SIT-TO-STAND TRANSFER MECHANICS: THE EFFECT OF AGE AND LIFTING-SEAT DEVICE DESIGN

by

Sean Hurley

Submitted in partial fulfilment of the requirements
for the degree of Master of Applied Science

at

Dalhousie University
Halifax, Nova Scotia
August, 2013

This thesis is dedicated to my parents, G and T Hurl, for their amazing support over the years.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

ABSTRACT

Objective: Lifting-seat devices are designed to raise the seat height to reduce biomechanical and neuromuscular demands of a sit-to-stand (STS). The goal of this thesis was to understand how seat height and lifting-seat devices with different mechanisms affect trunk, hip, and knee biomechanics and neuromuscular activity of surrounding muscles, and determine whether the effects are altered by age. Four conditions were tested; 1) no device normal seat height (ND-normal), 2) Seat Assist™ (SA), 3) Power Seat™ (PS), 4) no device raised height (ND-raised). Using a cross-sectional design, two objectives were completed. Objective one compared ND-normal and ND-raised to determine the effects of seat height. Objective two compared ND-raised, PS, SA to determine the effects of lifting-seats. **Design:** 10 healthy older and 10 healthy younger adults performed five trials of each STS condition. Bilateral lower limb and trunk three-dimensional motion, ground reaction forces and electromyography (EMG) from five muscles were collected. Peak values were extracted from kinematic waveforms. Peak and integrated values were extracted from net external moment and EMG waveforms. Two-way analysis of variance models tested for main effects (group x condition) and interactions ($p<0.05$) on the dependent variables. Bonferonni post-hoc testing was used to test all significant findings. **Results:** For objective one, there was one significant group by condition effect for integrated EMG of the vastus medialis (VM). ND-normal had greater peak angles, peak and impulse hip flexion moment values, knee flexion moment impulse, and peak and integrated EMG activity of all muscles (except integrated VM). For objective two, peak trunk, hip and knee flexion angles were greater for the SA. Peak and impulse hip flexion moments were greater for the SA, whereas, peak knee flexion moments were greater for the PS. Peak EMG activity of vastii muscles was greater for the PS and the SA compared to ND-raised with differences in the other muscles specific to device. Both devices had greater integrated EMG of the quadriceps. Only the SA had greater integrated EMG of lateral hamstrings and gluteus maximus. For both objectives, older adults had significantly greater peak and integrated EMG activity of most muscles. **Conclusions:** Older adults performed all conditions with similar kinetics and kinematics compared to the younger group, but had greater EMG activity as a percentage of maximum. Objective one results confirmed that the raised seat height reduced demands on the hip and the knee joints and surrounding muscles, affecting both groups similarly, except for VM activity, which had a greater reduction in overall activity in older adults with a raised seat height compared to younger adults. Objective two provided evidence that PS and SA devices have different effects on biomechanics and muscle activation compared to ND-raised. The pneumatic device (SA) changed the mechanics at the trunk, the hip, and the knee and increased muscle activation of all muscles. The electric-elevator device (PS) changed the mechanics at the knee and increased muscle activation of the quadriceps only. All three seating conditions had similar effects on both younger and older adults. This study provides a comprehensive comparison of seating mechanisms and age-related changes, which have implications for both design and prescription of the devices for those with sit-to-stand disabilities.

LIST OF ABBREVIATIONS USED

STS	Sit-To-Stand
EMG	Electromyography
GRF	Ground Reaction Force
COP	Center of Pressure
OA	Osteoarthritis
COM	Center of Mass
MVIC	Maximum Voluntary Isometric Contraction
ND-normal	No Device from normal seat height
SA	Seat Assist™
PS	Power Seat™
ND-raised	No Device from a raised seat height
LEFS	Lower Extremity Functional Scale
IRED	Infrared Emitting Diode
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
Ag/AgCl	Silver/Silver Chloride
VL	Vastus Lateralis
VM	Vastus Medialis
RF	Rectus Femoris
LH	Lateral Hamstring
GM	Gluteus Maximus
GΩ	Gigaohm
dB	decibel
Hz	Hertz
V	Volt
A-D	Analog to Digital
ms	millisecond
Nm/kg	Newton•meter/kilogram
COG	Center of gravity
GT	Greater Trochanter
Nms/kg	Newton•meter•second /kilogram
%MVICs	Percent Maximum Voluntary Isometric Contraction•second
BMI	Body Mass Index
ANOVA	Analysis of Variance
WHR	Waist to Hip Ratio
GOLD	Gold Standard Standing Event
TSP	Thoracic Spine Standing Event Method
HipExt	Hip Extension Standing Event Method
KneeExt	Knee Extension Standing Event Method
vGRF	Vertical Ground Reaction Force Standing Event Method
SH	Shoulder Standing Event Method

ACKNOWLEDGEMENTS

The list of acknowledgements for this project is lengthy.

Firstly, I would like to thank my supervisor, Dr. Cheryl Hubley-Kozey for providing a world-class research environment to grow and learn, as well as, for her leadership and guidance throughout the completion of this thesis.

I would also like to thank Dr. Derek Rutherford for his many contributions to all aspects of this project. Additionally, for always having an open door to answer one of my 'quick' questions and for sharing his research expertise and many thoughtful suggestions.

I wish to express many thanks to Dr. Stanish for providing me with every opportunity to succeed. I greatly appreciate all of the generosity and mentorship over the years.

And of course, the DOHM team (past and present), especially Kathryn, Gillian, Adam, Kerry, Janie, Dianne, Jereme, Shawn, and Nick. This project was far from a solo effort and I owe you all many thanks.

I would also like to acknowledge Uplift Technologies Inc. for their collaboration. Also, financial support of the Natural Sciences and Engineering Research Council of Canada, Nova Scotia Health Research Foundation, Dalhousie Biomedical Engineering NSERC CREATE:Biomedic program, and Dalhousie's Department of Engineering.

Finally, the entire Hurley Family (Mom, Dad, Kev, Mike, and Bea) - I could not ask for a better support team. And Andrea - for all her patience and encouragement over the past few years, as well as being a great study partner.

CHAPTER 1 INTRODUCTION

A sit-to-stand (STS) transfer is a complex maneuver that places demands on the musculoskeletal system that are greater than most other functional tasks (Mizner, Snyder-Mackler 2005, Hodge et al. 1989). STS involves the transfer from a stable three-point sitting position to a more unstable two-point standing position, leading to mechanical and physiological challenges in maintaining stability and generating adequate muscle force (Hughes, Myers & Schenkman 1996, Savelberg et al. 2007, Van der Heijden et al. 2009, Akram, McIlroy 2012, Fujimoto, Chou 2012). As a precursor to walking, STS limitations can decrease an individual's mobility (Guralnik et al. 1995), and physical activity, potentially increasing the risk of a loss of independence and mortality (Hirvensalo, Rantanen & Heikkinen 2000). Rehabilitation devices that lower the mechanical demands of rising from a chair have the potential to help older adults with lower limb impairments maintain a more independent and active lifestyle. Active lifestyle is associated with health benefits such as; lower risk of premature death, heart disease, stroke, colon and breast cancer, type-2 diabetes, improved muscular, bone, and joint health, as well as reduced depression (Physical Activity Guidelines 2008).

Older adults with muscle strength deficits often have difficulty rising from a chair or have adopted different kinematic strategies compared to healthy younger adults (Hughes, Myers & Schenkman 1996, Papa, Cappozzo 2000, Gross et al. 1998, Fujimoto, Chou 2012). Hughes et al. (1996) found older individuals with functional impairments used approximately 97% of their available knee extensor strength when rising from a lower than normal chair height compared to 39% in younger individuals. Given decreased strength with age (Hughes, Myers & Schenkman 1996, Fujimoto, Chou 2012, Gross et al. 1998), similar increases in muscle activity as a percentage of maximum during the STS transfer would be expected. However, no studies have compared muscle activation during a STS transfer between younger and older adults. This is important as inverse dynamics modeling used to calculate joint moments of force does not consider muscle activation and older adults have been shown to have higher antagonist-agonist co-activation during other functional activities (Nagai et al. 2011, Quirk, Hubley-Kozey 2012).

Older adults with knee extensor impairments have been shown to increase trunk flexion to shift their center of mass over their feet to reduce the knee flexion moments (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Scarborough, McGibbon & Krebs 2007). However, Savelberg et al. (2007) provided evidence that individuals do not change their kinematic and kinetic strategy until a threshold level when their knee extensor muscles become overloaded. This was determined by incrementally adding weight to a vest to simulate muscle impairments in healthy individuals. At 45% body weight, the strategy changed to a ‘stabilization strategy’ with more trunk flexion to keep the knee moments constant and redistributing a greater ratio to the hip and the ankle (Savelberg et al. 2007). Overall, older adults, particularly those with knee extensor muscle impairments often change their STS strategy compared to healthy younger adults and may respond differently to rehabilitative interventions designed to facilitate STS transfers.

Seating platforms with raised seat heights have been shown to alter joint motion and reduce the biomechanical demands of the hip and the knee joint during a STS transfer (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Schenkman, Riley 1996). More specifically, compared to a normal seat height, a raised seat decreases trunk, hip, and knee angular velocity and displacement (Munro et al. 1998, Schenkman, Riley 1996, Kuo, Tully & Galea 2009, Rodosky, Andriacchi & Andersson 1989, Su, Lai & Hong 1998), and hip and knee flexion moments (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi & Andersson 1989). Based on kinetic findings, decreased muscle activation of lower limb muscles from a raised seat height would be expected but there is limited objective electromyographic (EMG) data. Two studies have reported reduced peak muscle activation of the quadriceps and triceps brachii muscles with a raised seat height (Arborelius, Wretenberg & Lindberg 1992, Munro, Steele 2000). Only one study has examined a hip extensor muscle, reporting no effect of seat height on medial hamsting muscle activation (Arborelius, Wretenberg & Lindberg 1992). However, foot position in the sagittal plane was not controlled making between seat height comparisons difficult to interpret. Combining biomechanical data with EMG of muscles surrounding both the hip and the knee would provide comprehensive evidence to evaluate the effectiveness of seating platforms with raised seat heights.

Lifting-seat devices have been designed to raise the effective seat height with the goal of reducing lower limb muscular requirements and joint loading during a STS for frail elderly populations with pathologies that compromise the musculoskeletal system including; arthritis, injury or surgery rehabilitation strokes, and neurological problems such as multiple sclerosis, stroke, and Parkinson's disease. Lifting-seat devices can be broadly divided into two main groups; a spring-loaded or pneumatic design (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000, Wretenberg et al. 1993) and an elevator lifting design (Jeyasurya 2011). The former uses pneumatics to create a mechanical preload as the device lowers to provide an assistive force while the user actively performs a STS transfer. The latter uses an electric-elevator design that transfers the user to a raised position before performing a STS transfer. No studies have compared lifting-seat devices with different lifting mechanisms, and thus, a lack of information is available with respect to who will benefit from which type of device.

These devices have been reported to reduce pain and perceived effort (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000) but despite their purported mechanical influence on the STS transfer, a lack of biomechanical or neuromuscular evidence exists to understand the effect of either type of lifting seat device on STS transfer. Of the evidence that exists, two studies have shown lifting-seat devices to be effective at reducing the hip moments, knee moments, and quadriceps activity (Jeyasurya 2011, Wretenberg et al. 1993), whereas the two other studies have shown no change in knee moments and quadriceps activity (Munro, Steele 2000, Munro et al. 1998). Conflicting evidence could be related to several issues. Firstly, devices used in these studies differ with respect to their lifting mechanisms. Secondly, non-standardized methodologies were employed, such as varying foot position and use of armrests that make comparisons between unassisted and assisted transfers and between studies difficult. Thirdly, subject populations vary across studies making interpretation difficult. Studying the effect of these devices on healthy populations with standardized methodologies will develop a comprehensive base-point for future investigation and scientific inquiry pertaining to the development and prescription of STS assistive devices.

There remain several key questions with respect to how both types of lifting-seat devices affect a STS transfer and who might benefit from these devices. Firstly, is the design goal of raised seat height effective at reducing the biomechanical demands at the hip and knee and EMG activity of surrounding muscles and is the effect similar in younger and older adults? Secondly, whether the effect of the lifting-seat devices on STS is similar to their design goal of a raised seat height or is some other effect on STS strategy? And lastly, are lifting devices with different lifting-mechanisms effective at lowering these demands, if not how do they differ, and are the effects altered by age?

The goal of this thesis was to understand how seat height and lifting-seat devices with different lifting mechanisms affect trunk, hip, and knee joint biomechanics and neuromuscular activity of surrounding muscles, and to determine whether the effects are altered by age. A better understanding of lifting-seat devices, particularly, how they compare to an unassisted STS transfer from a raised seat height provides objective evidence that can be used to guide future device modifications and inform guidelines for prescription.

1.1 Purpose and Objectives of the Thesis

The main purpose of this research was to evaluate the effect of two different lifting seat devices during a STS transfer on the trunk, hip and knee joint biomechanics and surrounding muscles, and determine if these effects were influenced by age.

This study compared a healthy population of younger and older adults. The study purpose was addressed through two main objectives. The dependent variables were:

i. Peak trunk, hip and knee flexion angles.
ii. Peak and impulse hip flexion moments and knee flexion moments.
iii. Peak and integrated muscle activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstrings, and gluteus maximus.

Objective 1:

Objective one determined the effect of seat height and age on trunk, hip and knee biomechanics and muscle activation. Two different seating conditions were compared; an unassisted STS transfer from a normal seat height and an unassisted STS transfer from a raised seat height.

Objective 2:

Objective two determined the effect of two lifting-seat devices with different lifting mechanisms compared to an unassisted STS transfer from a raised seat height and age on trunk, hip and knee biomechanics and muscle activation. Three different seating conditions were compared; an electric-elevator design lifting-seat device (Power Seat™), a pneumatic design lifting-seat device (Seat Assist™), and no device at a raised seat height.

1.2 HYPOTHESES

The hypotheses for Objective 1 were;

1a) Compared to unassisted STS at raised seat height, a normal seat height will have greater peak trunk, hip, and knee angles, greater peak and impulse hip and knee moments, and greater peak and integrated EMG activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstring, and gluteus maximus.

1b) Compared to younger adults, older adults will have similar kinematics and kinetics of the trunk, hip, and knee, and greater peak and integrated EMG activity for all muscles.

1c) Both seat heights will affect both groups equally for kinematic, kinetic, and EMG dependent variables.

The hypotheses for Objective 2 were;

2a) There will be no differences between the two different lifting-seat devices and an unassisted STS from a raised seat height on biomechanical and neuromuscular outcomes.

2b) Compared to younger adults, older adults will have similar kinematics and kinetics of the trunk, hip, and knee, and greater peak and integrated EMG activity for all muscles.

2c) All seating conditions will affect both groups equally for kinematic, kinetic, and EMG dependent variables.

1.3 THESIS STRUCTURE

This thesis consists of six chapters. Chapter 1 provides an introduction to the problem, the rational and purpose of the study, and main study objectives and hypotheses. Chapter 2 provides a comprehensive overview of relevant literature pertaining to methods used to analyze a STS transfer, several characterized STS strategies, determinants of a STS transfer, and the effects of age, seat height, and lifting-seat devices on STS biomechanics. Chapter 3 provides detailed methodology used to address the research objectives of this thesis. Chapters 4-5 are organized in self-contained journal format papers based on objectives 1 and 2. Chapter 4 addresses the first objective: to determine the effects of seat height and age on a STS transfer. Chapter 5 addresses the second objective: to determine the effects of two different lifting-seat devices and age on a STS transfer compared to an unassisted STS from a raised seat height. Chapter 6 concludes the thesis, containing a summary of the results, implications, and directions for future research.

CHAPTER 2 BACKGROUND LITERATURE

The following sections provide the pertinent background literature and the rationale for this study and the methods used. A discussion of key dependent variables and what needs to be considered in assessing STS transfer, as well as effects of age, seat height, and lifting-seat devices on STS transfers illustrate the gaps in our knowledge and how this study addressed these gaps.

2.1 SIT-TO-STAND – INTRODUCTION

A STS transfer is a physiologically and mechanically demanding task on the musculoskeletal system that is performed approximately four times per hour by healthy adults (McLeod et al. 1975). As a precursor to walking, transferring from a chair to a standing position is a fundamental activity of daily living. STS limitations can decrease an individual's mobility (Guralnik et al. 1995), leading to an increased risk of a loss of independence and mortality (Hirvensalo, Rantanen & Heikkinen 2000). Almost 45% of individuals with severe knee osteoarthritis enrolled in various research studies at the Dynamics of Human Motion Laboratory (Dalhousie University, Halifax, Nova Scotia) have reported severe to extreme difficulty 'rising from sitting' using the Western Ontario and McMaster Universities Arthritis Index questionnaire. Based on objective biomechanical findings, hip joint contact forces and knee flexion moments have been shown to be approximately double those during walking (Mizner, Snyder-Mackler 2005, Hodge et al. 1989). Older adults with functional limitations have been shown to use 97% of available knee extensor muscle strength when rising from a lower than normal seat height (Hughes, Myers & Schenkman 1996). Rehabilitation interventions, such as lifting-seat devices that are purported to lower the mechanical and musclular demands of a STS transfer have the potential to help older adults with lower limb impairments maintain a more independent and active lifestyle. The benefits of an sctive lifestyle include; lower risk of premature death, heart disease, stroke, colon and breast cancer, type-2 diabetes, improved muscular, bone, and joint health, as well as reduced depression (Physical Activity Guidelines 2008).

While devices have been designed and shown to be helpful at reducing pain and increasing function using self report assessments, our understanding of the mechanism and hence who they will benefit most from these devices is lacking. To gain a comprehensive understanding of the effect of these devices requires objective evidence examining lower limb kinematics, kinetics, and muscle activation.

2.2 METHODS FOR MEASURING SIT-TO-STAND

There are a variety of methods for quantifying biomechanics of a STS transfer, which include kinematics, ground reaction forces, kinetics, and surface EMG. Kinematics typically includes measures of limb segment position, orientation, velocity and acceleration. These data can be used to derive joint angles and angular displacement (motion) (i.e. knee joint angle of the tibia with respect to the femur). Kinetic analysis typically measures the forces and moments acting on the limb segments and joints. Inverse dynamics modeling is a common approach that combines ground reaction forces, subject anthropometric measures, and motion data to calculate the net moments acting on joints (Vaughan, Davis, O'Connor 1992). Surface EMG provides an extracellular view of changes in membrane potential as action potentials propagate along muscle fibers to gain information about muscle activation.

2.2.1 Kinematics

STS studies have examined joint motion by using axial goniometers or accelerometers (Boonstra et al. 2010, Patsika, Kellis & Amiridis 2011), whereas others have used motion capture systems (Yoshioka et al. 2009, Ashford, Hospital 2000, Crockett, Lanovaz & Arnold 2012, de Souza et al. 2011, Dehail et al. 2007, Epifanio et al. 2008, Farquhar, Kaufman & Snyder-Mackler 2009, Fujimoto, Chou 2012, Gillette, Stevermer 2012, Khemlani, Carr & Crosbie 1999, Kuo, Tully & Galea 2009, Mizner, Snyder-Mackler 2005, Savelberg et al. 2007, Van der Heijden et al. 2009, Schenkman, Rlley & Mann 1990, Su, Lai & Hong 1998, Turcot et al. 2012, Yoshioka et al. 2007). STS studies often measure joint motion at more than just a single joint of the lower limb, as hip, knee, and ankle motion together help to characterize overall STS strategy. Monitoring trunk motion can also be important, as trunk flexion has been shown to increase in the elderly and pathological populations (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Van der Heijden et al. 2009, Savelberg et al. 2007). Trunk flexion angle has been defined in a number of different ways. The most common methods are 1) the angle between vertical/horizontal plane and the line from the greater trochanter to a positional marker on the shoulder or near the center of mass (Munro et al. 1998, Jeyasurya 2011, Burnfield et al. 2012, Schenkman, RIley & Mann 1990), and 2) the angle between the femur (greater trochanter to lateral femoral epicondyle) and the line from the greater trochanter to a positional marker on the shoulder or near the center of mass (Doorenbosch et al. 1994, Dehail et al. 2007, Savelberg et al. 2007, Van der Heijden et al. 2009). In general, kinematic data provide information on movement strategies and joint motion but do not provide any information on the forces that are causing the motions of the limbs and body.

