Future research should consider measuring peak muscle activations along with comparing isometric strength, and peak joint powers and moments for young and older adults in the lower body. Having this extra information will show how the peak magnitudes change with aging. The kinetic and energetic figures show the younger adults produce higher peaks and result in a lower time for the rising phase. This may be useful for rehabilitation specialists to improve STS performance, knowing how decreased muscular power is an indicator for falls risk (Cheng et al., 2014)

Posterior Foot Position

The current study examined the role of a posterior foot placement on the distribution and magnitude of joint kinetics and energetics and iEMG of the lower body muscles. The results indicate a posterior foot placement provided a significant decrease in hip joint angular impulse. The results reveal statistically significant increases in ankle and knee joint work with a statistically significant decrease in hip joint work. This study revealed that older adults produce less work in the hip and knee joints and a higher angular impulse in all the lower body joints from a resulting slower time to complete the rising phase. With a posterior foot placement, the time of the rising phase was significantly less than a naturally selected foot placement. Since this change of foot position results in less time in the rising phase, the joint angular impulse values are decreased and the knee and ankle joints are able to produce more work for this task. This simple strategy shift provides the ability to more effectively utilize energy from the knee and ankle joint while subsequently reducing the loading associated with the decrease in joint angular impulse.

The results of the kinetics, energetics and iEMG in regards to foot position may be explained by the initial kinematic configuration. With a posterior foot placement, the knee joint will become more flexed allowing a greater range of motion to produce more knee joint work. This increase in the knee joint work is useful for improving the quadriceps abilities to produce force. This increased force is not beneficial for knee joint angular impulse which would increase loading. However; this was not the case in the present study, there was not a statistically significant decrease in knee joint angular impulse. This phenomenon may be explained by the role of trunk flexion. Previous research has shown a positive correlation between knee extensor strength and the degree of trunk flexion (Gross et al., 1998). The more the trunk is flexed; the moment arm of the knee is decreased allowing a smaller production of knee joint extensor torque in the STS. This was the case for the present study as trunk flexion was significantly different between the two foot positions. Since joint angular impulse is the integral of the joint moment time curve, the decreased torque production should result in less joint angular impulse.

The musculature of the knee joint experienced an overall decrease in iEMG activity with a posterior foot placement. The only muscle to have a significant decrease was the VL. Previous research comparing anterior and posterior foot placements revealed no significant effects for condition on onsets of muscle activity (Dehail et al., 2007). It is possible that since there was less time in the rising phase with a posterior foot placement, that there would be less of an opportunity to apply iEMG over the time curve of the rising phase. Secondly, since the knee joint work increased with a posterior foot placement, this configuration may not require the magnitude of muscle activation to produce the necessary work to complete the task. This result may be another useful reason to suggest a posterior foot placement. Knowing that the quadriceps is the prime mover and the goal is to maximize strength, the ability to reduce the iEMG magnitude shows that a posterior foot placement is useful for this goal.

. The BF iEMG experienced a statistically significant reduction in amplitude with a posterior foot placement. It is likely that a posterior foot position requires increased knee flexion allowing a greater range of motion for knee joint work; there will be less co-activation of the antagonist. Previous research shows healthy participants exhibiting higher iEMG activation of the quadriceps over the hamstrings in a STS (Davidson et al., 2013). This was also examined in participants with muscle weakness from knee osteoarthritis. The hamstring activation was almost the same as the quadriceps activation in the unaffected limb and the hamstring activation was higher than the quadriceps in the limb affected by osteoarthritis (Davidson et al., 2013). With this in mind, the ability to produce more force and energy with the quadriceps should reduce the activation magnitude of the hamstrings. It is also noted that the BF and MH were the only variables to experience a statistically significant interaction between age and foot condition. This indicates that muscle weakness from aging and an anterior foot position will result in lower knee joint work and will require additional activation of the hamstrings.