2.2.2 Ground Reaction Forces

In order to quantify the forces needed to rise from a chair, force plates are used to measure ground reaction forces (GRFs) (Christiansen et al. 2011, Christiansen, Stevens-Lapsley 2010, Patsika, Kellis & Amiridis 2011, Boonstra et al. 2010). For example, GRFs provide information on weight-bearing asymmetry and studies have shown that pathological populations load their unaffected leg more than their affected leg (Christiansen et al. 2011, Christiansen, Stevens-Lapsley 2010). Weight-bearing asymmetry can also be used to quantify the effects of treatments, such as total knee arthroplasty (Christiansen et al. 2011). In other studies, GRFs have been used to track the position of the center of pressure (COP), which has been used in studies looking at stability during the rise. For example, elderly patients have greater center of pressure excursion in the medial/lateral and anterior/posterior directions and take longer to reach final COP position, which have implications for stability during STS (Akram, McIlroy 2012). Patsika et al. (2011) found patients with knee osteoarthritis (OA) have a more posterior center of pressure during the STS, which the authors suggested would cause an increase in the knee flexion moments. GRFs can provide important information about the forces exerted during a STS transfer but they do not capture any information at the joint level.

2.2.3 Kinetics

Joint kinetics are frequently studied in STS studies to provide information on the external forces acting on the joints. Kinetic outcomes can give an indication of the amount of muscle force (internal forces) needed to overcome external forces acting on the joints and are of greatest in magnitude at or just after seat-off for the knee and hip joints (Pai, Rogers 1991, Doorenbosch et al. 1994, Jeyasurya 2011) Knee joint moments are of particular interest as knee extensor strength has been identified as a limiting factor in rising from a chair in pathological populations (Hughes, Myers & Schenkman 1996, Savelberg et al. 2007, Van der Heijden et al. 2009). However, measuring moments of the hip and the ankle are also important to characterize overall kinetic strategy. For example, an elderly population with quadriceps weakness may use a STS strategy with more trunk flexion to reduce the knee moments, while increasing the hip moments (Yoshioka et al. 2007). Monitoring knee joint moments are of particular importance when studying STS but measuring other joint moments, particularly the hip is also important to identify any STS kinetic strategy alterations.

Inverse dynamics modeling uses a variety of assumptions to calculate joint moments of force, most importantly, that only one muscle group is activated (Vaughan, Davis, O'Connor 1992). However, in older adults or those with pathologies, there is often agonist-antagonist co-activation during functional movements (Hubley-Kozey et al. 2006, Patsika, Kellis & Amiridis 2011, Nagai et al. 2011, Quirk, Hubley-Kozey 2012), which may cause an underestimation of muscle activity in agonist muscles and joint contact forces needed to perform a task, such as STS transfer. For example, increased activation amplitude of the knee flexor muscles when the knee extensors are the main agonist results in a greater knee flexion moment than estimated by inverse dynamics. The knee extensor muscles (agonists) will have to generate a greater knee extension moment to overcome the additional knee flexion moment caused by the antagonistic muscle activation of the hamstrings.

2.2.4 Surface Electromyography

Surface EMG has been employed in STS studies to measure the amount of muscle activity or timing of muscle activity, such as onset, offset, and duration (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Wretenberg et al. 1993, Dehail et al. 2007, Doorenbosch et al. 1994, Khemlani, Carr & Crosbie 1999, Burnett et al. 2011, Camargos, Rodrigues-de-Paula-Goulart & Teixeira-Salmela 2009, Patsika, Kellis & Amiridis 2011). EMG provides information on individual muscles that are activated to provide internal forces to overcome the external forces acting on the body during a STS transfer. EMG is not as frequently measured in STS studies compared to kinematics and kinetics but there have been a few studies to characterize the timing and magnitude of key muscles during a STS transfer. One of the first muscles to become active is the tibialis anterior (Dehail et al. 2007, Ashford, Hospital 2000, Goulart, Valls-sole 1999, Khemlani, Carr & Crosbie 1999) and often precedes trunk flexion (Dehail et al. 2007). The tibialis anterior has been shown to activate first regardless of feet forward or backward and is thought to be an anticipatory postural adjustment to stabilize the feet before the body's center of mass (COM) moves forward (Khemlani, Carr & Crosbie 1999). The next muscles that follow are involved in flexing the trunk forward and consist of the rectus abdominus (Ashford, Hospital 2000, Dehail et al. 2007, Goulart, Valls-sole 1999) and the rectus femoris (Khemlani, Carr & Crosbie 1999). The knee extensors (quadriceps), the hip extensors (hamstrings, gluteus maximus), and the back extensors (lumbar paraspinal muscles) turn on prior to seat-off (Ashford, Hospital 2000, Dehail et al. 2007, Goulart, Valls-sole 1999, Khemlani, Carr & Crosbie 1999). They have been identified as 'prime movers' during the STS task based on correlation of their onset times across subjects (Dehail et al. 2007, Goulart, Valls-sole 1999), which differ from the preparatory postural muscles, such as the tibialis anterior, rectus abdominus, sternocleidomastoid (Goulart, Valls-sole 1999). The erector spinae muscle has its maximal activity near the end of trunk flexion and thus, plays a role in slowing trunk flexion (Dehail et al. 2007). The gastrocnemius muscles and soleus become active immediately after seat-off and are considered a postural muscle important for standing stability (Ashford, Hospital 2000, Doorenbosch et al. 1994, Khemlani, Carr & Crosbie 1999, Goulart, Valls-sole 1999).

2.2.5 Dependent Variables

The majority of studies use peak values as the main dependent variables (Doorenbosch et al. 1994, Hughes, Myers & Schenkman 1996, Lundin et al. 1995, Khemlani, Carr & Crosbie 1999, Pai, Rogers 1991, Rodosky, Andriacchi & Andersson 1989, Shepherd, Koh 1996, Jeyasurya 2011, Arborelius, Wretenberg & Lindberg. 1992, Farquhar, Kaufman & Snyder-Mackler 2009, Gillette, Stevermer 2012), which are important values in STS transfer task as individuals must overcome this magnitude of joint external moments or produce this level of muscle activity to rise from a chair. There have also been a few studies looking at muscle onset, offset, and total duration, which provides information on the temporal firing patterns of different muscles but does not provide any information on muscle activation amplitude (Dehail et al. 2007, Ashford, Hospital 2000, Goulart, Valls-sole 1999, Khemlani, Carr & Crosbie 1999, Munro, Steele 2000). There has been less work analyzing moment impulses or integrated EMG during the task that account for both amplitude and duration to give an indication of the overall joint loading and muscle effort during the task. To highlight the shortfalls of only analyzing peak measures or muscle onset/offset, Munro et al. (1998 & 2000) showed similar peak COM horizontal velocity between an unassisted STS and assisted STS with a lifting-seat device but earlier and more prolonged activation of the quadriceps and triceps muscles. Based on these measures, they had a mix of conclusions about whether or not the lifting-seat device caused a destabilizing horizontal force. Integrated EMG would have provided additional information about whether or not greater overall EMG activation was required with the lifting-seat device to maintain similar COM velocity.

2.2.6 STS Symmetry

In order to simplify analyses, most studies assume bilateral symmetry. Burnett et al. (2011) showed no significant differences between dominant and non-dominant limbs in vertical ground reaction forces or quadriceps muscle activity during STS in 35 healthy adults. Leg dominance was determined by the "leg they use to kick a ball". Lundin et al, (1995), investigated the validity of bilateral symmetry assumption for joint kinetics and found statistically significant differences in peak hip and knee moments between right and left legs. However, the differences were small in magnitude and the authors concluded they were of limited biomechanical significance. Assumption of symmetry may not be valid for individuals with unilateral pathologies that have been shown to load their unaffected limb more than their affected limb (Christiansen, Stevens-Lapsley 2010, Christiansen et al. 2011, Turcot et al. 2012).

In summary, to comprehensively characterize performance during a STS transfer, kinematics analysis is needed to assess joint motion and movement strategies, kinetic analysis is needed to assess joint loading, and EMG analysis is needed to assess muscle activation. Few studies have analyzed a STS transfer using all three analyses. This is important as inverse dynamics modeling used to calculate joint moments of force does not consider muscle co-activation and older adults have been shown to have higher antagonist-agonist co-activation during other functional activities (Nagai et al. 2011, Quirk, Hubley-Kozey 2012). Examining knee moments of force and muscle activity of the quadriceps are of particular importance but analyzing the moments of force and muscle activity at the hip joint help characterize tradeoffs between joints and overall STS strategy. Peak measures give an indication of the maximum moments of forces and maximum muscle activity needed to perform a STS transfer, related to peak joint loading and peak percentage of maximum muscle strength capability (Munro, Steele 2000, Munro et al. 1998, Doorenbosch et al. 1994, Hughes, Myers & Schenkman 1996, Lundin et al. 1995, Khemlani, Carr & Crosbie 1999, Pai, Rogers 1991, Rodosky, Andriacchi & Andersson 1989, Shepherd, Koh 1996, Jeyasurya 2011, Arborelius, Wretenberg & Lindberg. 1992, Farquhar, Kaufman & Snyder-Mackler 2009, Gillette, Stevermer 2012). Moment impulses and integrated EMG are less frequently analyzed, providing an indication of overall joint loading and muscle effort required during the task, which accounts for both magnitude and duration (Winter 2009, Robbins et al. 2009).

2.3 Sit-to-Stand Events and Phases

STS events separate a STS transfer into different phases of interest. STS studies vary in the number of events and phases used and largely depend on the research question and which phase(s) is needed to address study objectives. The selection of events is highly variable and non-standardized, which is surprising due to the numerous STS studies that exist.

The phase that is most commonly analyzed is from seat-off to standing, which is the period where the greatest lower limb muscle activity and joint moments occur as the center of mass moves from a stable three-point sitting position to a more unstable two-point standing position (Pai, Rogers 1991, Doorenbosch et al. 1994, Jeyasurya 2011, Goulart, Valls-sole 1999). Seat switches used to determine when buttocks lose contact with the seat are easy to employ and widely used to determine seat-off in STS studies (Munro et al. 1998, Goulart, Valls-sole 1999, Turcot et al. 2012, Christiansen, Stevens-

Lapsley 2010, Papa, Cappozzo 2000). However, standing event determination is much more variable. Standing event methods include; i) when knee extension angular velocity equals zero (Epifanio et al. 2008), ii) when hip extension angular velocity equals zero (Akram, McIlroy 2012, Schenkman, Rlley & Mann 1990, Ikeda et al. 1991), iii) when hip marker linear horizontal velocity or center of mass momentum equals zero (Khemlani, Carr & Crosbie 1999, Pai, Rogers 1991, Munro et al. 1998, Munro, Steele 2000, Shepherd, Koh 1996), iv) when the vertical velocity of a kinematic marker or segment equals zero (Kuo, Tully & Galea 2009, Turcot et al. 2012), v) when the vertical ground reaction force rate of change equals zero (Etnyre, Thomas 2007, Jeyasurya 2011). Of the proposed standing event methods, it is important to use one that consistently and accurately identifies standing for all study conditions and groups, particularly when calculating dependent variables that account for duration, such as integrated EMG and moment impulse. A comparison of the most common standing event methods is included in Appendix D.

2.4 Sit-to-Stand Transfer Strategies

The fundamental movement of STS transfer takes a person from a stable three-point position (both feet contacting the floor and buttocks on the chair) with flexed limbs to a more unstable two-point position on two extended legs. STS transfer is demanding on the neuromuscular system, and several different strategies exist to overcome the external joint moments and maintain the center of mass within its base of support.

Several different strategies have been characterized and examined in the literature. Scarborough et al. (2007) qualitatively describes three STS strategies, which are primarily characterized by the degree of trunk flexion. The ‘momentum-transfer’ strategy is one where some of the forward momentum from the upper-body is transferred to vertical momentum when the person lifts the buttocks off the chair with simultaneous back and knee extension (Scarborough, McGibbon & Krebs 2007, Scarborough, Krebs & Harris 1999, Schenkman, Rlley & Mann 1990). The ‘stabilization’ strategy is often used by older adults with functional limitations, such as muscle strength, in which they flex their trunk to place the center of mass over the feet before lifting off the seat with delayed trunk extension. In this strategy, lift-off from the seat is accomplished without assistance from horizontal momentum (Scarborough, McGibbon & Krebs 2007, Scarborough, Krebs & Harris 1999, Schenkman, Rlley & Mann 1990). Lastly, the dominant vertical rise strategy stops any limited trunk flexion immediately after seat-off and the participant lifts with predominantly vertical movement with trunk extension following knee and hip extension (Scarborough, McGibbon & Krebs 2007).

In the study by Scarborough et al. (2007), 95 older adults with functional limitations had their STS strategy characterized through visual observation. 65 used a momentum transfer strategy, 16 used a stabilization strategy, and 14 performed a dominant vertical rise strategy (Scarborough, McGibbon & Krebs 2007). There were no differences of age, gender, body mass index (BMI), or self-reported functional limitations between STS strategies. Quantitatively, trunk flexion angle was greatest for the stabilization strategy, followed by the momentum transfer strategy, and lowest for the dominant vertical rise strategy. The knee moments followed an opposite pattern with the lowest values for the stabilization strategy and the greatest for the dominant vertical rise strategy. Similarly, Yoshioka et al. (2007), found trunk flexion angle had a positive relationship with hip moments and inverse relationship with knee moments using a computer simulation model. The trunk segment was more upright in the STS strategy that had the lowest hip moment and greatest knee moment, whereas the trunk was more flexed in the STS strategy that produced the greatest hip moment and the lowest knee moment (Yoshioka et al. 2007).

Two studies have compared two different STS strategies; 1) self-selected strategy and 2) exaggerated trunk flexion prior to seat-off to simulate the 'stabilization' strategy (Doorenbosch et al. 1994, Fujimoto, Chou 2012). With trunk fully flexed, the COM was positioned more anteriorly (Doorenbosch et al. 1994, Fujimoto, Chou 2012), which was shown to decrease the knee moments and activity of rectus femoris (Doorenbosch et al. 1994). However, the hip and ankle moments were greater along with the hip extensors (gluteus maximus and hamstrings) and plantarflexor muscles (gastrocnemius and soleus), suggesting a potential trade-off from the knee joint to the hip and ankle joints (Doorenbosch et al. 1994). In addition, a more flexed trunk caused lower COM momentum and acceleration at seat-off (Fujimoto, Chou 2012).

There have been several studies suggesting that muscle deficits cause individuals to adopt the 'stabilization strategy' with more trunk flexion (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Van der Heijden et al. 2009, Savelberg et al. 2007). Two studies simulating the effects of muscle deficits were conducted in a younger population by means of adding weight to a vest (Savelberg et al. 2007, Van der Heijden et al. 2009). Ven der Heijden (2009) showed participants changed their STS strategy from a 'momentum-transfer' to a 'stabilization' strategy with an increase of trunk flexion of 11 degrees when using a weight vest set to 45% of the participant's body mass. The 'stabilization' strategy had higher hip and ankle moments but a reduction of knee moments. This study suggests that with added weight (simulated muscle deficits), the knee extensors are the first muscles to become overloaded, thus forcing a change in strategy from a 'momentum-transfer' to a 'stabilization' strategy that reduces knee moments but increases the hip and ankle moments. They found that the total impulse from the hip, knee, and ankle combined was higher for the 'stabilization' strategy, which suggests the 'momentum-transfer' strategy is the more efficient strategy but only up until the point when the knee extension moment overloads the knee extensor muscles, at which point the 'stabilization' strategy is preferred (Van der Heijden et al. 2009).

In a similar experimental design, Savelberg et al. (2007) incrementally added weight to a weight vest to simulate muscle impairments. As the load was increased, the strategy changed to a 'stabilization strategy' with more trunk flexion. However, the change did not occur gradually but with the majority of change occurring from an increase of 30% to 45% of the participant's body weight, where the joint moments were redistributed from the knee joint to the hip joint with accompanying increase in the biceps femoris muscle. This study supports the findings by Van der Heijden et al. (2009), that the strategy change only occurs when the knee extensors become overloaded, which occurs around 45% of the participant's body weight for the nine healthy adult females used in this study (Savelberg et al. 2007).

In summary, the most common method to characterize STS strategies is the degree of trunk flexion. The degree of trunk flexion plays an important role in the position of the COM, hence the location of the ground reaction force vector in the sagittal plane and the demands on the lower limb joints and muscles. Trade-offs exist amongst

STS strategies on the demands on the knee and the hip joints. More trunk flexion reduces the demands on the knee joint but increases the demands on the hip joint. It is clear from these studies that age and quadriceps muscle strength influences the strategy used. STS strategy and trunk flexion need to be considered in musculoskeletal rehabilitation and in designing assistive devices to reduce the demands on either the hip or knee joints, depending on an individual's pathology.

2.5 DETERMINANTS OF STS TRANSFER

Determinants of STS are "factors influencing how the movement is performed… and …should be independent from the techniques used to study movement." (Janssen, Bussmann & Stam 2002). The authors of an extensive review paper on the determinants of STS suggest that knowledge of the determinants is necessary in order to conduct research on STS or to interpret results of reported studies because the results can be a function of a determinant (Janssen, Bussmann & Stam 2002). The determinants can be separated into three different categories; chair-related, subject-related; and strategy-related (Janssen, Bussmann & Stam 2002). Manipulation of determinants affect STS transfer and should be standardized when comparing between different subject populations or between different seating conditions, such as unassisted STS transfer compared to assisted STS transfers with lifting-seat devices. Several key determinants that have been studied in detail including; use of arms, foot position, and speed are discussed in this section. Age and seat height are also identified as determinants of STS (Janssen, Bussmann & Stam 2002) and are discussed in detail in sections 2.5 and 2.6, respectively.

2.5.1 Foot Position

Foot position has been shown to alter STS strategies and demands on lower limbs (Akram, McIlroy 2012, Khemlani, Carr & Crosbie 1999, Gillette, Stevermer 2012, Goulart, Valls-sole 1999). In studies comparing feet in a more anterior position to those in a more posterior position, the more anterior position was shown to increase STS duration (Akram, McIlroy 2012, Khemlani, Carr & Crosbie 1999) with a greater anterior/posterior ground reaction force (Akram, McIlroy 2012, Gillette, Stevermer 2012), greater hip flexion and peak angular velocity (Khemlani, Carr & Crosbie 1999), greater hip moments (Gillette, Stevermer 2012), and decreased ankle moments (Khemlani, Carr & Crosbie 1999, Gillette, Stevermer 2012). In studies examining muscle activity, feet placed more anteriorly caused earlier onset of gastrocnemius and soleus muscles (Khemlani, Carr & Crosbie 1999), delayed onset of the tibialis anterior (Goulart, Valls-sole 1999, Khemlani, Carr & Crosbie 1999), and greater activity of the soleus (Khemlani, Carr & Crosbie 1999), abdominals, sternocleidomastoid, and trapezius muscles (Goulart, Valls-sole 1999). Farquhar et al. (2009) compared constrained foot position with unconstrained and the constrained position resulted in greater hip flexion, reduced knee flexion, and less medial gastrocnemius activity. This study also found that asymmetries in pathological and non-pathological limbs were more pronounced with feet in constrained position (Farquhar, Kaufman & Snyder-Mackler 2009).

Most studies have examined alterations of foot position in the sagittal plane (i.e. feet forward, feet backward, feet staggered) (Akram, McIlroy 2012, Khemlani, Carr & Crosbie 1999, Goulart, Valls-sole 1999, Gillette, Stevermer 2012). Although frontal and transverse plane angles of the foot are not expected to significantly alter the sagittal plane motion and moments, constraining foot position in these planes should be considered and is often employed in STS studies (i.e. feet parallel to each other or symmetrical) (Boonstra et al. 2010, Doorenbosch et al. 1994, Ikeda et al. 1991, Patsika, Kellis & Amiridis 2011, Rodosky, Andriacchi & Andersson 1989, Savelberg et al. 2007, Van der Heijden et al. 2009, Schenkman, Rlley & Mann 1990, Schenkman, Riley 1996).

2.5.2 Use of Arms

The use of the upper extremities, particularly when using armrests play an important role during a STS transfer in offloading the lower extremities. Use of arms during a STS has been shown to reduce the knee moments (Seedhom, Terayama K 1976, Arborelius, Wretenberg & Lindberg. 1992, Burdett et al. 1985) and hip moments (Burdett et al. 1985). More specifically, use of arms has been shown to reduce hip moments by about 50% and knee moments by about 60% (Arborelius, Wretenberg & Lindberg. 1992). Use of arms does not change the range of motions of the hip, knee, or ankle during a STS (Burdett et al. 1985). In order to avoid the effect of upper extremities, many studies instruct participants to fold their arms across their chests but other studies have them place them on their hips or thighs. Both use and positioning of the upper extremities impact the STS transfer and should be considered in STS methodology.

2.5.3 Speed

Pai and Rogers (1991) compared the joint motion and moments of the lower-limb when rising with self-selected speed, slower than self-selected, and faster than self-selected. At greater speeds, the knee flexion moments increased but the hip flexion moments and plantarflexion moments and the overall hip and knee motion did not significantly differ (Pai, Rogers 1991). Yoshioka et al. (2009) did a computer simulation model of increasing speed of STS on the joint moments and found that for fast and moderate speed (less than 2.5 seconds), the hip flexion, knee flexion, and plantar flexion moments increased exponentially with increasing speed but at speeds higher than 2.5 seconds, the joint moments stayed relatively constant (Yoshioka et al. 2009).

Speed of rise is typically self-selected in experimental design (Epifanio et al. 2008, Chen et al. 2010, Pai, Rogers 1991) but some studies have attempted to control the speed by use of a metronome (Schenkman, Rlley & Mann 1990, Ikeda et al. 1991). STS speed primarily affects the lower limb joint moments of force (Pai, Rogers 1991, Yoshioka et al. 2009) and needs to be considered either in the study methodology or when comparing between groups and conditions.

Use of arms, foot position, and STS speed are three determinants of STS. Manipulation of these constraints affects STS transfer and should be standardized when comparing between different subject populations or between different seating conditions.

2.6 Effect of Age on a STS Transfer

Age is a subject-related determinant that can effect kinematic and kinetics strategies during a STS transfer (Janssen, Bussmann & Stam 2002). STS duration has been shown to be similar between older and younger adults at self-selected speed (Akram, McIlroy 2012, Papa, Cappozzo 2000) but not at higher speeds due to inability to generate joint moments as a result of reduced muscle capacity (Papa, Cappozzo 2000). Kinematic and kinetics variables, such as knee moments (Hughes, Myers & Schenkman 1996, Ikeda et al. 1991), hip moments, hip and knee angular velocities, and peak ankle, knee, hip, and trunk angles, as well as the timing of the kinematic and kinetic variables have been shown to be similar amongst older and younger adults (Ikeda et al. 1991). Despite these similarities, knee moments as a percentage of knee extensor strength have been shown to be 97% in older adults with functional impairments compared to 39% in younger adults while rising from a lower than normal chair height (Hughes, Myers & Schenkman 1996). Given decreased strength with age measured by isometric strength testing (Hughes, Myers & Schenkman 1996, Fujimoto, Chou 2012, Gross et al. 1998), similar increases in muscle activity, as a percentage of maximum would be expected. However, no studies could be found that have compared muscle activation during a STS transfer between younger and older adults, which is important as older adults have been shown to have higher antagonist-agonist co-activation during other functional activities (Nagai et al. 2011, Quirk, Hubley-Kozey 2012).

There have been several studies suggesting that older adults with muscle deficits use a 'stabilization strategy' with more trunk flexion (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Van der Heijden et al. 2009, Savelberg et al. 2007, Gross et al. 1998). Older adults have been shown to have a greater peak horizontal trunk and COM velocity than the young adults prior to seat-off but at seat-off, both groups showed similar COM positions and velocities (Fujimoto, Chou 2012). Additional postural differences were found in older adults, including less trunk-to-pelvis flexion, and greater head-to-trunk flexion (Ikeda et al. 1991). Differences between older and younger adults also exist near the termination of the STS movement. Older adults have also been shown to have a longer stabilization phase (after the hip was extended to 0 degrees) with a greater COP excursion in the anterior/posterior and medial/lateral directions, which suggests older adults have greater difficulty controlling stability at the end of a STS transfer compared to younger adults (Akram, McIlroy 2012).

Overall, STS kinematics and kinetics of the lower limbs are similar between older adults and younger adults. However, the effect of age on muscle activation of the lower limb muscles has yet to be established, which is important as older adults have been shown to have been shown to have higher antagonist-agonist co-activation during other functional activities (Nagai et al. 2011, Quirk, Hubley-Kozey 2012). For example, increased activation amplitude of the hamstrings (antagonists) during knee extension will cause a greater knee flexion moment than calculated by inverse dynamics, of which the knee extensors (agonists) must produce. Some older adults, particularly those with muscle strength deficits have been shown to change their STS strategy (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Van der Heijden et al. 2009, Savelberg et al. 2007, Gross et al. 1998)and to have difficulty controlling stability compared to healthy younger adults (Akram, McIlroy 2012). Therefore, older adults may respond differently to rehabilitative interventions designed to facilitate STS transfers, such as a raised seat height or lifting-seat devices but have yet to be established. This is important as these interventions are targeted towards older adults and those with pathologies.

2.7 EFFECT OF SEAT HEIGHT ON A STS TRANSFER

Raising the seat height has been shown to alter trunk and lower limb biomechanics during a STS transfer (Arborelius, Wretenberg & Lindberg. 1992, Munro et al. 1998, Su, Lai & Hong 1998, Schenkman, Riley 1996, Gillette, Stevermer 2012). Compared to a normal seat height, a raised seat has been shown to decrease trunk, hip, and knee angular velocity and displacement (Munro et al. 1998, Schenkman, Riley 1996, Kuo, Tully & Galea 2009, Rodosky, Andriacchi & Andersson 1989, Su, Lai & Hong 1998), and hip and knee flexion moments (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi & Andersson 1989). Peak knee flexion moments have been shown to be 60% less and the peak hip flexion moments 50% less from a raised seat height with knee flexion angle of 30° compared to a normal chair height with knee flexion angle of 90° (Arborelius, Wretenberg & Lindberg. 1992). Similar decreases in muscle activation would be expected but there has been limited evidence on the effect of seat height using EMG, which is important as inverse dynamics modeling used to calculate joint moments of force does not consider antagonist-agonist muscle co-activation. Two studies using EMG have shown a raised seat height reduces muscle activation of the quadriceps and triceps brachii muscles (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000). One study showed no change in muscle activation of the medial hamstring with seat height(Arborelius, Wretenberg & Lindberg 1992). However, foot position in the sagittal plane was unconstrained making between seat height comparisons difficult to interpret. Overall, there is limited information on the effect of seat height on muscles used to extend the hip, such as the gluteus maximus and the hamstrings, which have also been identified as prime movers during a STS (Dehail et al. 2007, Goulart, Valls-sole 1999). Establishing the effects of raised seat height on neuromuscular demands of the musculature surrounding the hip and the knee is important to affirm biomechanical findings, providing further comprehensive evidence that seating platforms with raised seat heights are effective rehabilitation devices for individuals with STS limitations.

Only two studies have previously studied the effect of seat height between healthy older and younger adults (Chen et al. 2010, Schenkman, Riley 1996) providing evidence that both groups were affected similarly to different seat heights in terms of subjective ratings (Chen et al. 2010) and angular excursions and velocities (Schenkman, Riley 1996). No studies were found that have examined the effect of seat height between healthy older and younger adults on joint moments or neuromuscular outcomes, which is important in establishing if the effects of a raised seat height on joint and muscle demands are altered by age.

Combining biomechanical data with EMG of muscles surrounding both the hip and the knee would provide comprehensive evidence to evaluate the effectiveness of seating platforms with raised seat heights as a rehabilitative intervention for individuals with STS limitations from aging or pathology. Lifting-seat devices have been developed with the design goal of raising seat height to facilitate STS transfers from seating platforms of normal heights.

2.8 Effect of Lifting-Seat Devices on a STS transfer

There have been several different types of assistive devices developed that aim to reduce the lower limb joint forces and muscular requirements during a STS transfer. Edlich et al. (2003) characterized adaptive seating systems in three main categories 1) systems without a mechanical lift (i.e. raised seat height, hand rails, etc.), 2) Systems with mechanical lifts (i.e. ejector seats, spring-loaded flap-seats). 3) Systems that lift, tilt, recline, or rock (i.e. elevator chairs) (Edlich et al. 2003). Jeyasura et al. (2011) compared different assistive devices, and found that a lifting-seat design had the best performance in terms of biomechanical outcomes (stability measures and peak knee flexion moments) and subjective outcomes (stability and effort) compared to arm, bar, and waist designs.

In general, the purpose of lifting-seat devices is to raise the effective seat height, which has been shown to reduce the knee flexion moments (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi & Andersson 1989) and muscle activity of the quadriceps (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000). Lifting-seat devices can be broadly divided into two main groups; a spring-loaded or pneumatic design (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000, Wretenberg et al. 1993) and an elevator lifting design (Jeyasurya 2011). The former uses pneumatics to create a mechanical preload as the device lowers to provide an assistive force while the user actively performs a STS transfer. The device will only lower if the weight of the individual is greater than the seat's adjustable force setting. Therefore, in order to rise, the subject will need to off-load the seat in some capacity by the use of arms or by flexing the trunk forward (decreasing the moment arm from the anterior seat hinge) The latter uses an electric-elevator design that transfers the user to a raised position before performing a STS transfer. It is important to differentiate the lifting-seat devices based on their lifting mechanisms. There have been several studies on devices using a spring-loaded or flap-seats (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000, Wretenberg et al. 1993) but there has only been one study found that examined chairs with an electric-elevator design (Jeyasurya 2011). No studies have compared lifting-seat devices with different lifting mechanisms, and thus, a lack of information is available with respect to how they differ and who will benefit from which type of device.

Lifting-seat devices, such as those developed by Uplift Technologies Inc. can have other design features that significantly affect lifting mechanism and consequently, the effect on a STS transfer. Most lifting-seat devices are designed to pivot about a single hinge at the anterior portion of the chair (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000, Wretenberg et al. 1993). Along with an upward force, this type of device also provides an accompanying anterior force vector, which may provide a potentially dangerous destabilizing perturbation (Young 1972). Uplift Technologies Inc. has developed two lifting-seat devices that incorporate a two-hinge design; one at the anterior portion of the chair and one near the middle. This design feature allows the front of the chair to tilt incrementally while the rear portion of the chair is flatter during the rising period in attempt to lessen the horizontal force but has yet to be studied.

These devices have been reported to reduce pain and perceived effort (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000) but despite their mechanical influence on the STS transfer, a lack of biomechanical or neuromuscular evidence exists to understand the effect of either type of lifting seat device on STS transfer. Five studies have compared assisted STS transfers with lifting-seat devices to unassisted STS transfers from normal seat heights and of these, only four provided kinetic or neuromuscular analyses (Bashford et al. 1998, Jeyasurya 2011, Munro, Steele 2000, Wretenberg et al. 1993). Two studies have used both kinetics and EMG to study the effect of a lifting-seat device on a STS transfer (Munro et al. 1998, Munro, Steele 2000). However, only the quadriceps muscles were analyzed. Muscles used to extend the hip, such as the gluteus maximus and the hamstrings, which have also been identified as prime movers during a STS have not been studied (Dehail et al. 2007, Goulart, Valls-sole 1999). There is a need to understand how both types of devices affect the hip and knee joints and surrounding muscles by comprehensively studying the STS transfer with and without lifting-seat devices using EMG to assess muscle activation, and a biomechanical analysis of kinetics to assess joint moments of force, and kinematics to assess motion and transfer strategies.

Of the evidence that exists, two studies have shown lifting-seat devices to be effective at reducing the hip moments, knee moments, and quadriceps activity (Jeyasurya 2011, Wretenberg et al. 1993), whereas the other two studies have shown no change in knee moments and quadriceps activity (Munro, Steele 2000, Munro et al. 1998). Conflicting evidence could be a related to non-standardized methodologies used in these studies, such as varying foot position and use of armrests that make comparisons between unassisted and assisted transfers and between studies difficult. For example, arm assistance was used in some studies (Munro et al. 1998, Munro, Steele 2000), whereas in others it was not (Jeyasurya 2011, Wretenberg et al. 1993) and one study permitted participants to use additional aids, such as canes or walkers (Bashford et al. 1998). Additionally, subject populations vary across studies making interpretation difficult. (Table 2.1). Establishing comprehensive data on healthy populations with standardized methodology is important before studying their effect on pathological populations. Additionally, different types of lifting-seat devices have been studied and how they differ remains unclear.

Table 2.1. Subject populations in studies assessing the effect of lifting-seats on STS transfer.

Study	Subject Population
Wretenberg et al. (1993)	9 healthy males (20-32 years), 8 severe medial knee osteoarthritis (65-79 years, 4 M/F)
Bashford et al. (1997)	7 knee osteoarthritis, 3 knee rheumatoid arthritis, 2 myopathies, 1 paraparesis, 1 cerebrovascular accident, 1 Parkinson's disease, 1 brain tumour (16 total)
Munro et al. (1998 & 2000)	12 elderly females (mean 65.5 +- 8.6 years) with rheumatoid arthritis.
Jeyasurya (2010)	17 healthy older adults (>60 years)

Overall, there is limited comprehensive evidence of how lifting-seat devices affect STS transfers and as a result, there remain several key questions with respect to how both types of lifting-seat devices affect a STS transfer. Specific to the present study, are these devices effective at lowering the demand on the hip and knee joints and musculature surrounding both joints? Secondly, are the effects similar in young and older adults? Thirdly, do lifting-seat devices with different lifting mechanisms have the same effect on a STS transfer? Lastly, whether the effect of the lifting-seat devices on STS is simply a result of an increase in seat height or some other effect on STS strategy?

CHAPTER 3 GENERAL METHODOLOGY

This was a cross sectional comparative study with an experimental manipulation. Each participant attended one testing session at the Dynamics of Human Motion laboratory at Dalhousie University. Bilateral lower extremity and trunk three-dimensional motion, ground reaction force data and surface EMG from five lower extremity muscle sites were collected while the participants completed five trials of each STS condition; 1) no device from normal seat height (ND-normal), 2) Seat Assist™ (SA), 3) Power Seat™ (PS), and 4) no device from raised seat height (ND-raised). The methodology is described in more detail in the following sections.

3.1 General Participant Recruitment

This study population consists of 10 healthy older adults (above 65 years) and 10 healthy younger adults (20-30 years) recruited from the Dalhousie University community and surrounding area. They were considered healthy based on the absence of any cardiovascular, neurological (i.e. stroke, Parkinson's disease, myocardial infarct, arrhythmias) or musculoskeletal issues (i.e. osteoarthritis, major surgeries) that would place them at risk of injury or that would alter their ability to perform a STS transfer. Participants were recruited using posters and advertisements on university and public access forums. Written consent was obtained in accordance with Dalhousie Research Ethics Board (Ethics Approval #2011-2508).

3.2 General Data Collection Procedure

An overview of the experimental protocol is illustrated in Figure 3.1. Detailed information on each component is provided in the following sections.

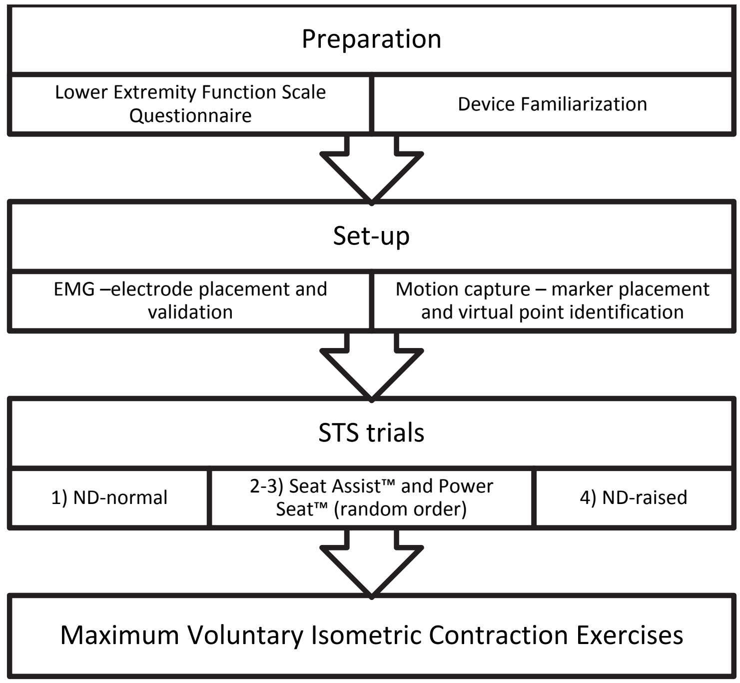


Figure 3.1. Overview of experimental protocol

3.2.1 Preparation

Upon arrival to the Dynamics of Human Motion Laboratory, each participant was introduced to the lab testing equipment and study protocol. Participants were instructed to familiarize themselves with both lifting seat devices by reading user manuals and practicing until they were comfortable with both devices. Participants completed the Lower Extremity Functional Scale (LEFS) (Appendix C.1), which is a self-reported functional status measure questionnaire that has been validated and shown to be valid and reliable (Binkley et al. 1999). Standard anthropometric measures were recorded including; height, mass, hip and waist circumferences, and bilateral knee heights, foot widths, thigh and calf circumferences. Leg dominance was determined by asking each participant "Which leg do you use to kick a ball?" (Burnett et al. 2011).

3.2.2 Set-up

Subjects wore spandex shorts for unobstructed placement of surface EMG electrodes and motion capture infrared emitting diode (IRED) skin surface markers. Surface EMG preparation and collection protocols were based on previous studies (Hubley-Kozey et al. 2006, Rutherford, Hubley-Kozey & Stanish 2011) and were consistent with the International Society of Electrophysiology and Kinesiology and SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines (Stegeman, Hermens 1999). One researcher trained in EMG techniques applied the electrodes. Prior to electrode placement, skin was prepared by light shaving with a disposable razor and abrading with 70% alcohol wipes. Disposable silver/silver-chloride (Ag/AgCl) surface electrodes (10mm diameter) with 20mm inter-electrode distance (3M™, Red Dot™, Repositionable Monitoring Electrodes, St.Paul, MN, USA), were placed in a bipolar configuration along the muscle fibre orientation of the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH), and gluteus maximus (GM) (Table 3.1). One ground electrode was placed over the tibia shaft. Validation of the EMG signal and proper gain adjustment was obtained using muscle palpation and a series of muscle specific isometric contractions (Winter, Fuglevand & Archer 1994).