Conclusion

There were many changes that occurred with the STS from the results of a posterior foot position and the influence of age. Many of these changes were expected from the initial kinematic configuration and what we know about physiological changes with aging. What the current study did validate from previous research is the initial foot position is an important component to the ease of standing up (Dehail et al., 2007). The current investigation can express the ease of standing up by quantifying and identifying the time, joint angular impulse, joint work, trunk flexion and the iEMG of the muscles considered to be prime movers and anticipatory muscles.

An important factor to consider with a posterior foot placement is that the base of support is now closer to the COM of the participant. Since the COM has a smaller distance to travel to the base of support, this should require less time to complete the rising phase. More importantly, a posterior foot position results in less trunk flexion range and time. This is important for improving the dynamic stability of the COM; with less time in the STS, the COM can stabilize quicker. Secondly, less trunk flexion will also result in a smaller range in COM displacement which is important for maintaining dynamic stability. Finally, increasing trunk flexion will decrease the moment arm at the knee joint, resulting in less energy and torque production. Since the feet back position results in less trunk flexion; both younger and older adults were able to generate more energy from the knee joint, suggesting enhanced quadriceps utilization for the STS.

What this study revealed was that a posterior foot position shift of 0.10 m significantly reduced the time to complete the STS. The time to complete a STS is relevant to the amplitude of integrated values for EMG, kinetics and energetics knowing these variables is scaled to a time curve. It is also important to consider the nature of the five times STS test that a score is given based on the time to complete the 5 STS. If this test is intended to be repeated to track progress, it is essential to have a standardized foot position (Kwong et al., 2014).

This study revealed the TA will have larger amplitude of iEMG with a posterior foot placement. Although the TA is not muscle that is a prime mover to displace the COM in the STS, it inherits an important feature of anticipation of the STS movement.

The early activation of the TA can be viewed as an anticipated postural adjustment with a purpose of stabilizing the foot before the body moves forward. It is suggested that the lack of initial activation of the TA could exemplify degradation of the STS (Dehail et al., 2007). The results of this study can prove useful to clinicians on how to effectively utilize a posterior foot position to improve the kinetics and energetic requirements for a STS in young and older adults.

Limitations

This study also reveals the limitation to a posterior foot position. Knowing that a posterior foot position has several significant effects, the foot positions were not standardized and there is a possibility that there may be a large variance in the initial kinematics of the lower body joints and trunk segments. This is an important factor to consider since the derived kinetic and energetic variables are based off inverse dynamics which requires kinematic positions. Lastly the nature of the MVIC may not have been the best approach to compare the iEMG of younger and older adults. There is a possibility that the older adults are not capable of providing a true MVIC which may have negative consequences to the accuracy of the results.

Future Considerations

A posterior foot placement produced a significant increase in knee joint work while decreasing the iEMG amplitude of the TA, trunk flexion and the time in the rising phase. With this in mind, there are likely changes in kinematics. Future research should consider measuring kinematics to demonstrate the variance in the initial kinematic configuration. Using the MVIC approach isn't necessarily out of the question; however, future research should also scale the EMG to the peak amplitude to the trial and take an average of the trials and make comparisons from there.

There are clear changes from a posterior placement of both feet. Future work should also investigate the STS in a staggered stance with one foot more posterior than the other. This stagger position should impose higher kinetic, energetic and EMG demands to one side which may be a useful component to an exercise program to target unilateral efforts to improve STS performance. Lastly, future work should investigate both limbs and include older adults that are at risk of falling. This would be beneficial to determine what kinetic, energetic and EMG aspects are important to train to ensure the best quality of life possible for older adults.

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Appendix A: Informed Consent Document

INFORMED CONSENT DOCUMENT
OLD DOMINION UNIVERSITY

PROJECT TITLE: THE EFFECT OF POSTERIOR FOOT POSITIONS ON ELECTROMYOGRAPHY, JOINT KINETICS AND ENERGETICS FOR DURING A SIT-TO-STAND IN YOUNG AND OLDER ADULTS

INTRODUCTION

The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. This project is being carried out to determine the effects of foot placement and the changes in force and EMG activity on a sit-to-stand maneuver. The research will be conducted at Old Dominion University in the Neuromechanics Laboratory.