Table 3.1. Electrode placement for the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH), and gluteus maximus (GM)

Muscle	Electrode location
VL	75% of the distance between the anterior superior iliac spine and the lateral joint line of the knee at 45° knee flexion
VM	80% of the distance between the anterior superior iliac spine and the medial joint line of the knee at 45° knee flexion
RF	50% of the distance from the anterior superior iliac spine and the superior part of the patella
LH	50% between the ischial tuberosity and the fibular head
GM	50% on the line between the sacral vertebrae and the greater trochanter

For motion capture, IRED skin surface markers were affixed to the lateral aspect of both lower extremities and trunk based on previous work (Landry et al. 2007, McKean et al. 2007, Rutherford et al. 2008). Single IRED markers were positioned on the skin at the greater trochanter and lateral epicondyle of the femur, the lateral malleolus, and the lateral aspect of the shoulder. Clusters of three markers on a rigid body (triad) were placed roughly midway on the foot, shank, thigh, the sacrum of the pelvis, and the thoracic spine (approximately over the spinous process of thoracic vertebra) (Figure 3.2).

Other important anatomical points were collected as virtual points using a digitizing probe with the subject standing in a neutral position including right and left thoracic spine, right and left anterior superior and posterior superior iliac spines, medial epicondyle of the femur, fibular head, tibial tuberosity, medial malleolus, base of the second metatarsal and center of the posterior calcaneous.

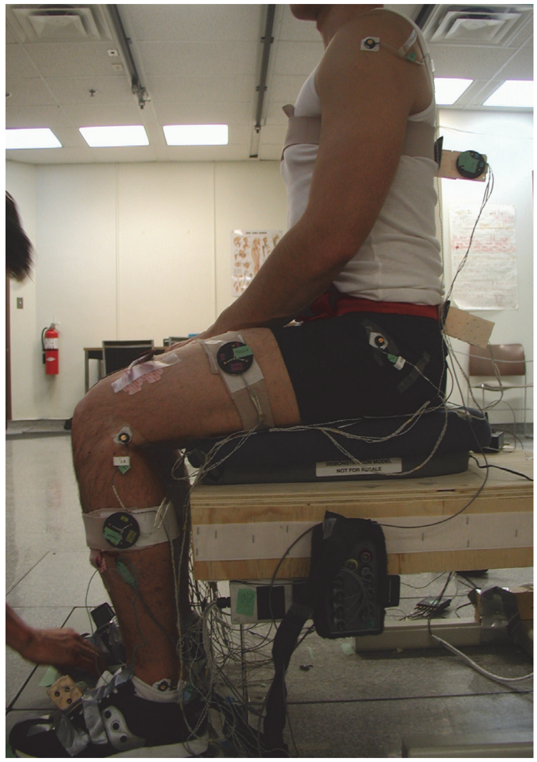


Figure 3.2. Infra-red light-emitting diode single marker and triad set-up on left lower extremity

3.2.3 Sit-to-Stand Trials

Each participant performed five trials of four different seating conditions with an armless, backless chair with an adjustable seat height; 1) ND-normal, 2) SA (Seat Assist™ UPE-1, Uplift Technologies Inc., Dartmouth, NS, Canada), 3) PS (Power Seat™ UPE-P100Ex 24VDC, Uplift Technologies Inc., Dartmouth, NS, Canada), 4) ND-raised. Before every ND-Normal, PS, and SA trial, the participant was positioned on the chair with the knee angle positioned at 90° using a standard goniometer and the feet positioned parallel and approximately shoulder width apart. The participant performed several trials to ensure they were performing each condition at a consistent self-selected speed and movement strategy. Participants were instructed to remain on the seat pan in an erect posture facing forward until a signal was given to begin the STS trial. In addition, they were instructed to fold their arms across their chest, keep feet on the floor, and face forward until trial completion. The ND-normal condition was collected first with the participant sitting on the seat pan of Power Seat™ device. The two lifting-seat conditions followed in a random order. The ND-raised trials followed with the participant seating on the Power Seat™ in the extended position. For the PS trials, participants pulled the lever to engage the seat when the signal was given. Participants were instructed to keep their feet planted on the ground and remain on the seat until maximum elevation and stand up at self-selected pace after the Power Seat™ stops. For the SA trials, the participants were instructed to perform a comfortable STS movement with self-elected speed and movement strategy. For all conditions, once standing, participants remained standing for the duration of the trial recording and looking forward.

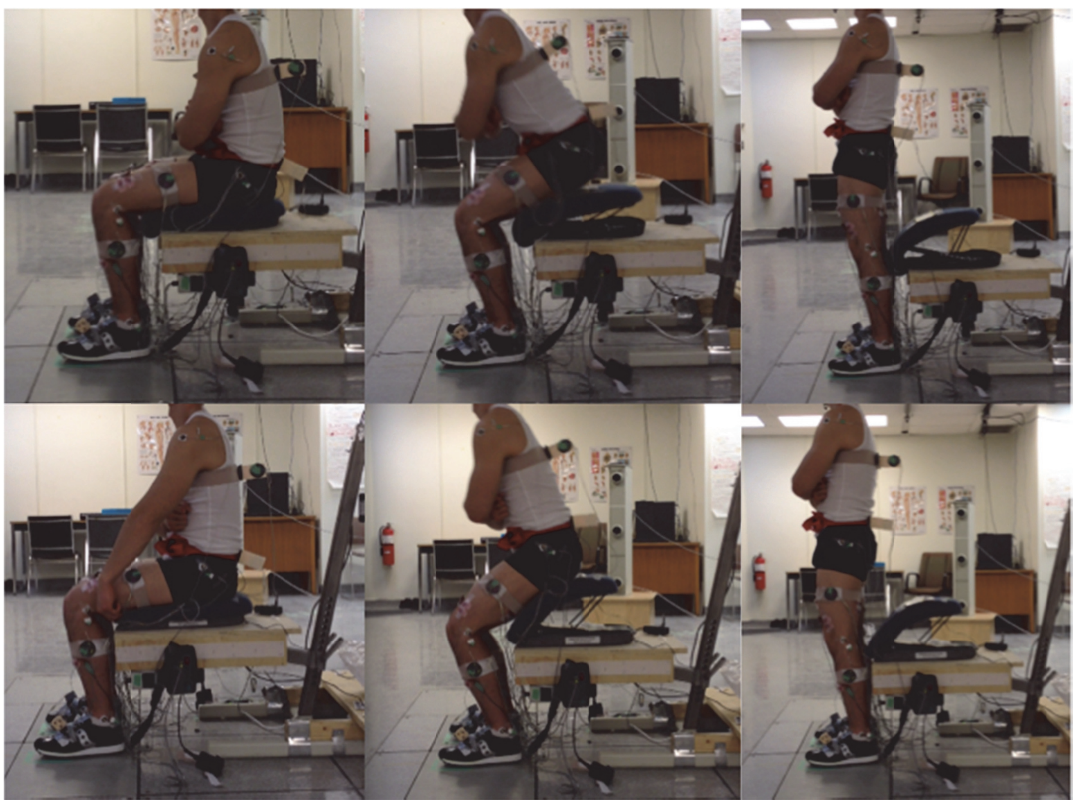


Figure 3.3. Example trials of Seat Assist™ (top) and Power Seat™ (bottom). Start (left), seat-off (center), standing (right)

3.2.4 Maximum Voluntary Isometric Contractions

After the STS trials, a subject bias trial of resting muscle activity was collected with the participant lying in a supine position. Then a series of maximum voluntary isometric contractions against a Cybex isokinetic dynamometer (Cybex International Inc, MA, USA) were performed. The EMG amplitudes from this series was used to amplitude normalize the STS electromyograms and the torque recorded was used a measure of muscle strength (Nm) for the ankle dorsiflexors and plantflexors, and for the knee flexors and extensors. The exercises included i) knee extension at 45° with the participant seated, ii) hip flexion combined with knee extension with the knee at 45° with the participant seated, iii) knee flexion at 55° in a prone position, iv) sitting plantarflexion with the ankle in neutral, v) standing unilateral plantarflexion (no cybex), vi) sitting dorsiflexion with the ankle in neutral, vii) hip extension lying in prone position (no cybex), and viii) trunk extension in a prone position (no cybex). Participants were given one practice trial prior to performing two trials of each exercise and standardized verbal encouragement was given to elicit maximum effort during each trial. Visual feedback for isometric torque was also provided after all practice and test trials. Each exercise was held for 3-seconds with a one-minute rest period provided. A gravitational moment correction trial was recorded prior to each normalization exercise with the subject completely relaxed.

3.3 DATA ACQUISITION

EMG signals were pre-amplified (500X) then amplified using two eight channel EMG measurement systems (Bortec Inc., Calgary, AB, Canada) (impedance = ~10 GΩ, common mode rejection ratio = 115dB at 60 Hz, band-pass 10-1000 Hz). EMG signals were sampled at 2000 Hz (16bit, +/-5V) using an analog-to-digital converter (BNC 2090 National Instruments, Austin, TX, USA) and custom programs in LabView 2009 9.0 (National Instruments, Austin, TX, USA) and stored for later processing.

Force plate data were collected with two AMTI force platforms (Advanced Medical Technology Inc, Watertown, Mass) and sampled at 500 Hz (16bit, +/-2V) using and analog-to-digital (A-to-D) converter (Optotrak Data Acquistion Unit II, Northern Digital Inc., Waterloo, ON, Canada). Each force plate was aligned to the global coordinate system for each limb. Three-dimensional lower limb motion data was collected at 50 Hz using two Optotrak 3020™ camera banks (Northern Digital Inc., Waterloo, ON, Canada). Motion and force plate data were stored for later processing using Northern Digital First Principles software (Northern Digital Inc., Waterloo, ON, Canada). A one second calibration trial with the subject standing in a neutral position was used to calibrate the position of the markers relative to the triad markers in a reference position.

3.4 DATA PROCESSING

3.4.1 Surface Electromyography

All data processing was completed using custom programs written in MATLAB version 7.4 (Mathworks, Natick, MA, USA) based on our standard lab procedures (Hubley-Kozey et al. 2006, Landry et al. 2007) with modifications for bilateral analysis. EMG waveforms were band-pass filtered (20-500Hz), corrected for participant bias, converted to microvolts, full-wave rectified and low-pass filtered, recursive (4^{th} order Butterworth filter) at 6 Hz. EMG data during the STS trials were amplitude-normalized to the maximum EMG amplitude during the MVIC exercises. The maximum activation from all MVIC exercises, regardless of the exercise was obtained using a 100ms moving-average window (99 ms overlap) (Hubley-Kozey et al. 2006).

3.4.2 Isometric Torque

Maximum torque for each exercise trial and gravitational moment trial was identified using a 500ms moving-average window (0ms overlap) (Hubley-Kozey et al. 2006). The muscle moment (Mmusc) was calculated using equations based on static equilibrium (Equation 3.1). Gravitational moments (Mgrav) were either subtracted from or added to isometic torque values recorded by the Cybex (Mcybex), depending on whether the direction of the exercise force vector was with or opposed to the gravitational force vector. The average of two trials was recorded as muscle strength in Newton-meters (Nm).

[3.1] $M_{musc} = M_{cybex} \pm M_{grav}$

3.4.3 Kinematics

A custom MATLAB™ program was written to calculate bilateral kinematics. The one-second standing calibration trial was used to find the invariant pose matrices from the global coordinate system (GCS) to the technical coordinate system and the anatomical coordinate system (ACS) of the pelvis, thigh, shank, and foot segments ($[T]_{TCS/GCS}$, $[T]_{ACS/GCS}$), using the segment center of gravities (COGs) as the origins. Equation 3.2 provides an example of a pose matrix.

$$[3.2] \qquad [T]_{ACS/GCS} = \begin{array}{cc} 1 & 0 \quad 0 \quad 0 \\ \begin{matrix} COGx \\ COGy \\ COGz \end{matrix} & \left[\; [R]_{ACS/GCS} \quad \right] \end{array}$$

Using these pose matrices, a new pose matrix was found from the technical coordinate system (triads) to the anatomical coordinate system for each segments. During the motion trials, the triad markers and lateral point markers were tracked in the global coordinate system and the pose matrices were used to define the motion in the anatomical coordinate system for each segment. The Joint Coordinate System was used to describe the axes of the joints, which used a Cardan sequence of rotations first about the Y-axis (medial-lateral) for flexion/extension, second about the X-axis (anterior-posterior) for abduction/adduction, and the Z-axis (proximal-distal) for internal/external rotation (Grood, Suntay 1983). Joint motion was described as the distal segment moving about the proximal segment. Two-dimensional sagittal plane trunk flexion angle was determined by equation 3.3 that finds the angle between the vertical and a line between the greater trochanter marker and the shoulder marker (Figure 3.4) (Mak et al. 2003, Burnfield et al. 2012, Munro et al. 1998).

$$[3.3] \; Trunk\ Flexion\ Angle = \cos^{-1}\left(\frac{u \cdot v}{|u||v|}\right)$$

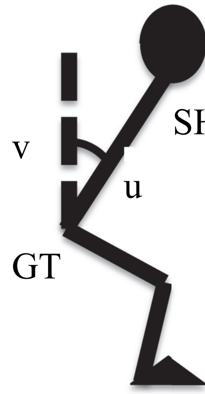


Figure 3.4. Trunk flexion angle between the vertical (v) and the line (u) between the greater trochanter (GT) and the shoulder (SH) markers.

3.4.4 Kinetics

A custom MATLAB™ program was written to calculate bilateral kinetics. Net external knee and hip moments were calculated using an inverse dynamics model, projected on to the joint coordinate system, and normalized to body mass (Vaughan, Davis, O'Connor 1992). Inverse dynamics is a link-segment model that combines ground reaction force, motion data, and limb anthropometrics to calculate net forces and moments acting on the joint, starting at the distal segment and working proximally (Costigan et al. 1992, Vaughan, Davis, O'Connor 1992). The pelvis, thigh, shank, and foot segments are assumed to be rigid bodies that act independently.

[3.4] $\sum F_{segment} = ma + F_{\mathrm{dist}} + mg$

Using equation 3.4, net segment forces were calculated by combining the gravitational and the segmental accelerations, the segment mass, and the reactive force from the distal segment (ground reaction force in the case of the foot segment).

[3.5] $\sum M_{joint} = M_{COG} + ((DJC - COG)\ X\ F_{dist}) + ((PJC - COG)\ X\ F_{prox}) - (-M_{dist})$

Using equation 3.5, net segment moments were calculated by combining, the moment acting on the segment center of gravity (M_{COG}) using segment moment of inertia and angular accelerations (α) and velocities (ω), the distal joint contact moment (cross product of the distal segment net force with the distance of the line of action from the distal joint center (DJC) to segment center of gravity (COG)), the proximal joint contact moment (cross product of the proximal segment net force with the distance of the line of action from the proximal joint center (PJC) to segment center of gravity (COG)), and the net moment of the distal joint (M_{DIST}). Segment center of gravity positions, masses, and moments of inertia were calculated using regression equations based on cadaveric measurements (Vaughan, Davis, O'Connor 1992).

3.4.5 Sit-to-Stand Event Determination

A custom MATLAB™ program was written to calculate STS events. The start of the STS transfer was determined by visually identifying when the horizontal velocity of the trunk increases above base line prior to reaching maximum horizontal velocity. An electric switch was placed between the buttocks and the seat to determine seat-off. The termination of the STS transfer was identified when the vertical velocity of an upper body marker (shoulder marker) returned to zero (Kuo, Tully & Galea 2009, Turcot et al. 2012)

Appendix D addressed a methodological question of determining which standing event selection method was 1) suitable for all conditions and groups, 2) worked for all trials, and 3) accurately selected time of standing. In order to determine the accuracy, key biomechanical measures at the instant of each standing event were measured (knee and hip flexion angles and moments, and the vertical position of the shoulder marker). In addition to when vertical velocity of the shoulder marker equaled zero, four other standing events were calculated and compared. Other standing event methods included; i) when knee extension angular velocity equals zero (Epifanio et al. 2008), ii) when hip extension angular velocity equals zero (Akram, McIlroy 2012, Schenkman, RIley & Mann 1990, Ikeda et al. 1991), iii) when the vertical velocity of thoracic spine marker equals zero (Kuo, Tully & Galea 2009, Turcot et al. 2012), iv) when the vertical ground reaction force rate of change equals zero (Etnyre, Thomas 2007, Jeyasurya 2011). Velocities and rates of changes were calculated using finite differences (Equation 3.6). Of these, the shoulder vertical velocity method was the only method to accurately select time of standing (based on biomechanical outcomes at time of standing) that was also absent of any error trials, making it the preferred standing event method across seating conditions and groups.

[3.6] $Velocity(n) = \frac{[position(n+1) - position(n-1)] * (sampling\ rate)}{2}$

3.5 DEPENDENT VARIABLES

The main dependent variables were calculated using a custom MATLAB™ program (Table 3.1). Peak values were extracted from time-normalized joint angles, moments, and EMG waveforms. Cubic spline interpolation was used to time-normalize waveforms to 100% from seat-off to standing and ensemble average waveforms were calculated from at least three STS trials that were absent of any event timing or IRED data errors. Hip and knee flexion moment impulses (Nm·s/kg) and integrated EMG (%MVIC·s) were calculated for positive areas of non-time-normalized waveforms seat-off to standing (Equations 3.7, 3.8). Legs were collapsed for all dependent variables based on the assumption of STS symmetry in healthy populations (Burnett et al. 2011, Christiansen, Stevens-Lapsley 2010, Lundin et al. 1995).

[3.7] $Integrated\ EMG = \int \%MVIC\ dt$

[3.8] $Moment\ Impulse = \int moment\ dt$

Table 3.2. Dependent variables. Sagittal plane angles and moments, and electromyography

Category	**Variables**
Sagittal Plane Angles (Trunk, Hip, Knee)	1) Peak
Sagittal Plane Net External Moments (Hip, Knee)	1) Peak 2) Impulse
Electromyography *Muscles* –VL, VM, RF, LH, GM	1) Peak 3) Integrated EMG

(VL – vastus lateralis, VM, - vastus medialis, RF – rectus femoris, LH – lateral hamstring, GM – gluteus maximus)

3.6 STATISTICAL ANALYSIS

3.6.1 Objective 1

Student t-tests determined significant differences in age, mass, height, body mass index (BMI) and LEFS scores. All dependent variables were normally distributed with equal variances confirmed by Kolmogorov-Smirnov and Levene's tests, respectively. For muscle strength, a two-factor mixed model Analysis of Variance model (ANOVA) tested for significant group (between) and leg (within) main effects and interactions (alpha=0.05). For peaks, moment impulses and integrated EMG, group (young and old), seat height (ND-normal, ND-raised) differences were tested using a two-factor mixed model ANOVA that accounts for between and within group main effects and interactions (alpha=0.05). Participants were the only random factor in the ANOVA model. Post-hoc testing determined pair-wise significant findings using Bonferonni adjusted alpha levels. Statistical procedures were completed in Minitab™ Ver.16 (Minitab Inc. State College, PA, USA).

3.6.2 Objective 2

Student t-tests determined significant differences in age, mass, height, BMI and LEFS scores. Normality and equal variance of the dependent variables were determined from Kolmogorov-Smirnov and Levene's tests, respectively. A Johnson transformation selected a function to optimally transform data with non-normal distributions or unequal variances. For muscle strength, a two-factor mixed model Analysis of Variance model (ANOVA) tested for significant group (between) and leg (within) main effects and interactions (alpha=0.05). For peaks, moment impulses, and integrated EMG, group (young and old), seat height (ND-raised, PS, SA) differences were tested using a two-factor mixed model ANOVA that accounts for between and within group main effects and interactions (alpha=0.05). Participants were the only random factor in the ANOVA model. Post-hoc testing determined pair-wise significant findings using Bonferonni adjusted alpha levels. Statistical procedures were completed in Minitab™ Ver.16 (Minitab Inc. State College, PA, USA).

CHAPTER 4 THE EFFECT OF SEAT HEIGHT AND AGE ON SIT-TO-STAND TRANSFER BIOMECHANICS AND MUSCLE ACTIVATION

4.1 INTRODUCTION

Sit-to-stand (STS) transfer is an essential activity of daily living and limitations can decrease an individual's overall mobility (Guralnik et al. 1995). STS involves the transfer from a stable three-point sitting position to a more unstable two-point standing position, leading to mechanical and physiological challenges in maintaining stability and generating adequate muscle force (Hughes, Myers & Schenkman 1996, Savelberg et al. 2007, Van der Heijden et al. 2009, Akram, McIlroy 2012, Fujimoto, Chou 2012). Rehabilitation interventions that lower the mechanical demands of rising from a chair have the potential to help older adults with lower limb impairments maintain a more independent and active lifestyle.

Raising the seat height has been shown to be effective at altering motion and reducing the biomechanical demands of a STS transfer (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Schenkman, Riley 1996). Compared to a normal seat height, a raised seat has decreased trunk, hip, and knee angular velocity and displacement (Munro et al. 1998, Schenkman, Riley 1996, Kuo, Tully & Galea 2009, Rodosky, Andriacchi & Andersson 1989, Su, Lai & Hong 1998), and hip and knee flexion moments (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi & Andersson 1989). For example, peak knee flexion moments have been shown to be 60% less and the peak hip flexion moments 50% less from a raised seat height with knee flexion angle of 30° compared to a normal chair height with knee flexion angle of 90° (Arborelius, Wretenberg & Lindberg. 1992). Similar decreases in muscle activation from a raised seat height would be expected but there is limited objective electromyographic (EMG) data. Muscle activation data can help to interpret joint moments of force, as inverse dynamics modeling does not account for additional joint moments caused by antagonist-agonist muscle co-activation. Two studies that used EMG showed that a raised seat height reduced peak muscle activation of the quadriceps and triceps brachii muscles (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000), indicating an effect at the knee and elbow. Only one study has examined a hip extensor muscle, reporting no effect of seat height on muscle activation amplitude of the medial hamstring (Arborelius, Wretenberg & Lindberg 1992). Combining biomechanical data with EMG of muscles surrounding both the hip and the knee would provide comprehensive evidence to evaluate the effectiveness of seating platforms with raised seat heights as a rehabilitative intervention for individuals with STS limitations from aging or pathology.

Older adults, particularly those with quadriceps strength deficits often have difficulty rising from a chair or adopt different kinematic strategies compared to healthy younger adults (Hughes, Myers & Schenkman 1996, Papa, Cappozzo 2000, Gross et al. 1998, Fujimoto, Chou 2012). Hughes et al. (1996) found older individuals with functional impairments used approximately 97% of their available knee extensor strength when rising from a lower than normal chair height compared to 39% in younger individuals. Given decreased strength with age (Hughes, Myers & Schenkman 1996, Fujimoto, Chou 2012, Gross et al. 1998), similar increases in muscle activity, as a percentage of maximum, would be expected. However, no studies were found that compared muscle activation during a STS transfer between younger and older adults. This is important as older adults have been shown to have higher antagonist-agonist co-activation during other functional activities (Nagai et al. 2011, Quirk, Hubley-Kozey 2012). Additionally, only two studies have previously studied the effect of seat height between healthy older and younger adults (Chen et al. 2010, Schenkman, Riley 1996) providing evidence that both groups were affected similarly to different seat heights in terms of subjective ratings (Chen et al. 2010) and angular excursions and velocities (Schenkman, Riley 1996). Whether the effects of a raised seat height on joint and muscle demands are altered by age has not been examined.

The study purpose was to determine the effects of seat height (normal and raised) and age on lower limb biomechanics and EMG dependent variables during a STS transfer by examining trunk flexion angles, hip and knee flexion angles and moments, and muscle activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstrings, and gluteus maximus. Our hypotheses included; 1) Compared to unassisted STS at raised seat height, a normal seat height will have greater peak hip and knee flexion angles and moments, greater hip and knee flexion moment impulses, greater peak and integrated EMG activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstrings, and gluteus maximus. 2) Compared to younger adults, older adults will have similar kinematics and kinetics of the hip and knee and greater peak and integrated EMG activity of all muscles. 3) Both seat heights will have a similar effect on both groups.

4.2 Methods

4.2.1 Participants

Ten healthy older adults (above 65 years) and ten healthy younger adults (20-30 years) recruited from the Dalhousie University community and surrounding area participated in this study. They were considered healthy based on the absence of any cardiovascular, neurological or musculoskeletal issues that would place them at risk of injury or that would alter their ability to perform a STS transfer. Participants were recruited using posters and advertisements on university and public access forums. Written consent was obtained in accordance with Dalhousie Research Ethics Board.

4.2.2 Test Procedure

Each participant attended one testing session at the Dynamics of Human Motion laboratory at Dalhousie University. Participants completed the Lower Extremity Functional Scale (LEFS) (Appendix C.1), which is a self-reported functional status measure questionnaire that has been validated and shown to be valid and reliable (Binkley et al. 1999). Standard anthropometric measures were recorded including; height, mass, hip and waist circumferences, and bilateral knee heights, foot widths, thigh and calf circumferences. Leg dominance was determined by asking each participant “Which leg do you use to kick a ball?” (Burnett et al. 2011).

Participants wore spandex shorts for unobstructed placement of surface EMG electrodes and motion capture infrared emitting diode (IRED) skin surface markers. Surface EMG preparation and collection protocols were based on previous studies (Hubley-Kozey et al. 2006, Rutherford, Hubley-Kozey & Stanish 2011) and were consistent with the International Society of Electrophysiology and Kinesiology and SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines (Stegeman, Hermens 1999). One researcher trained in EMG techniques applied the electrodes. Prior to electrode placement, skin was prepared by light shaving with a disposable razor and abrading with 70% alcohol wipes. Disposable Ag/AgCl surface electrodes (10mm diameter) with 20mm inter-electrode distance (3M™, Red Dot™, Repositionable Monitoring Electrodes, St.Paul, MN, USA), were placed in a bipolar configuration along the muscle fibre orientation of the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH), and gluteus maximus (GM). One ground electrode was placed over the tibia shaft. Validation of the EMG signal and proper gain adjustment were obtained using muscle palpation and a series of muscle specific isometric contractions (Winter, Fuglevand & Archer 1994).

For motion capture, IRED skin surface markers were affixed to the lateral aspect of both lower extremities and trunk based on previous work (Landry et al. 2007, McKean et al. 2007, Rutherford et al. 2008). Single IRED markers were positioned on the skin at the greater trochanter and lateral epicondyle of the femur, the lateral malleolus, and the lateral aspect of the shoulder. Clusters of three markers on a rigid body (triad) were placed roughly midway on the foot, shank, thigh, the sacrum of the pelvis, and the thoracic spine (approximately over the spinous process of thoracic vertebra). Other important anatomical points that were obstructed from cameras during STS trials were collected as virtual points. The subject stood in a neutral position and a digitizing probe located the right and left thoracic spine, right and left anterior superior and posterior superior iliac spines, medial epicondyle of the femur, fibular head, tibial tuberosity, medial malleolus, base of the second metatarsal and center of the posterior calcaneous.

Each participant performed five trials of two different seating conditions using an armless, backless chair with adjustable seat height; 1) normal seat height set to each participant's knee height and 2) raised seat height. Before every normal seat height trial, the participant was positioned on the chair with the knee angle positioned at 90° using a standard goniometer and the feet positioned parallel about shoulder width apart. The participant performed several trials to ensure they were performing each condition at a consistent self-selected speed and movement strategy. Participants were instructed to remain on the chair in an erect posture facing forward until a signal was given to begin the STS trial. In addition, they were instructed to fold their arms across their chest, keep their feet on the floor, and face forward until trial completion. The normal seat height condition was collected first with the participant sitting on the seat pan of Power Seat™ device (Power Seat™ UPE-P100Ex 24VDC, Uplift Technologies Inc., Dartmouth, NS, Canada). The raised seat height trials followed with the participant seating on the Power Seat™ in the extended position. For both conditions, once standing, participants remained standing for the duration of the trial recording.

After the STS trials, a subject bias trial of resting muscle activity was collected with the participant lying in a supine position. A series of maximum voluntary isometric contractions (MVIC) exercises against a Cybex isokinetic dynamometer (Cybex International Inc, MA, USA) were performed to elicit maximum muscle activity and to provide a measure of muscle strength (Nm) including; i) knee extension and ii) hip flexion combined with knee extension at 45° with the participant seated, iii) knee flexion at 55° in a prone position, iv) sitting plantarflexion with the ankle in neutral, v) standing unilateral plantarflexion (no cybex), vi) sitting dorsiflexion with the ankle in neutral, vii) hip extension and viii) trunk extension lying in prone position (no cybex). Participants were given one practice trial prior to performing two trials of each exercise and standardized verbal encouragement was given to elicit maximum effort during each trial. Visual feedback of isometric torque production was also provided after all practice and test trials. Each exercise was held for 3-seconds with a one-minute rest period provided. A gravitational moment correction trial was recorded prior to each normalization exercise with the subject completely relaxed.

4.2.3 Data Acquisition and Signal Processing

EMG signals were pre-amplified (500X) then amplified using two eight channel EMG measurement systems (Bortec Inc., Calgary, AB, Canada) (impedance = ~10 GΩ, common mode rejection ratio = 115dB at 60 Hz, band-pass 10-1000 Hz). EMG signals were sampled at 2000 Hz (16bit, +/-5V) using an analog-to-digital (A-to-D) converter (BNC 2090 National Instruments, Austin, TX, USA) and stored data for later processing using custom programs in LabView 2009 9.0 (National Instruments, Austin, TX, USA). Force plate data were collected with two AMTI force platforms (Advanced Medical Technology Inc, Watertown, Mass) and sampled at 500 Hz (16bit, +/-2V) using an A-to-D converter (Optotrak Data Acquistion Unit II, Northern Digital Inc., Waterloo, ON, Canada). Each force plate was aligned to the global coordinate system for each limb.

Three-dimensional lower limb motion data were collected at 50 Hz using two Optotrak 3020™ camera banks (Northern Digital Inc., Waterloo, ON, Canada). Data were stored for later processing using Northern Digital First Principles software (Northern Digital Inc., Waterloo, ON, Canada). A one-second calibration trial with the subject standing in a neutral position was used to calibrate the position of the markers relative to the triad markers in a reference position.

All data processing was completed using custom programs written in MATLAB version 7.4 (Mathworks, Natick, MA, USA) based on our standard lab procedures (Hubley-Kozey et al. 2006, Landry et al. 2007) with modifications for bilateral analysis. EMG waveforms were band-pass filtered (20-500Hz), corrected for participant bias, converted to microvolts, full-wave rectified and low-pass filtered (4th order Butterworth filter) at 6 Hz. EMG data during the STS trials were amplitude-normalized to the maximum EMG amplitude during the MVIC exercises. The maximum activations from all MVIC exercises, regardless of the exercise were obtained using a 100ms moving-average window (99 ms overlap) (Hubley-Kozey et al. 2006). Maximum torque for each exercise was identified using a 500ms moving-average window (0ms overlap) and corrected for gravitational moments (Hubley-Kozey et al. 2006). The average of two trials was recorded as muscle strength in Newton-meters (Nm).

The Joint Coordinate System described the axes of the joints, using a Cardan sequence of rotations (Grood, Suntay 1983). Joint motion was described as the distal segment moving about the proximal segment. Two-dimensional sagittal plane trunk flexion angle was described by the angle between the vertical and a line between the greater trochanter marker and the shoulder marker. Net external knee and hip moments were calculated using an inverse dynamics model and normalized to body mass (Nm/kg) (Vaughan, Davis, O'Connor 1992).

The start of the STS transfer was determined by visually identifying when the horizontal velocity of the trunk increased above base line prior to reaching maximum horizontal velocity. An electric switch placed between the buttocks and the seat determined seat off. The termination of the STS transfer was identified by the sample where vertical velocity of the shoulder marker reached zero (Kuo, Tully & Galea 2009, Turcot et al. 2012).

The main dependent variables are in Table 4.1. Peak values were extracted from time-normalized joint angles, moments, and EMG ensemble average waveforms. Cubic spline interpolation was used to time-normalize waveforms to 100% from seat-off to standing and ensemble averages were calculated from at least three STS trials that were absent of any event timing or IRED data errors. Hip and knee flexion moment impulses (Nm·s/kg) and integrated EMG (%MVIC·s) were calculated for positive areas of non-time-normalized waveforms from seat-off to standing to account for both amplitude and duration.

Table 4.1. Dependent variables. Sagittal plane angles and moments, and electromyography.

Category	Variables
Sagittal Plane Angles (Trunk, Hip, Knee)	1) Peak
Sagittal Plane Net External Moments (Hip, Knee)	1) Peak 2) Impulse
Electromyography *Muscles* –VL, VM, RF, LH, GM	1) Peak 2) Integrated EMG

(VL – vastus lateralis, VM, - vastus medialis, RF – rectus femoris, LH – lateral hamstring, GM – gluteus maximus)

4.2.4 Statistical Analysis

Student t-tests determined significant differences in age, mass, height, BMI and LEFS scores. All dependent variables were normal with equal variances determined from Kolmogorov-Smirnov and Levene's tests, respectively. For muscle strength, a two-factor mixed model Analysis of Variance model (ANOVA) tested for significant group (between) and leg (within) main effects and interactions (alpha=0.05).

For peaks, moment impulses, and integrated EMG, the legs were collapsed. Group (young and old), seat height (normal, raised) differences were tested using a two-factor mixed model ANOVA that accounts for between and within group main effects and interactions (alpha=0.05). Post-hoc testing determined pair-wise significant findings using Bonferonni adjusted alpha levels. Statistical procedures were completed in Minitab™ Ver.16 (Minitab Inc. State College, PA, USA).

4.3 RESULTS

4.3.1 Participant Demographics

Twenty healthy participants were recruited, including ten young adults (20-30 years old) and ten older adults (>65 years old). Group demographics and anthropometrics are in Table 4.2. Older adults were significantly older and reported significantly lower scores on the lower extremity functional scale.

Table 4.2. Mean (SD) of participant demographics

	Young Adults	Older Adults
N	10	10
Percent Female	50%	50%
Age (years)	**25 (2)**	**69 (3)***
Mass (kg)	77 (13)	75 (15)
Height (m)	1.77 (0.08)	1.70 (0.10)
BMI (kg/m^2)	24.3 (2.6)	25.8 (4.0)
WHR	0.79 (0.07)	0.89 (0.09)
LEFS (/80)	**79.9 (0.3)**	**73.1 (4.8)***

BMI = Body Mass Index
WHR = Waist to Hip Ratio
LEFS = Lower Extremity Functional Scale
* (Bold) indicate significant differences (p<0.05)

Isometric torque generated for different muscle groups are in table 4.3. No significant (p>0.05) group by leg interactions were found. Younger adults generated 25%, 20%, and 25% greater isometric torque than the older adults for the knee flexors, knee extensors, and ankle plantarflexors, respectively. Only knee flexion was significantly different (p<0.05) between groups.

Table 4.3. Mean (SD) of isometric torque output (Nm).

Muscle group	Young Adults		Older Adults		Group	Leg	Interaction
	Dom	Non	Dom	Non			
Knee Extensors	151.1 (33.6)	163.9 (39.0)	129.9 (41.9)	124.4 (35.7)	0.078	0.933	0.195
Knee Flexors	83.6 (20.4)	85.8 (21.3)	64.2 (20.9)	62.5 (19.8)	**0.022**	0.832	0.930
Ankle Plantarflexors	102.7 (27.7)	108.7 (30.0)	77.2 (28.5)	80.3 (28.5)	0.067	0.211	0.562
Ankle Dorsiflexors	40.8 (7.7)	41.8 (6.9)	36.8 (13.9)	37.9 (13.3)	0.446	0.366	0.902

Dom (dominant), Non (non-dominant)

4.3.2 Temporal Characteristics

Table 4.4 provides descriptive statistics of the temporal characteristics associated with STS for each seat height for both age groups. There were no significant group by condition interactions (p>0.05) for any temporal variables. Time from seat-off to standing was significantly greater for the normal seat height by 0.2 seconds (p<0.05).

Table 4.4. Mean (SD) of temporal characteristics

	Young Adults		Older Adults		Group	Height	Inter
	Normal	Raised	Normal	Raised			
Total Time (s)	1.8 (0.3)	1.7(0.3)	2.1 (0.5)	2.1 (0.7)	0.1	0.181	0.377
Seat-off to stand (s)	1.1 (0.1)	0.9 (0.1)	1.2 (0.3)	1.1 (0.4)	0.137	**<0.001**	0.535

4.3.3 Angles

Peak sagittal plane trunk, hip, and knee angles are in Table 4.5. There were no significant group by condition interactions (p>0.05). The greatest difference in angle between groups was approximately 9° for peak hip flexion angle, but this was not significant (p=0.071).

Table 4.5. Mean (SD) peak flexion angles (degrees).

		Young Adults		Older Adults		Group	Height	Inter
Peak		Normal	Raised	Normal	Raised			
Trunk	Peak	45 (3)	26 (7)	46 (5)	28 (8)	0.58	**<0.001**	0.633
Hip	Peak	81 (12)	53 (9)	89 (8)	62 (11)	0.071	**<0.001**	0.989
Knee	Peak	78 (7)	59 (7)	78 (10)	56 (10)	0.696	**<0.001**	0.217

Flexion (+)

Peak trunk, hip, and knee flexion angles were significantly lower for the raised seat height compared to the normal seat height (p<0.05) by approximately 20° for the trunk, 25° for the hip, and 20° for the knee. Kinematic waveforms are in Appendix E.

4.3.4 Moments of Force

Table 4.6 provides the peak and impulse values of net external moments for each chair height in younger and older adults. There were no significant group by condition interactions (p>0.05) for any flexion moment variables.

Table 4.6. Mean (SD) peak net external flexion moments (Nm/kg) and impulse (Nms/kg).

		Young Adults		Older Adults		Group	Height	Inter
		Normal	Raised	Normal	Raised			
Hip	Peak	0.90(0.19)	0.62(0.22)	0.90(0.17)	0.61(0.16)	0.983	**<0.001**	0.891
	Imp	0.53(0.22)	0.28(0.19)	0.60(0.21)	0.36(0.16)	0.417	**<0.001**	0.865
Knee	Peak	0.62(0.11)	0.60(0.19)	0.56(0.13)	0.49(0.19)	0.22	0.097	0.355
	Imp	0.38(0.13)	0.26(0.12)	0.33(0.09)	0.22(0.10)	0.88	**<0.001**	0.593

Flexion (+)

Peak and impulse net external hip flexion moment values were significantly lower for the raised seat height compared to the normal seat height (p<0.05) (Figure 4.1). Peak hip moment values decreased by approximately 30%, whereas, impulse values decreased by approximately 45%. Knee flexion moment impulse was significantly lower (approximately 30%) for the raised seat height compared to the normal seat height (p<0.05). Kinetic waveforms are in Appendix E.

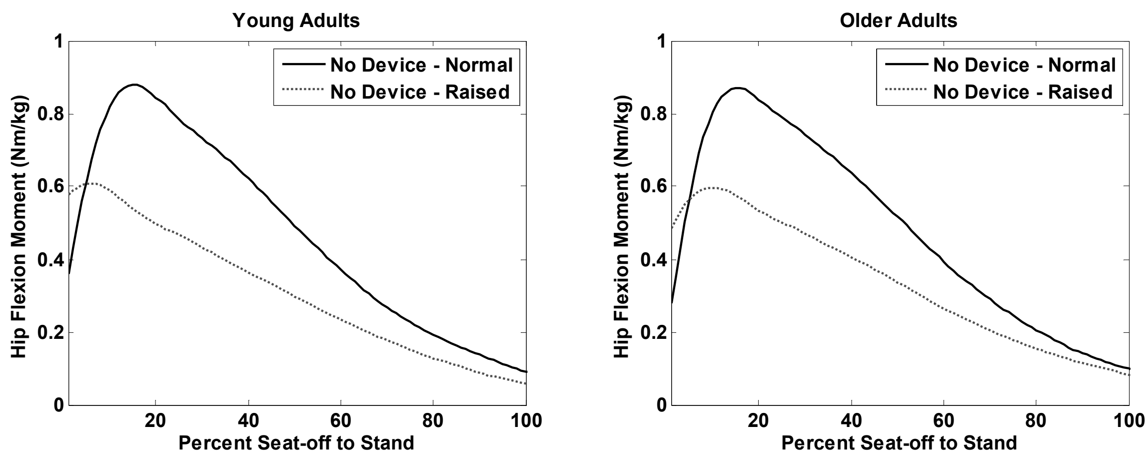


Figure 4.1. Ensemble average hip flexion moment waveforms for two different seat heights between the younger adults (left) and older adults (right).