RESEARCHERS

Joshua Weinhandl, PhD, Responsible Project Investigator, Assistant Professor, Darden College of Education, Department of Human Movement Sciences, Old Dominion University

Eric Pitman, Graduate Student, Department of Human Movement Sciences, Old Dominion University

DESCRIPTION OF RESEARCH STUDY

The purpose of this study is to determine the influence of varied foot positions on sit-to-stand performance in young and older adults. This study will compare performance of this task with the subject's feet initially in a forward and back position. A total of 40 subjects will participate in this study. You will report to the Neuromechanics Laboratory, SRC 1007, for one session. You will be asked to participate in one, two hour-long testing session. You will rise from the seated position at a self-selected natural speed and no restrictions in forward leaning of your torso. During the sit to stands, you will be accompanied by a spotter to provide safety and minimize the risk of falling.

The session will involve performing 10 sit-to-stand maneuvers with the subject's arms across their chest and seated comfortably towards the edge of the seat. The 10 stands will be divided evenly into 2 placements; self-selected and then placing both feet back by 10 centimeters.

Data collection will consist of measurements of height and weight. Proper attire for physical activity is suggested, however clean, tight-fitting shorts will be provided. Single reflective markers will be placed on specific anatomical landmarks. Furthermore, eight surface electromyography (EMG) electrodes will be placed on eight dominant leg muscles. Placement of the EMG electrodes will require the removal of hair followed by skin abrasion and then the site will be cleaned with rubbing alcohol. After reflective markers and EMG electrodes have been applied, three-dimensional kinematics and kinetics will be collected during the sit-to-stand. During the visit to the lab, you will be asked fill out a Medical History, osteoarthritis screening and Physical Activity Questionnaire. This includes questions pertaining to age, physical activity, lower body injury (ies), recent head injury (ies), or any medications that may cause dizziness, etc.

EXCLUSIONARY CRITERIA

To be eligible to participate you must be physically active three days a week for at least 30 minutes and perform at least one session of lower body resistance training per week.

You will not be able to participate in the study if you:

- have suffered any injuries to the lower extremities within the last six months.
- have ever had surgery to your ankles, knees, hips, and or lower back.
- are pregnant or think you may be pregnant.
- if you experience pain in standing up
- if you experience chronic osteoarthritis in ankles, knees, hips, and or lower back.
- young participants must be between 18-35 years of age and older participants must be between 65-85 years of age.

RISKS AND BENEFITS

RISKS:

If you decide to participate in this study, then you may experience minor skin irritation due to the skin preparation and adhesive on the external EMG electrodes. It is possible you may experience a loss of balance resulting in falling or a musculoskeletal injury such as a muscle strain.

To reduce the above risks, care will be taken when applying and removing the external markers. To minimize the effects of the use of EMG electrodes, aloe Vera will be provided after the study to alleviate skin irritation. If you feel any soreness or irritation while participating in this study, please tell the investigators as soon as possible. The investigators will provide you with the appropriate information for treating these problems, based on our level of expertise. To minimize loss of balance and/or falling, a spotter will be present when performing the sit-to-stand task. If you are injured while participating in this research study, you will initially be provided care by the investigator(s) and will then be referred to the Student Health Services (students) or your personal physician (non-students) for follow-up care.

There is a small risk of loss of confidentiality. To minimize this risk all information gathered from you will be confidential in nature and stored in the laboratory of the principle investigator (SRC 1007). Only the principal investigator and research personal will have access to these files.

BENEFITS:

There are no direct benefits from participation in the research study. The information obtained in this study will expand our knowledge base leading to beneficial changes in the future.

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study. If you are a student in Exercise Science courses that offer extra research credit, then you may be offered extra credit depending on the instructor's approval. If you are an older adult participant, you are eligible for 1 month free membership with the Wellness Institute and Research Center exercise program at Old Dominion University.

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

The researchers will provide confidentiality to all documents regarding patient information, including questionnaires and test results, by storing information in a safe, locked location. The results of this study may be used within reports, presentations or publication; but all personal identifiers will be disregarded. However, your records may be subpoenaed by court order or inspected by government bodies with oversight authority. Following the completion of the study, all subject information will be destroyed.

WITHDRAWAL PRIVILEGE

In the event that you no longer wish to participate, you have the right to discontinue participation for this study. Even if you initially wish to participate you can withdraw at any time. Your decision will not affect your relationship with Old Dominion University or cause a loss of benefits to which you might otherwise be entitled. In addition, the researcher has the right to withdraw your participation if they find potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you participate, then your consent in this document does not waive any of your legal rights. In the event that you are harmed or injured from participating in this study, neither Old Dominion University nor the researchers are able to give you any compensation, including money, insurance coverage, or free medical care. In the event that you suffer any injury from participation in this study, you may contact Dr. Joshua Weinhandl at (757) 683-4754, Dr. George Maihafer, the current IRB chair, at (757) 683-4520, or Office of Research (757) 683-3460 at Old Dominion University.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them:

Joshua Weinhandl 757-683-4754

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. George Maihafer, the current IRB chair, at 757-683-4520, or the Old Dominion University Office of Research, at 757-683-3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	**Date**

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	**Date**

Appendix B: Disablement in the Physically Active Scale

Instructions: Please answer **each statement** with one response by shading in the box that most closely describes your problem(s) within the past **24 hours**. Each problem has possible descriptors under each and not all descriptors may apply to you but are given as common examples.

KEY
0 - No Problem
1 - I have the problem(s), but it does not affect me
2 - The problem(s) slightly affects me
3 - The problem(s) moderately affects me
4 - The problem(s) severely affects me

	No Problem 0	Does not affect 1	Slight 2	Moderate 3	Severe 4
Pain: "Do I have pain?"	☐	☐	☐	☐	☐
Motion: "Do I have impaired motion?" e.g., decreased range/ease of motion, flexibility, and/or increased stiffness	☐	☐	☐	☐	☐
Muscular Functioning: "Do I have impaired muscle function?" e.g., decreased strength, power, endurance, and/or fatigue	☐	☐	☐	☐	☐
Stability: "Do I have impaired stability?" e.g., the injured area feels loose, gives out, or gives way	☐	☐	☐	☐	☐
Changing Directions: "Do I have difficulty with changing directions in activity?" e.g., twisting, turning, starting/stopping, cutting, pivoting	☐	☐	☐	☐	☐
Daily Actions: "Do I have difficulty with daily actions that I would normally do?" e.g., walking, squatting, getting up, lifting, carrying, bending over, reaching, and going up/down stairs	☐	☐	☐	☐	☐
Maintaining Positions: "Do I have difficulty maintaining the same position for a long period of time?" e.g., standing, sitting, keeping the arm overhead, or sleeping	☐	☐	☐	☐	☐
Skill Performance: "Do I have difficulty with performing skills that are required for physical activity?"					
1) e.g., running, jumping, kicking, throwing & catching	☐	☐	☐	☐	☐
2) e.g., coordination, agility, precision & balance	☐	☐	☐	☐	☐
Overall Fitness: "Do I have difficulty maintaining my fitness level?" e.g., conditioning, weight lifting & cardiovascular endurance	☐	☐	☐	☐	☐

Participation in Activities: "Do I have difficulty with participating in activities?"					
1) e.g., participating in leisure activities, hobbies & games	☐	☐	☐	☐	☐
2) e.g., participating in my sport(s) of preference	☐	☐	☐	☐	☐
Well-Being: "Do I have difficulties with the following…?"					
1) increased uncertainty, stress, pressure, and/or anxiety	☐	☐	☐	☐	☐
2) altered relationships with team, friends, and/or colleagues	☐	☐	☐	☐	☐
3) decreased overall energy	☐	☐	☐	☐	☐
4) changes in my mood and/or increased function	☐	☐	☐	☐	☐

Appendix C: Godin Leisure-Time Exercise Questionnaire

1. During a typical **7-day period** (a week), how many times on average do you do the following kinds of exercise for **more than 15 minutes** during your free time (write on each line the appropriate number).