4.3.5 Muscle Activation

Table 4.7 provides the peak and integrated EMG activity for each chair height in younger and older adults. There was a significant group by condition interaction ($p<0.05$) for VM integrated EMG, which was greatest for the normal seat height in older adults, followed by the raised seat height in older adults and the normal seat height in younger adults, and lastly, the raised seat height in younger adults (Figure 4.2). The older adult group had a greater reduction of integrated EMG of the VM (23 %MVIC·s) with the raised seat height compared to the younger adults (12 %MVIC·s). A similar trend was found for integrated EMG of the VL ($p=0.053$) and peak activity of the RF ($p=0.089$) and GM ($p=0.061$) but these did not reach significance.

Older adults had significantly greater peak EMG activity of the VM, RF, LH and GM as a percentage of maximum (% MVIC) ($p<0.05$) and significantly greater integrated EMG activity of VL, RF, LH and GM (% MVIC·s) ($p<0.05$).

Table 4.7. Mean (SD) of peak EMG activity (%MVIC) and integrated EMG (%MVICs).

		Young Adults		Older Adults		Group	Height	Interaction
		Normal	Raised	Normal	Raised			
VL	Peak	58 (26)	31 (10)	69 (20)	45 (22)	0.159	**<0.001**	0.717
	Int	28 (6)	16 (5)	47 (11)	27 (7)	**<0.001**	**<0.001**	0.053
VM	Peak	52 (14)	29 (10)	79 (25)	50 (24)	**0.009**	**<0.001**	0.278
	Int	26 (5)	14 (4)	51 (19)	28 (9)	**<0.001**	**<0.001**	**0.005**
RF	Peak	23 (9)	11 (3)	39 (15)	22 (14)	**0.011**	**<0.001**	0.089
	Int	11 (4)	7 (3)	24 (11)	16 (8)	**0.002**	**<0.001**	0.192
LH	Peak	12 (5)	10 (8)	24 (7)	19 (8)	**0.002**	**0.022**	0.182
	Int	8 (3)	7 (4)	20 (9)	16 (9)	**0.002**	**0.017**	0.166
GM	Peak	19 (7)	16 (6)	32 (13)	23 (8)	**0.016**	**<0.001**	0.061
	Int	14 (6)	10 (4)	25 (11)	19 (9)	**0.007**	**<0.001**	0.378

Device main effects ($p<0.05$) were found for peak EMG activity of the all muscles and for integrated EMG of VL, RF, LH and GM with significantly lower values for the raised seat height. EMG waveforms are in Appendix E.

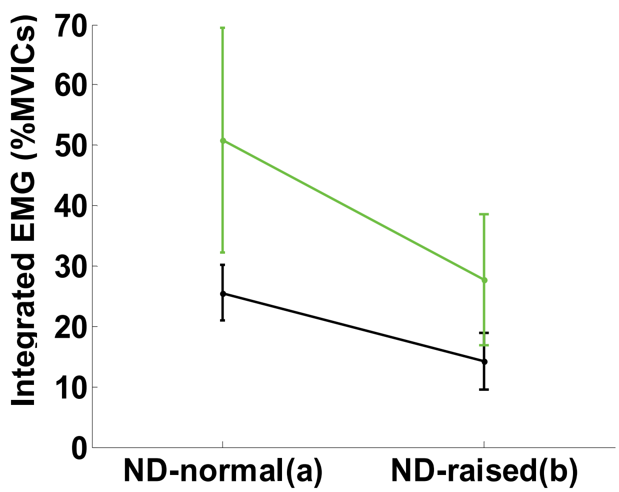


Figure 4.2. Mean (SD) of integrated EMG of vastus medialis (%MVICs). ND-normal (left) and ND-raised (right).

4.3.6 Summary of Results

A summary of key significant findings for seat height and group main effects are in Table 4.8. These illustrate the between group differences for the EMG dependent variables and between seat height differences for biomechanical and EMG dependent variables.

Table 4.8. Key significant findings ($p<0.05$).

	Seat Height	Group
Peak Angles	**Trunk, Hip, Knee** Normal>Raised	No group differences
Peak Moments	**Hip** Normal>Raised **Knee** No device differences	No group differences
Moment impulse	**Hip and Knee** Normal>Raised	No group differences
Peak EMG	**All Muscles** Normal>Raised	**VM, RF, GM, LH** Older>Younger
Integrated EMG	**VL, RF, GM, LH** Normal>Raised **VM** (Interaction) Normal(old)>Normal(young), Raised(old)>Raised(young)	**VL, RF, GM, LH** Older>Younger

4.4 DISCUSSION

The goal of this study was to provide a comprehensive assessment of the effect of seat height on a STS transfer and determine whether there were differences between age groups. Overall, the results of this study support previous findings that a higher than normal seat height reduces the biomechanical and neuromuscular demands (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Schenkman, Riley 1996).

Both groups were similar in their height, mass, BMI, and waist-to-hip ratio. The older group had significantly lower self-reported Lower Extremity Function Score (73.1/80) compared to younger adults (79.9/80) but within the minimal clinically important difference of 9 points (Binkley et al. 1999). Both groups were considered high physically functioning groups, reporting little to no difficulty for many simple and physically demanding daily tasks. The younger group had significantly greater isometric torque output of knee flexors (25%), which is similar to previous studies, reporting decreased strength with age (Hughes, Myers & Schenkman 1996, Fujimoto, Chou 2012, Gross et al. 1998). There younger group also had 20% greater isometric torque output of the knee extensors and 25% greater isometric torque output for the ankle plantarflexors, though both were not significant. This lack of difference in torque output could in part be due to a lack of statistical power associated with the small sample, which is a limitation of our study. Both groups did not have any strength differences between legs, which was expected, as they were both healthy populations. In addition, there were no group differences in temporal characteristics of the STS transfer, which is similar to previous findings on STS at self-selected speed (Akram, McIlroy 2012, Papa, Cappozzo 2000).

Both groups had similar kinematics and kinetics, which is consistent with our hypothesis and previous findings (Hughes, Myers & Schenkman 1996, Ikeda et al. 1991). Several studies have shown increased trunk flexion in older adults with strength deficits, which is characteristic of a 'stabilization strategy' to reduce the knee flexion moments and demands on the quadriceps muscles but with a trade-off of increased hip flexion moments (Fujimoto, Chou 2012, Schenkman, RIley & Mann 1990, Papa, Cappozzo 2000, Gross et al. 1998, Yoshioka et al. 2007). In our study, there were no differences in peak trunk flexion angles between groups, suggesting the older adult individuals did not adopt this strategy. Savelberg et al. (2007), provided evidence that individuals do not change their kinematic and kinetic STS strategy until a threshold level when their knee extensor muscles become overloaded. The older adults who participated in this study had high self–reported measures of lower extremity function and were 20% weaker in their knee extensors, which may not be enough to see a change in their STS strategy.

In general, in order to generate similar peak and impulse moment values, older adults used greater peak and integrated EMG activity. Similar to our findings, Hughes et al. (1996) also reported no differences in knee moments between groups but the older group used greater knee moments as a percentage of their maximum isometric knee extensor strength. Two contributing factors help to explain greater muscle activation in the older adult group. Firstly, older adults had greater strengths deficits; 20-25% less isometric torque during knee flexion, knee extension, and ankle plantarflexion exercises compared to the younger group. Hip extension strength was not measured, which is a limitation of the study. Overall, the older adults had to work at a greater percentage of their maximum muscle activation to generate similar joint moments. Secondly, consistent with findings during other functional tasks (Nagai et al. 2011, Quirk, Hubley-Kozey 2012), older adults had greater antagonist-agonist muscle co-activation. Despite producing 25% less knee flexion torque, the older adults had 100% greater peak lateral hamstrings activity during the STS transfers compared to the younger adults. Similarly, older adults produced 20% less knee extension torque but had 25-75% greater peak quadriceps activity during the STS transfers. Greater co-activation causes greater muscle effort and potentially greater joint loading. This is the first study to report the values of peak and integrated EMG activity for both healthy younger and older adults. Older adults had peak EMG values for the vastus medialis of 79 %MVIC and for the vastus lateralis of 69 %MVIC from a normal chair height, highlighting the high demands on the quadriceps muscles during a STS transfer even for this healthy older group.

Comparing between seat heights, there were no differences in the total STS time. The raised seat height had significantly shorter seat-off to standing duration, which is likely related to shorter distance travelled from the raised seat height to a standing position. The raised seat height significantly reduced peak trunk, hip, and knee angles, which is consistent with previous findings of decreased joint angular excursions with higher seat height (Munro et al. 1998, Su, Lai & Hong 1998, Kuo, Tully & Galea 2009, Rodosky, Andriacchi & Andersson 1989, Schenkman, Riley 1996). Peak hip flexion moments were also significantly reduced, confirming previous findings of lower demands at the hip joint (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi &

Andersson 1989). Arborelius et al. (1992) found a raised seat height to a knee angle of 30° decreased hip moments by roughly 50% compared to a knee angle of 90°. We found the hip moments to decrease approximately 30% for when the seat height is raised approximately 15cm with a knee angle of approximately 60° at seat-off. Contrary to other studies (Arborelius, Wretenberg & Lindberg. 1992, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Hughes, Myers & Schenkman 1996, Rodosky, Andriacchi & Andersson 1989) and our hypothesis, there was no effect of a raised seat height on peak knee flexion moments but there was a reduction in knee moment impulse. Lack of statistical significance (p=0.097) of the peak knee flexion moment could be due to a small sample size and limited statistical power, particularly in the older adult group who had 10-15% reduction of the mean peak knee flexion moment with a raised seat height. The reduction of the knee flexion moment impulse with the raised seat height may reflect the shorter time from seat-off to standing and thus, a reduction of overall knee moments during STS from a raised seat height.

In general, peaks and integrated EMG activity of all muscles were lower with the raised seat height by approximately 35-47% for the quadriceps, 20% for the lateral hamstring, and 25% for the gluteus maximus. Therefore, lower maximum and overall muscle effort are needed from a raised seat height for muscles that are used to extend the knee and the hip during a STS transfer. Quadriceps integrated EMG was reduced with the raised seat height affirming the knee flexion moment impulse findings. Despite no significant differences in peak knee flexion moments, we found a significant reduction of peak EMG activity of the vastus lateralis, vastus medialis, and rectus femoris with the raised seat height, which could be a result of reduced antagonist lateral hamstrings activity. Two other studies have used EMG to study the effect of raised seat height on lower limb muscles (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000). In this study, the peak muscle activity of the vastus medialis and vastus lateralis was also reduced with a raised seat height (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000). Contrary to our findings, Arborelius et al. (1992) found no differences in peak EMG activity of the rectus femoris or hamstrings. Our study was the first study to examine and show a reduction of gluteus maximus activity with a raised seat height, which affirms findings of lower peak hip flexion moments. This was also the first study to analyze integrated EMG during a STS transfer. This measure was used to give an indication of the over all muscle effort from seat-off to standing in addition to the peak muscle activity.

Only two studies have examined the effect of seat height between healthy older and younger adults (Chen et al. 2010, Schenkman, Riley 1996) and reported that both groups were affected similarly to different seat heights in terms of subjective ratings (Chen et al. 2010) and angular excursions and velocities (Schenkman, Riley 1996). This is the first study to examine the effect of seat height between healthy older and younger adults on joint moments or neuromuscular outcomes. Overall, the effects of a raised seat height on joint and muscle demands are not altered by age. Only one significant interaction effect was found for integrated EMG of the VM; a greater reduction of overall EMG activity with the raised seat height for older adults compared to younger adults. Despite a lack of statistical significance ($p>0.05$), integrated EMG of the VL and peak activity of the RF and the GM showed a similar trend. This suggests that older adults may benefit more from raised seat height compared to younger adults in terms of muscular demands.

The findings of this study need to be interpreted within the study limitations and cannot be extrapolated beyond. Set-up constraints used to isolate the effect of seat height and avoid confounding effects of other determinants of STS transfers (Janssen, Bussmann & Stam 2002) limit generalizing findings to non-constrained STS tasks. Foot position was controlled and participants were instructed to fold their arms across their chest to avoid the use of arms, which may limit their natural STS movements at different seat heights. This study is part of a larger study on examining the effects of lifting-seat devices and our raised seat height condition was performed on a lifting-seat device in its fully extended position. Differences may exist compared to raised seats used in other studies.

4.5 Summary

Older adults had similar kinetics and kinematics compared to the younger adult group to perform at STS task at two different seat heights. In order to generate similar moments of force, the older adult group had greater EMG activity as a percentage of maximum. Raised seat height reduced peak hip and knee flexion angles, peak hip flexion moments, hip and knee flexion moment impulses, and the peak and integrated EMG activity of lower limb muscles, affecting both groups similarly, except for VM activity, which had a greater reduction in overall activity in older adults with a raised seat height compared to younger adults. Overall, this study provides comprehensive findings supporting a raised seat height as a rehabiliation intervention to facilitate STS transfers and as a design goal for lifting-seat devices placed on normal seating platforms.

CHAPTER 5 SIT-TO-STAND TRANSFER BIOMECHANICS AND MUSCLE ACTIVATION: COMPARISON OF LIFTING-SEAT DEVICES TO A RAISED SEAT HEIGHT IN YOUNGER AND OLDER ADULTS

5.1 INTRODUCTION

A sit-to-stand (STS) transfer places greater demands on lower limb joints and muscles than other functional tasks, such as walking (Mizner, Snyder-Mackler 2005, Hodge et al. 1989). Older adults with muscle, bone, and joint impairments, particularly those with quadriceps muscle weakness often have difficulty rising from a chair or adopt different kinematic strategies (Hughes, Myers & Schenkman 1996, Papa, Cappozzo 2000, Gross et al. 1998, Fujimoto, Chou 2012). Seating platforms with raised seat heights have been shown to be an effective method for reducing the biomechanical demands of a STS transfer (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Schenkman, Riley 1996). In chapter four, a raised seat height reduced biomechanical demands on the hip and the knee joints and EMG activity of surrounding muscles.

Lifting-seat devices have been designed to raise the effective seat height with the goal of reducing lower limb muscular requirements during a STS for frail elderly populations with pathologies that compromise the musculoskeletal system including; arthritis, surgical rehabilitation, strokes, and neurological problems such as multiple sclerosis, stroke, and Parkinson's disease. Lifting-seat devices have been reported to reduce pain and perceived effort (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000) but there is limited biomechanical or neuromuscular objective evidence supporting the use of these devices. Of the evidence that exists, two studies have shown lifting-seat devices to be effective at reducing the hip moments, knee moments, and quadriceps activity (Jeyasurya 2011, Wretenberg et al. 1993), whereas the other two studies have shown no change in knee moments and quadriceps activity (Munro, Steele 2000, Munro et al. 1998).

Conflicting evidence could be a related to several issues. Firstly, non-standardized methodologies have been employed, such as varying foot position and use of armrests that make comparisons between unassisted and assisted transfers and between studies difficult. Secondly, subject populations also vary across studies making interpretation difficult. Studying the effect of these devices on healthy populations with standardized methodology will develop a comprehensive base-point for future investigation on pathological populations. Lastly, lifting-seat devices can be categorized into two main groups; a spring-loaded or pneumatic design (Bashford et al. 1998, Munro et al. 1998, Munro, Steele 2000, Wretenberg et al. 1993) and an elevator lifting design (Jeyasurya 2011). The former uses pneumatics to create a mechanical preload as the device lowers to provide an assistive force while the user actively performs a STS transfer. The latter uses an electric-elevator design that transfers the user to a raised position before performing a STS transfer. No studies have compared lifting-seat devices with different lifting mechanisms, and thus, how they differ and who will benefit from which type of device remains unclear.

As a result of limited comprehensive evidence, there remain several key questions with respect to how both types of lifting-seat devices affect a STS transfer. Firstly, are these devices effective at lowering the demand on the hip and knee joints and musculature surrounding both joints? Secondly, is the effect similar in younger and older adults? The latter is important as lifting-seat devices are targeted to older adults and those with pathologies. Thirdly, do both types of devices with different lifting mechanisms affect the user the same? And lastly, whether the effect of the lifting-seat devices is simply a result of an increase in seat height or some other effect on STS strategy? A better understanding of how these devices function, particularly, how they compare to their design goal (an unassisted STS from a raised seat height) will provide objective evidence that can be used to guide future device modifications and inform guidelines for prescription.

The study purposes were to determine the effects of two different lifting-seat devices compared to an unassisted STS from a raised chair height and the effects of age on lower limb biomechanics and neuromuscular characteristics during a STS transfer by examining hip, knee flexion angles and moments, and muscle activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstrings, and gluteus maximus. The hypotheses were; 1) there will be no differences between either type of lifting-seat device or the unassisted STS from a raised seat height on biomechanical and neuromuscular outcomes, 2) compared to younger adults, older adults will have similar kinematics and kinetics of the hip and knee and greater peak and integrated EMG activity for all muscles. 3) All seating conditions will affect both groups equally.

5.2 METHODS

5.2.1 Participants

Ten healthy older adults (above 65 years) and ten healthy younger adults (20-30 years) absent of any cardiovascular, neurological or musculoskeletal issues that would alter their ability to safely perform a STS transfer participated in one testing session at the Dynamics of Human Motion laboratory. Participants were recruited from the Dalhousie University community and surrounding area using posters and advertisements on university and public access forums. All subjects provided written informed consent in accordance with Dalhousie University Research Ethics Board.

5.2.2 Test Procedure

Upon arrival, participants completed the Lower Extremity Functional Scale (LEFS) (Appendix C.1), a self-reported functional status measure questionnaire that has been validated and shown to be reliable (Binkley et al. 1999). Height, mass, hip and waist circumferences, and bilateral knee heights, foot widths, thigh and calf circumferences were recorded. Each participant was asked, "Which leg do you use to kick a ball?" to determine leg dominance (Burnett et al. 2011).

Surface EMG electrode placement and properties were consistent with the International Society of Electrophysiology and Kinesiology and SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines (Stegeman, Hermens 1999). One researcher trained in EMG techniques applied the electrodes and collection protocols were based on previous studies (Hubley-Kozey et al. 2006, Rutherford, Hubley-Kozey & Stanish 2011). Skin was prepared by light shaving with a disposable razor and abrading with 70% alcohol wipes prior to placement of disposable Ag/AgCl surface electrodes (10mm diameter) with 20mm inter-electrode distance (3M™, Red Dot™, Repositionable Monitoring Electrodes, St.Paul, MN, USA). Muscle palpation during a series of muscle specific isometric contractions and assessment of the EMG recordings were performed to validate EMG signal and for proper gain adjustment

(Winter, Fuglevand & Archer 1994). Electrodes were placed in a bipolar configuration along the orientation of the muscle of the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), lateral hamstring (LH), and gluteus maximus (GM). One ground electrode was placed over the tibia shaft. EMG signals were pre-amplified (500X) then amplified using two eight channel EMG measurement systems (Bortec Inc., Calgary, AB, Canada) (impedance = ~10 GΩ, common mode rejection ratio = 115dB at 60 Hz, band-pass 10-1000 Hz). EMG signals were sampled at 2000 Hz (16bit, +/-5V) using an analog-to-digital converter (BNC 2090 National Instruments, Austin, TX, USA) and custom programs in LabView 2009 9.0 (National Instruments, Austin, TX, USA) and stored for later processing.