	Times per week
a. **STRENUOUS EXERCISE** (Heart Beats Rapidly) e.g., running, jogging, hockey, football, soccer, squash, basketball, cross country skiing, judo, roller skating, vigorous swimming, vigorous long distance bicycling	________
b. **MODERATE EXERCISE** (Not Exhausting) e.g., fast walking, baseball, tennis, easy bicycling, volleyball, badminton, easy swimming, alpine skiing, popular and folk dancing	________
c. **MILD EXERCISE** (Minimal Effort) e.g., yoga archery, fishing from river bank, bowling, horseshoes, golf, snow-mobiling, easy walking	________

2. During a typical **7-day period** (a week), in your leisure time, how often do you engage in any regular activity **long enough to work up a sweat** (heart beats rapidly)?

OFTEN	SOMETIMES	NEVER/RARELY
☐	☐	☐

Appendix D: Western Ontario and McMaster Universities Osteoarthritis Index

(WOMAC)

Please rate the activities in each category according to the following scale of difficulty:

0 - No difficulty
1 - A little bit of difficulty
2 - Moderate difficulty
3 - Quite a bit of difficulty
4 - Extreme difficulty or unable to perform activity

Mark **one box** for each activity:

		None 0	Slight 1	Moderate 2	Very 3	Extreme 4
Pain	1. Walking	☐	☐	☐	☐	☐
	2. Stair climbing	☐	☐	☐	☐	☐
	3. Nocturnal	☐	☐	☐	☐	☐
	4. Rest	☐	☐	☐	☐	☐
	5. Weight bearing	☐	☐	☐	☐	☐
Stiffness	1. Morning stiffness	☐	☐	☐	☐	☐
	2. Stiffness occurring later in the day	☐	☐	☐	☐	☐
Physical Function	1. Descending stairs	☐	☐	☐	☐	☐
	2. Ascending stairs	☐	☐	☐	☐	☐
	3. Rising from sitting	☐	☐	☐	☐	☐
	4. Standing	☐	☐	☐	☐	☐
	5. Bending to floor	☐	☐	☐	☐	☐
	6. Walking on flat surface	☐	☐	☐	☐	☐
	7. Getting in/out of car	☐	☐	☐	☐	☐
	8. Going shopping	☐	☐	☐	☐	☐
	9. Putting on socks	☐	☐	☐	☐	☐
	10. Lying in bed	☐	☐	☐	☐	☐
	11. Taking of socks	☐	☐	☐	☐	☐
	12. Rising from bed	☐	☐	☐	☐	☐

13. Getting in/out of bath	☐	☐	☐	☐	☐
14. Sitting	☐	☐	☐	☐	☐
15. Getting on/off toilet	☐	☐	☐	☐	☐
16. Heavy domestic duties	☐	☐	☐	☐	☐
17. Light domestic duties	☐	☐	☐	☐	☐

Appendix E: Lower Extremity Functional Scale

We are interested in knowing whether or not you are having any difficulty at all with the activities listed below. Please provide an honest answer for each activity.

KEY

0 - Extreme difficulty or unable to perform activity
1 - Quite a bit of difficulty
2 - Moderate difficulty
3 - A little bit of difficulty
4 - No difficulty