Motion data were collected based on previous work (Landry et al. 2007, McKean et al. 2007, Rutherford et al. 2008) using two Optotrak camera banks (Northern Digital Inc., Waterloo, ON, Canada) sampled at 50 Hz. Force plate data were collected with two AMTI force plates (Watertown, MA) sampled at 500 Hz (16bit, +/-2V) using an A-to-D converter (Optotrak Data Acquistion Unit II, Northern Digital Inc., Waterloo, ON, Canada). Data was stored for later processing using Northern Digital First Principles software (Northern Digital Inc., Waterloo, ON, Canada). Three active infrared emitting diode (IRED) markers were placed on a rigid body (triad) were placed on the foot, shank, thigh, sacrum of the pelvis, and thoracic spine (approximately over the spinous process of thoracic vertebra) segments, and individual markers were placed on the skin at the greater trochanter and lateral epicondyle of the femur, the lateral malleolus, and the lateral aspect of the shoulder. Eight virtual markers were identified during neutral standing using a digitizing probe.

EMG, motion, and force plate data were collected from five trials of three different seating conditions using an armless, backless chair with adjustable seat height; 1) Seat-Assist™ (SA) (Seat Assist™ UPE-1, Uplift Technologies Inc., Dartmouth, NS, Canada), 2) Power Seat™ (PS) (Power Seat™ UPE-P100Ex 24VDC, Uplift Technologies Inc., Dartmouth, NS, Canada), 3) no device from a raised seat height (ND-raised). Before every PS and SA trial, the participant was positioned fully on the chair with the knee angle positioned at 90° using a standard goniometer and the feet positioned parallel approximately shoulder width apart. The participant performed several trials to ensure they were performing each condition at a consistent self-selected speed and movement strategy. Participants were instructed to remain on the seat pan in an erect posture facing forward until a signal was given to begin the STS trial. In addition, they were instructed to fold their arms across their chest, keep feet on the floor, and face forward until trial completion. The two lifting-seat conditions were performed first in a random order. The ND-raised trials followed with the participant sitting on the Power Seat™ in the extended position. For the PS trials, participants pulled the lever to engage the seat when the signal was given. Participants were instructed to keep their feet planted on the ground and remain on the seat until maximum elevation and stand up at self-selected pace after the Power Seat™ stops. For the SA trials, the participants were instructed to perform a comfortable STS movement with self-selected speed and movement strategy. For all conditions, once standing, participants remained standing, facing forward for the duration of the trial recording.

Following the STS trials, a subject bias trial of resting muscle activity was collected with the participant completely relaxed and lying in a supine position. Participants then performed a series of maximum voluntary isometric contractions (MVIC) exercises against a Cybex isokinetic dynamometer (Cybex International Inc, MA, USA). The EMG amplitudes from this series was used to amplitude normalize the STS electromyograms and the torque recorded was used a measure of muscle strength (Nm) for the ankle dorsiflexors and plantflexors, and for the knee flexors and extensors. The exercises included; i) knee extension and ii) hip flexion combined with knee extension at 45° with the participant seated, iii) knee flexion at 55° in a prone position, iv) sitting plantarflexion with the ankle in neutral, v) standing unilateral plantarflexion (no cybex), vi) sitting dorsiflexion with the ankle in neutral, vii) hip extension and viii) trunk extension lying in prone position (no cybex). Participants were instructed to give a maximal effort and hold each exercise for three seconds. One practice trial was performed prior to two trials of each exercise with one-minute rest periods provided. Standardized verbal encouragement and visual feedback on torque production was also provided after each practice and test trial. A gravitational moment correction trial was recorded prior to each normalization exercise with the subject completely relaxed.

5.2.3 Data Processing

All data processing was completed using custom programs written in MATLAB version 7.4 (Mathworks, Natick, MA, USA) based on standard lab procedures (Hubley-Kozey et al. 2006, Landry et al. 2007) with modifications for bilateral analysis. Each force plate was aligned to the global coordinate system for each limb. A one second calibration trial with the subject in quiet standing was used to define anatomical coordinate systems in each lower limb segment. The Joint Coordinate System described the axes of the joints using a Cardan sequence of rotations (Grood, Suntay 1983). Joint motion was described as the distal segment moving about the proximal segment. Two-dimensional sagittal plane trunk flexion angle was described by the angle between the vertical and a line between the greater trochanter marker and the shoulder marker. Net external knee and hip moments normalized to body mass (Nm/kg) were calculated using an inverse dynamics model which combines ground reaction force and motion data, limb anthropometrics and inertial properties (Vaughan, Davis, O'Connor 1992).

EMG data were band-pass filtered (20-500Hz), corrected for participant bias, converted to microvolts, full-wave rectified and low-pass filtered at 6Hz using a 4th order recursive Butterworth filter. For MVIC exercises, a 100ms moving-average window (99 ms overlap) determined the maximum amplitude and a 500ms moving-average window (0ms overlap) determined maximum torque for each exercise (Hubley-Kozey et al. 2006). EMG data during the STS trials was normalized to the MVIC amplitude for each muscle, regardless of the exercise. For the calculation of muscle strength values, isometric torque values in Newton-meters (Nm) corrected for gravitational moments were averaged from the two test trials.

STS initiation was determined by visually identifying when the horizontal velocity of the trunk increased above base line prior to reaching maximum horizontal velocity. An electric switch determined seat-off when the buttocks lost contact with the seat. STS termination was identified by the sample where vertical velocity of the shoulder marker reached zero (Kuo, Tully & Galea 2009, Turcot et al. 2012), which was determined to be the preferred standing event method from an analysis in Appendix D.

The main dependent variables are in Table 5.1. Peak values were extracted from time-normalized joint angles, moments, and EMG ensemble average waveforms. Cubic spline interpolation was used to time-normalize waveforms to 100% from seat-off to standing and ensemble averages were calculated from at least three STS trials that were absent of any event timing or IRED data errors. Hip and knee flexion moment impulses (Nm·s/kg) and integrated EMG (%MVIC·s) were calculated for positive areas of non-time-normalized waveforms from seat-off to standing to account for both amplitude and duration.

Table 5.1 Dependent variables. Sagittal plane angles and moments, and electromyography.

Category	**Variables**
Sagittal Plane Angles (Trunk, Hip, Knee)	1) Peak
Sagittal Plane Net External Moments (Hip, Knee)	1) Peak 2) Impulse
Electromyography *Muscles* –VL, VM, RF, LH, GM	1) Peak 2) Integrated EMG

(VL – vastus lateralis, VM, - vastus medialis, RF – rectus femoris, LH – lateral hamstring, GM – gluteus maximus)

5.2.4 Statistical Analysis

All statistical tests were completed in Minitab™ Ver.16 (Minitab Inc. State College, PA, USA). Student t-tests determined significant group differences in age, mass, height, BMI and LEFS scores. The Kolmogorov-Smirnov test determined normality and the Levene's test determined equal variance. All data with unequal variances or non-normal distributions were transformed using a Johnson transformation method. For muscle strength, a two-factor mixed model Analysis of Variance model (ANOVA) tested for significant group (between) and leg (within) main effects and interactions (alpha=0.05).

For angle, moment, and EMG dependent variables, the legs were collapsed. A two-factor mixed ANOVA model tested for group (young and old), device (Power Seat™, Seat Assist™, ND-raised) differences accounting for between and within group main effects and interactions (alpha=0.05). Post-hoc testing determined pair-wise significant findings using Bonferonni adjusted alpha levels.

5.3 RESULTS

5.3.1 Participant Demographics and Strength Measures

Group demographics and anthropometrics are in Table 5.2. Ten healthy young adults (20-30 years old) and ten healthy older adults (>65 years old) participated in this study. Older adults were significantly older and had significantly worse self-reports of lower extremity function.

Table 5.2. Mean (SD) of participant demographics.

	Young Adults	Older Adults
N	10	10
Percent Female	50%	50%
Age (years)	**25 (2)**	**69 (3)***
Mass (kg)	77 (13)	75 (15)
Height (m)	1.77 (0.08)	1.70 (0.10)
BMI (kg/m^2)	24.3 (2.6)	25.8 (4.0)
WHR	0.79 (0.07)	0.89 (0.09)
LEFS (/80)	**79.9 (0.3)**	**73.1 (4.8)***

BMI = Body Mass Index
WHR = Waist to Hip Ratio
LEFS = Lower Extremity Functional Scale
* (Bold) indicate significant differences ($p<0.05$)

Isometric torque generated for different muscle groups are in table 5.3. No significant ($p>0.05$) group by leg interactions were found. Knee flexion torque was significantly different ($p<0.05$) between groups (25% greater in younger adults). Despite a lack of statistical significance ($p>0.05$), younger adults generated 20% and 25% greater isometric torque for the knee extensors and ankle plantarflexors, respectively.

Table 5.3. Mean (SD) of isometric torque output (Nm).

Muscle group	Young Adults		Older Adults		Group	Leg	Inter
	Dom	Non	Dom	Non			
Knee Extensors	151.1 (33.6)	163.9 (39.0)	129.9 (41.9)	124.4 (35.7)	0.078	0.933	0.195
Knee Flexors	83.6 (20.4)	85.8 (21.3)	64.2 (20.9)	62.5 (19.8)	**0.022**	0.832	0.930
Ankle Plantarflexors	102.7 (27.7)	108.7 (30.0)	77.2 (28.5)	80.3 (28.5)	0.067	0.211	0.562
Ankle Dorsiflexors	40.8 (7.7)	41.8 (6.9)	36.8 (13.9)	37.9 (13.3)	0.446	0.366	0.902

Dom (dominant), Non (non-dominant)

5.3.2 Temporal Characteristics

Table 5.4 provides descriptive statistics of the temporal characteristics associated with each seating condition for both age groups. There were no significant group by condition interactions ($p>0.05$) for any temporal variables. Older adults had significantly greater total STS time (approximately 0.4 seconds) and seat-off to standing time (approximately 0.3 seconds) compared to younger adults ($p<0.05$).

Total STS time with the SA was significantly greater compared to the PS and ND-raised (approximately 0.5 seconds) ($p<0.05$). Time from seat-off to standing was significantly greater for both the SA and PS compared to ND-raised (approximately 0.2 seconds) ($p<0.05$).

5.3.3 Angles

Peak sagittal plane trunk, hip, and knee angles are in Table 5.5. There were no significant group by condition interactions ($p>0.05$) for any of the peak flexion angles. Older adults performed the STS transfer with significantly greater peak hip flexion angles compared to younger adults by approximately 10° ($p<0.05$).

Peak trunk (figure 5.1) and hip flexion angles were significantly greater for the SA compared to the PS and ND-raised ($p<0.05$) by approximately 20° for the trunk and approximately 20-25° for the hip. Peak knee flexion angle was greatest for the SA, followed by the PS, and then the ND-raised ($p<0.05$). Peak knee flexion angles were approximately 20° greater for the SA and 4° greater for the PS compared to ND-raised. Other kinematic waveforms are in Appendix F.

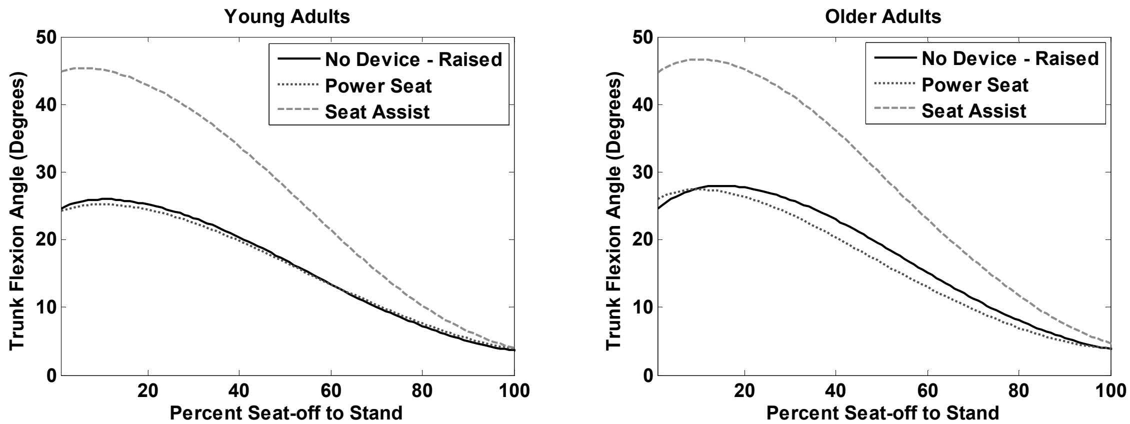


Figure 5.1. Ensemble average trunk flexion angle waveforms for the three different seating conditions between the younger adults (left) and older adults (right).

5.3.4 Moments of Force

Table 5.6 provides the peak and impulse values of the net external joint flexion moments for each seating condition in younger and older adults. There were no significant group by condition interactions ($p>0.05$).

Peak and impulse net external hip flexion moments values were significantly greater for the SA compared to the PS and ND-raised ($p<0.05$). Peak hip flexion moment were approximately 20% lower and impulse values were approximately 40% lower for the Seat Assist™. Peak net external knee flexion moments were significantly greater (approximately 15%) for the PS compared to the SA and ND-raised ($p<0.05$). Knee flexion moment impulse was significantly greater (approximately 25%) for the PS and the SA compared to ND-raised ($p<0.05$). Kinetic waveforms are in Appendix F.

5.3.5 Muscle Activation

Table 5.7 provides the peak and integrated EMG activity for each seating condition in younger and older adults. Integrated EMG of all five muscles were non-normally distributed. VL, VM, and GM were transformed using a sinusoidal function and RF and LH were transformed using a natural logarithm function. There were no significant group by condition interactions ($p>0.05$) for any EMG variables. Older adults had significantly greater peak EMG activity (% MVIC) and significantly greater integrated EMG (% MVIC/s) for all muscles ($p<0.05$).

Peak EMG activity of VL and VM was significantly greater for the PS and the SA compared to ND-raised ($p<0.05$). Peak EMG activity of the RF was significantly greater for the PS only compared to ND-raised ($p<0.05$). Peak EMG activity of the LH was greater for the SA compared to the PS only ($p<0.05$) and the peak activity of the GM was greater for the SA compared to both the PS and ND-raised ($p<0.05$). For the three quadriceps muscles, the integrated EMG was greater for the SA and the PS compared to ND-raised. For the LH and the GM, the integrated EMG was greater for the SA compared to the PS and ND. EMG waveforms are in Appendix F.

Table 5.4. Mean (SD) of temporal characteristics.

	Young Adults			Older Adults			Group	Device	Interaction
	ND-raised	PS	SA	ND-raised	PS	SA			
Total Time (s)	1.7(0.3)	1.7 (0.2)	2.2 (0.4)	2.1(0.7)	2.1 (0.5)	2.7 (0.8)	**0.048**	**<0.001**	0.661
Seat-off to stand (s)	0.9 (0.1)	1.0 (0.1)	1.1 (0.1)	1.1 (0.4)	1.3 (0.4)	1.4 (0.3)	**0.021**	**<0.001**	0.352

Table 5.5. Mean(SD) of peak flexion angles (degrees).

		Young Adults			Older Adults			Group	Device	Interaction
		ND-raised	PS	SA	ND-raised	PS	SA			
Trunk	Peak	26 (7)	26 (5)	46 (7)	28 (8)	28 (5)	48 (7)	0.49	**<0.001**	0.980
Hip	Peak	53 (9)	56 (8)	76 (13)	62 (11)	67 (7)	90 (11)	**0.017**	**<0.001**	0.208
Knee	Peak	59 (7)	63 (7)	73 (7)	56 (10)	61 (14)	74 (18)	0.619	**<0.001**	0.474

Table 5.6. Mean (SD) peak net external flexion moments (Nm/kg) and impulse (Nms/kg).

		Young Adults			Older Adults			Group	Device	Interaction
		ND-raised	PS	SA	ND-raised	PS	SA			
Hip	Peak	0.62 (0.22)	0.58 (0.20)	0.75 (0.19)	0.61 (0.16)	0.60 (0.14)	0.76 (0.19)	0.966	**<0.001**	0.909
	Impulse	0.28 (0.19)	0.29 (0.17)	0.50 (0.22)	0.36 (0.16)	0.41 (0.24)	0.61 (0.23)	0.259	**<0.001**	0.63
Knee	Peak	0.60 (0.19)	0.64 (0.12)	0.53 (0.09)	0.49 (0.19)	0.58 (0.14)	0.45 (0.14)	0.179	**<0.001**	0.495
	Impulse	0.26 (0.12)	0.29 (0.09)	0.34 (0.12)	0.22 (0.10)	0.32 (0.12)	0.32 (0.18)	0.905	**<0.001**	0.321

Table 5.7. Mean (SD) peak EMG activity (%MVIC) and integrated EMG (%MVICs).

		Young Adults			Older Adults			Group	Device	Interaction
		ND-raised	PS	SA	ND-raised	PS	SA			
VL	Peak	31 (10)	36 (11)	40 (10)	45 (22)	50 (18)	55 (18)	**0.042**	**<0.001**	0.925
	Int	16 (5)	19 (5)	23 (5)	27 (7)	49 (34)	52 (34)	**<0.001**	**<0.001**	0.280
VM	Peak	29 (10)	35 (9)	37 (8)	50 (24)	55 (24)	61 (24)	**0.011**	**<0.001**	0.566
	Int	14 (4)	17 (4)	21 (4)	28 (9)	49 (37)	56 (37)	**<0.001**	**<0.001**	0.705
RF	Peak	11 (3)	13 (5)	14 (4)	22 (14)	26 (14)	25 (13)	**0.017**	**0.024**	0.450.
	Int	7 (3)	9(4)	10 (3)	16 (8)	31 (36)	34 (36)	**0.001**	**0.001**	0.906
LH	Peak	10 (8)	8 (4)	11 (5)	19 (8)	16 (6)	24 (11)	**0.004**	**<0.001**	0.116
	Int	7 (4)	7(3)	8 (4)	16 (9)	26 (38)	34 (40)	**0.005**	**<0.001**	0.850
GM	Peak	16 (6)	15 (7)	19 (7)	23 (8)	23 (8)	30 (11)	**0.021**	**<0.001**	0.120
	Int	10 (4)	11(5)	14(6)	19 (9)	33 (36)	38 (36)	**0.003**	**<0.001**	0.398

5.3.6 Summary of Results

A summary of key significant findings for device and group main effects are in Table 5.8. These illustrate the between group differences for trunk angle only and for all muscle EMG peaks and integrals, and then several between device and between raised seats differences.

Table 5.8. Key significant findings ($p<0.05$).

	Device	Group
Peak Angles	**Trunk, Hip** SA>PS,ND-raised **Knee** SA>PS>ND-raised	**Trunk, Knee** No group differences **Hip** OA>YA
Peak Moments	**Hip** SA>PS,ND-raised **Knee** PS>SA,ND-raised	No group differences
Moment impulse	**Hip** SA>PS,ND-raised **Knee** PS,SA>ND-raised	No group differences
Peak EMG	**VL, VM** PS,SA> ND-raised **RF** PS>ND-raised *(SA not significantly different from either) **LH** SA>PS *(ND-raised not significantly different from either) **GM** SA>PS,ND-raised	All muscles OA>YA
Integrated EMG	**VL, VM, RF** PS,SA> ND-raised **GM, LH** SA>PS,ND-raised	All muscles OA>YA

5.4 DISCUSSION

The purpose of this chapter was to compare the Power Seat™ and the Seat Assist™ with an unassisted STS from a raised seat height to better understand how both devices affect lower limb biomechanics and muscle activation during a STS transfer. We hypothesized no differences between seating conditions, as the design goal of these lifting-seat devices is to raise the seat height. Several differences existed among devices and with the ND-raised condition, providing information for device prescription and for potential device modifications to alter biomechanical and neuromuscular demands.