Today, ***do you*** or ***would you*** have any difficulty at all with:	Extreme 0	Quite a bit 1	Moderate 2	Minimal 3	None 4
1. Any of your usual work, housework or school activities	☐	☐	☐	☐	☐
2. Your usual hobbies, recreational or sporting activities	☐	☐	☐	☐	☐
3. Getting into or out of the bath	☐	☐	☐	☐	☐
4. Walking between rooms	☐	☐	☐	☐	☐
5. Putting on your shoes or socks	☐	☐	☐	☐	☐
6. Squatting	☐	☐	☐	☐	☐
7. Lifting an object, like a bag of groceries from the floor	☐	☐	☐	☐	☐
8. Performing light activities around your home	☐	☐	☐	☐	☐
9. Performing heavy activities around your home	☐	☐	☐	☐	☐
10. Getting into or out of a car	☐	☐	☐	☐	☐
11. Walking 2 blocks	☐	☐	☐	☐	☐
12. Walking a mile	☐	☐	☐	☐	☐
13. Going up or down 10 stairs (about 1 flight)	☐	☐	☐	☐	☐
14. Standing for 1 hour	☐	☐	☐	☐	☐
15. Sitting for 1 hour	☐	☐	☐	☐	☐
16. Running on even ground	☐	☐	☐	☐	☐
17. Running on uneven ground	☐	☐	☐	☐	☐
18. Making sharp turns while running fast	☐	☐	☐	☐	☐
19. Hopping	☐	☐	☐	☐	☐
20. Rolling over in bed	☐	☐	☐	☐	☐

Appendix F: Falls Risk Assessment Questionnaire

Please mark "yes" or "no" for each statement.

1.	I have fallen in the last 6 months.	yes ☐	no ☐
2.	I use or have been advised to use a cane or walker to get around safely.	yes ☐	no ☐
3.	Sometimes I feel unsteady when I am walking.	yes ☐	no ☐
4.	I steady myself by holding onto furniture when I walk.	yes ☐	no ☐
5.	I am worried about falling.	yes ☐	no ☐
6.	I need to push with my hands to stand up from a chair.	yes ☐	no ☐
7.	I am often dizzy when I first stand up.	yes ☐	no ☐
8.	I have trouble stepping up onto a curb.	yes ☐	no ☐
9.	I often have to rush to the toilet.	yes ☐	no ☐
10.	I have lost some feeling or have pain in my feet.	yes ☐	no ☐
11.	I take medicine that sometimes makes me feel light-headed or more tired than usual.	yes ☐	no ☐
12.	I take medicine to help me sleep or improve my mood.	yes ☐	no ☐
13.	I often feel sad or depressed.	yes ☐	no ☐

Curriculum Vita

EDUCATION
Old Dominion University
M.S. (Exercise Science and Wellness)-August 2015
GPA: 3.51

Old Dominion University
B.S. (Physical Education) – December 2012
GPA: 2.81

PROFESSIONAL EXPERIENCE

Graduate Teaching Assistant
Exercise Science 415, Exercise Testing laboratory instructor
August 2014- May 2015

Graduate Research assistant
Neuromechanics Laboratory, Old Dominion University
October 2013- May 2014

Peer Reviewed Abstracts:

1. Pitman E, Sievert ZA Fontenot K, Mutchler JA & Weinhandl JT. (2015). The effect of posterior foot placements on Sit-to-Stand Kinetics, Energetics and Muscle activity. Proceedings of the 39th American Society of Biomechanics. Columbus, OH.

2. Pitman E, Fontenot K, Sievert ZA & Weinhandl JT. (2014). Comparison of unilateral drop landings and land-and-cut maneuvers. Proceedings of the 7th World Congress of Biomechanics. Boston, MA.

3. Fontenot K, Pitman E, Sievert ZA & Weinhandl JT. (2014). The influence of gender and limb dominance on lower extremity joint mechanics during land-and-cut maneuvers. Proceedings of the 7th World Congress of Biomechanics. Boston, M

Research Presentations:

1. Pitman E, Fontenot K, Sievert ZA & Weinhandl JT. Comparison of unilateral drop landings and land-and-cut maneuvers. Graduate Research achievement day, Old Dominion University. March 27, 2014.

2. Pitman E, Fontenot K, Sievert ZA & Weinhandl JT. Comparison of unilateral drop landings and land-and-cut maneuvers. 7th World Congress of Biomechanics July7, 2014.

PROFESSIONAL MEMBERSHIPS

National Strength and Conditioning Association
American Society of Biomechanics
American Council on

PROFESSIONAL MEETINGS ATTENDED

2014 – 7th World Congress of Biomechanics. Boston, MA.