Evidence provided in Chapter 4 showed that biomechanical and neuromuscular demands of a STS transfer were reduced during STS from a raised seat height. Lifting-seat devices have been designed to raise the seat height when placed on normal seating platforms. In a preliminary study (appendix A and B), both types of lifting-seat devices were compared with an unassisted rise at a normal seat height (peak angles, moments, and EMG). The Seat Assist™ reduced peak hip and knee moments and peak quadriceps muscle activity. The Power Seat™ reduced peak hip moments and peak activity of all muscles. Overall, both devices were effective at reducing some biomechanical and neuromuscular demands compared to an unassisted STS at a normal seat height

This was the first study comparing two lifting-seat devices with different mechanisms and also the first to compare either type of lifting-mechanism to a raised seat height. Both lifting-seat devices were designed with similar start and end points, raising the seat about 15cm and also anterior about 9cm. However, their lifting mechanisms are different and as a result, temporal, kinematic, kinetic, and EMG dependent variables differed between conditions. Both devices took longer from seat-off to standing than ND-raised and the Seat Assist™ had longer total STS duration. The Seat Assist™ significantly increased the peak trunk and hip flexion angles compared to the other two conditions, which suggests a different STS strategy was used with this device. The Seat-Assist™ is a pneumatic device that creates a mechanical preload as the user sits on the device and is thought to provide an assistive force as the user actively performs a STS. This mechanism differs from the Power Seat™ that transfers the user to a raised position before performing a STS transfer. In addition to greater peak trunk flexion angle, the Seat Assist™ also has the greatest peak knee flexion angle, which suggests the user does not maintain contact with the device to the completion of its rise. The Power Seat™ also has greater peak knee flexion angle compared to the ND-raised but less than the Seat Assist™. Overall, peak flexion angles for the Seat Assist™ were more similar to the ND-normal condition (found in chapter 4), whereas, Power Seat™ was more comparable to the ND-raised condition. Kinematic findings help to explain differences in some of the kinetic and EMG dependent variables between conditions.

Compared to no device at a raised seat, the Seat Assist™ had greater peak and impulse hip moments, peak EMG activity of the gluteus maximus, and integrated EMG of both the lateral hamstrings and gluteus maximus. From these results, the Seat Assist™ had greater demands on the hip joint and the hip extensor muscles compared to an unassisted STS from a raised seat height. The Seat Assist™ had similar peak knee flexion moments but had greater knee flexion moment impulse, which is likely attributed to greater duration from seat-off to standing. The Seat Assist™ resulted in more muscle agonist-antagonist co-activation at the knee joint compared to ND-raised, consistent with previous findings on a similar pneumatic design device (Munro, Steele 2000). The Seat Assist™ had greater peak EMG activity of the vastii muscles despite no difference in peak knee flexion moments. The increase in integrated EMG of the vastii muscles was approximately 75%, which was much greater than the increase of the knee flexion moment impulse of approximately 35%. Higher quadriceps activity related to high hamstring co-activity with the Seat Assist may lead to greater muscle effort and joint loading. While the EMG peaks for the quadriceps were lower for the SA compared to results for ND-Normal (from Chapter 4), integrated EMG of the quadriceps were not much different and integrated EMG of the LH and GM were much higher. The hip and knee biomechanical and neuromuscular findings suggest that this device may not be as beneficial for those with hip or knee joint pathologies or hip extensor or quadriceps strength deficits compared to its design goal of a raised seat height.

Compared to ND-raised, the Power Seat™ did not increase peak or impulses of hip flexion moments or the EMG activity of the hip extensor muscles and these values were all lower than the ND-Normal results in Chapter 4. The Power Seat™ did have greater peak and impulse of the knee flexion moment with greater corresponding peaks and integrated EMG activity of the quadriceps compared to ND-raised. High agonist-antagonist co-activation (lateral hamstring activity) was not found with the Power Seat. Overall, the Power Seat™ is the more suitable device for those with hip pathologies or hip extensor muscle deficits. However, the Power Seat™ has greater knee flexion moments and demands on the knee extensors muscles compared to an unassisted STS from a raised seat height.

This study provides support that healthy older adults maintain a similar STS kinematic and kinetic strategy compared to healthy younger adults, which is consistent with our hypothesis and previous studies (Hughes, Myers & Schenkman 1996, Ikeda et al. 1991, Akram, McIlroy 2012, Papa, Cappozzo 2000). However, older adults had greater peak and impulse EMG activity for all muscles, of which, lesser muscle strength could be a contributing factor. Older adults were at 20% weaker for their knee extensors and 25% for their knee flexors, though only significant for the knee flexors. Strength differences may also help to explain greater total STS and seat-off to standing durations in the older adult group. Interestingly, these temporal group differences for these three seating conditions differ from our previous analysis in chapter 4, showing no temporal group differences for unassisted STS transfers at different seat heights.

This was the first study to determine if the effects of lifting-seat devices were altered by age. No interaction effects were found for any of the biomechanical or neuromuscular variables, suggesting both lifting-seat devices have a similar effect on joint and muscle demands for both healthy younger adults and healthy older adults. Interestingly, Wretenberg et al. (1993) reported individuals with knee osteoarthritis had a reduction in hip flexion moments with a lifting-seat device, while the healthy younger group did not. These findings suggest that lifting-seat devices may affect individuals with pathologies differently. Therefore, this work cannot be extrapolated to individuals with pathologies, particularly frail, older adults with muscle impairments, but provides a baseline for comparison.

Interpretation of study findings need to be considered within the limitations of this study. In attempt to isolate the effect of both types of lifting-seat device and avoid the confounding effects of other determinants of STS transfers (Janssen, Bussmann & Stam 2002), set-up constraints were implemented. Foot position was controlled and participants were instructed to fold their arms across their chest to avoid the use of arms.

These constraints may limit their natural STS movements with the lifting-seats; particularly arm assistance, as it is advised in device instructions. Only healthy subjects were included in this study and the effect of lifting-seat devices may differ from target populations with musculoskeletal impairments.

5.5 SUMMARY

Older adults used similar mechanics but had greater peak muscular demands and overall muscle activation. Both devices had different effects on trunk, hip, and knee motion, hip and knee moments and overall muscle activation compared their design goal of a raised seat height. The pneumatic device changed the mechanics at the trunk, the hip, and the knee and increased muscle activation of all muscles. The electric-elevator device changed the mechanics at the knee and increased muscle activation of the quadriceps only. All three seating conditions had similar effects on both younger and older adults.

CHAPTER 6 CONCLUSION

The goal of this thesis was to understand how seat height and lifting-seat devices with different lifting mechanisms affect trunk, hip, and knee joint biomechanics and neuromuscular activity of surrounding muscles, and to determine whether the effects were altered by age. There were two main objectives addressed specifically in Chapters 4 and 5. Firstly, establishing the effect of seat height and age on lower limb biomechanics and muscle activation to provide information on whether a raised seat height is an appropriate design goal for lifting-seat devices. Secondly, to determine whether two different lifting-seat devices differ from their design goal (an unassisted STS from a raised seat height), and whether the effects are altered by age. Peak values were included to give an indication of the required moments of force and muscle activation (percentage of maximum) needed to complete the task. Integrated values that account for both amplitude and duration were included to give an indication of the overall muscle and joint demands during the STS. The findings for the specific objectives are summarized below:

6.1 SUMMARY OF KEY FINDINGS

6.1.1 Chapter 4

Chapter 4 results are summarized in table 4.8. Normal seat height was hypothesized (1a) to have greater peak trunk, hip, knee flexion angles and moments, greater hip and knee flexion moment impulses, greater peak and integrated EMG activity of the vastus lateralis, vastus medialis, rectus femoris, lateral hamstrings, and gluteus maximus compared to a raised seat height. Overall, the results presented in chapter 4 support this hypothesis with the exception of peak knee flexion moment. Though not statistically significant ($p=0.097$), the mean peak knee flexion moment decreased by 10-15% for older adults with a raised seat height. Overall, this study provides comprehensive findings of reduced biomechanical demands on the hip and the knee joints and neuromuscular demands of surrounding muscles with a raised seat height.

Older adults were hypothesized (1b) to have similar kinematics and kinetics of the trunk, hip, and knee and greater peak and impulse EMG activity for all muscles. Study results support this hypothesis. Muscle strength deficits in older adults could be a contributing factor to greater relative muscle demands (as a percentage of maximum). Older adults were 20% weaker for their knee extensors ($p>0.05$) and 25% weaker for their knee flexors ($p<0.05$). Additionally, older adults had greater antagonist-agonist muscle co-activation leading to greater peak and impulse EMG activity. Overall, older adults used similar mechanics but had greater muscular demands compared to younger adults.

Our results mainly support our third hypothesis (1c). Both seat heights had a similar effect on kinematic, kinetic, and EMG dependent variables for both groups. There was only one significant interaction effect ($p<0.05$). A greater reduction of integrated EMG activity with the raised seat height was found for older adults compared to younger adults. Despite a lack of statistical significance ($p>0.05$), integrated EMG of the VL and peak activity of the RF and the GM showed a similar trend. If the goal is to lower muscular demands, older adults may benefit more from raised seat height compared to younger adults.

6.1.2 Chapter 5

Chapter 5 results are summarized in table 5.8. No differences between seating conditions were hypothesized (2a) but was not supported by the results of this study. Both devices had different effects on trunk, hip, and knee motion, hip and knee moments and overall muscle activation compared to an unassisted rise from a raised seat height. The Seat Assist had greater peak flexion angles, knee moment impulse, peak and impulse hip flexion moments, and increased muscle activation of all muscles. The Power Seat had similar trunk and hip biomechanics as ND-raised but had greater peak knee flexion angle, peak and impulse knee flexion moments, and greater peak and integrated EMG of the quadriceps only.

Our results, presented in chapter 5, support hypothesis 2b. In order to maintain similar kinematic and kinetic strategies, older adults required greater peak and integrated EMG of all muscles, which is likely, in part due to strength differences and greater antagonist-agonist muscle co-activation. Older adults generated 20% less isometric torque output for their knee extensors and 25% less for their knee flexors, though only significant for the knee flexors.

There were no interaction effects supporting hypothesis 2c. The PS, the SA, and the ND-raised conditions affected both groups the same in terms of kinematic, kinetic, and EMG dependent variables.

6.2 IMPLICATIONS

The knowledge gained from this thesis should have direct implications on several aspects of STS research including; differences between older and younger adults, the effect of seat height, and the effect of lifting seat devices.

6.2.1 Effect of Age on a Sit-to-Stand Transfer

This study provides support that healthy older adults maintain a similar STS kinematic and kinetic strategy compared to healthy younger adults, which is consistent with previous studies comparing between different healthy age groups (Hughes, Myers & Schenkman 1996, Ikeda et al. 1991, Akram, McIlroy 2012, Papa, Cappozzo 2000). This study was the first to examine hip and knee flexion moment impulses to provide information on the overall joint demands and were similar between groups. Contrary to other studies, older adults did not have greater trunk flexion characteristic of a 'stabilization strategy' to reduce the knee flexion moments and demands on the quadriceps muscles (Schenkman, Rlley & Mann 1990, Fujimoto, Chou 2012, Papa, Cappozzo 2000, Gross et al. 1998, Yoshioka et al. 2007). The older adults in this study were only 20% weaker in their knee extensor muscles and as such, may not have had to change their STS strategy. Savelberg et al. (2007), provided evidence that individuals do not change their kinematic and kinetic STS strategy until a threshold level when their knee extensor muscles become overloaded.

This study was the first to examine muscle activation between healthy older and younger adults. Consistent with findings by Hughes et al. (1996), older adults had to work at a greater percentage of maximum to generate similar joint moments. Older adults had greater EMG activity (peaks and integrated) of all muscles, which could be related to a couple of factors. Firstly, older adults had greater strength deficits by approximately 20-25% for the knee flexors, knee extensors, and ankle plantarflexors. Secondly, older adults had greater antagonist-agonist muscle co-activation during the STS transfers, which is consistent with findings during other functional tasks (Nagai et al. 2011, Quirk, Hubley-

Kozey 2012). Compared to younger adults, older adults had 100% greater peak lateral hamstrings activity and 25-75% greater peak quadriceps activity during the STS transfers, though only 20-25% less maximum isometric torque output in these muscle groups. Greater co-activation causes greater muscle effort and potentially greater joint loading. Additionally, this is the first study to report the values of peak and integrated EMG activity for both healthy younger and older adults. Older adults had peak EMG values for the vastus medialis of 79 %MVIC and for the vastus lateralis of 69 %MVIC from a normal chair height, highlighting the high demands on the quadriceps muscles during a STS transfer for this age group.

This study provided comprehensive evidence on the effect of age on STS transfer analyzing trunk, hip, and knee kinematics, kinetics, and EMG activity surrounding muscles. Both healthy younger and older adults have similar joint kinematics and kinetics but older adults have greater peak and overall muscle activation during the task.

6.2.2 Effect of Seat Height on a Sit-to-Stand Transfer

Overall, the results of this study support previous findings that a higher than normal seat height reduces the biomechanical demands during a STS transfer (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000, Gillette, Stevermer 2012, Su, Lai & Hong 1998, Schenkman, Riley 1996). The raised seat height condition reduced peak trunk, hip, and the knee flexion angles and reduced peak hip and knee flexion moments, though the peak knee flexion moment was not significant ($p>0.05$). This was the first study to show decreased hip and knee flexion moment impulses with a raised seat height, indicating a reduction in overall joint demands from seat-off to standing.

Similar to previous findings, the raised seat height reduced peak muscle activity of the vastii muscles (Arborelius, Wretenberg & Lindberg. 1992, Munro, Steele 2000). In this study, there was also a reduction in peak muscle activity of the rectus femoris and a hamstring muscle, which disagrees with findings by Arborelius et al. (1992). This study was the first to examine and show a reduction of peak gluteus maximus activity with a raised seat height. This was also the first study to show decreased integrated EMG for all muscles with a raised seat height (except VM), which has implications for reducing overall muscle effort during the task.

There have only been two previous studies comparing between seat heights in healthy older and younger adults (Chen et al. 2010, Schenkman, Riley 1996). Both groups were affected similarly by different seat heights in terms of subjective ratings (Chen et al. 2010) and angular excursions and velocities (Schenkman, Riley 1996). This was the first study to examine the effect of seat height between healthy older and younger adults on joint moments and muscle activation. Overall, the effects of a raised seat height on joint and muscle demands were not different between age groups. Only one significant interaction effect was found for integrated EMG of the VM; a greater reduction of overall EMG activity with the raised seat height for older adults compared to younger adults. Similar trends were found for integrated EMG of the VL and peak activity of the RF and the GM, though not significant. This suggests that older adults may benefit more from raised seat height compared to younger adults in terms of muscular demands.

This study has provided evidence that a raised seat height reduces the maximum and overall biomechanical and neuromuscular demands required during a STS transfer task. As such, seating platforms with raised seat heights could be used for individuals with STS limitations, particularly for frail older adults with muscle impairments or pathologies. Lifting-seat devices have been designed to raise the effective seat height when placed on normal seating platforms. This design goal is supported by the findings of this study. How these devices differ from an unassisted STS from a raised seat height was the focus of chapter 5.

6.2.3 Effect of Lifting-Seat Devices on a Sit-to-Stand Transfer

There is limited comprehensive evidence on the effect of either type of lifting-seat device on a STS transfer. This was the first study to combine a biomechanical analysis in addition to a surface EMG analysis from muscles surrounding both the hip and the knee joints to give a more comprehensive picture of how these two devices affect a STS transfer.

Evidence provided in Chapter 4 showed that biomechanical and neuromuscular demands of a STS transfer were reduced during STS from a raised seat height. Lifting-seat devices have been designed to raise the seat height when placed on normal seating platforms. In a preliminary study (appendix A and B), both types of lifting-seat devices were compared with an unassisted rise at a normal seat height. The Seat Assist™ reduced peak hip and knee moments and peak quadriceps muscle activity. The Power Seat™ reduced peak hip moments and peak activity of all muscles. Overall, both devices were effective at reducing some biomechanical and some neuromuscular demands. A question that remained and formed the motivation for chapter 5 - how does either type of lifting-seat device differ from their design goal of a raised seat height?

We hypothesized that there would be no differences between the three seating conditions but we did find several key differences, highlighting how these devices differ from their design goal. Both types of lifting-seat devices were designed with similar start and end points, raising the seat about 15cm and forward about 9cm. However, their lifting mechanisms are different and as a result, their effect on STS kinematic strategy, joint kinetics, and lower limb muscle activation differed as well. Participants had greater trunk flexion with the Seat Assist™ similar to ND-normal (chapter 4) as the user actively performs a STS from the starting position of this device. Subjects used less trunk flexion (similar to ND-raised) with the Power Seat™ as the user passively rides the device to its extended position before performing the STS. In addition to greater peak trunk flexion angle, the Seat Assist™ also has the greatest peak knee flexion angle, which suggests the user does not maintain contact with the device to the completion of its rise. Overall, the STS kinematic strategy with the Seat Assist™ in terms of peak trunk, hip, and knee angle is much closer to the ND-normal condition, whereas, the Power Seat™ is much closer to the ND-raised condition.

Compared to no device at a raised seat, the Seat Assist™ had greater peak and overall hip moments and greater corresponding activation of the hip extensor muscles. The Power Seat™ did not have this effect and the hip flexion moments and activity of the lateral hamstring and gluteus maximus were all lower than the ND-normal results in Chapter 4. From these results, only the Power Seat™ may be beneficial for those with hip extensor strength deficits or hip pathologies.

Compared to ND-raised, the Seat Assist™ did not increase peak knee flexion moments but did increase knee flexion moment impulse. Additionally, this device caused more muscle agonist-antagonist co-activation at the knee joint compared to ND-raised, consistent with previous findings on a similar pneumatic design device (Munro, Steele 2000). Higher quadriceps activity related to high hamstring co-activity with the Seat

Assist may lead to greater muscle effort and joint loading. The Power Seat™ had greater peak and impulse of the knee flexion moment with greater corresponding peaks and integrated EMG activity of the quadriceps compared to ND-raised. From these results, both devices increased knee biomechanical and neuromuscular demands compared to ND-raised, which suggests that these devices would not be as beneficial to those with knee joint pathologies or knee extensor strength deficits, which have been shown to be the limiting muscle group during a STS transfer (Hughes, Myers & Schenkman 1996, Savelberg et al. 2007, Van der Heijden et al. 2009).

This was the first study to determine if the effects of lifting-seat devices were altered by age. No interaction effects were found for any of the biomechanical or neuromuscular outcomes, suggesting both lifting-seat devices have a similar effect on joint and muscle demands for both healthy younger adults and healthy older adults. Wretenberg et al. (1993) found individuals with knee osteoarthritis had a reduction in hip flexion moments with a lifting-seat device, while the healthy younger group did not. These findings suggest that lifting-seat devices affect individuals with pathologies differently.

6.3 FUTURE WORK

1) This pilot study investigated the effect seat height and two types of lifting-seat devices had on a STS transfer with a secondary goal of assessing experimental protocol feasibility and safety on healthy younger and healthy older adults. Lifting-seat devices are designed for individuals with STS impairments due to pathologies or muscle impairments and future studies are needed on these target populations. Evidence exists to suggest lifting-seat devices do not effect healthy and pathological in the same manner (Wretenberg et al. 1993). Information on how either type of lifting-seat device affects individuals with STS impairments and how they differ from healthy groups will provide further evidence to guide device prescription and innovation.

2) Several constraints were imparted on the subject in order to avoid confounding effects in an attempt to isolate the effects of lifting-seat devices. Foot position was kept constant and arms were folded across the subjects' chest to avoid their involvement. These

constraints may have prevented subjects' natural STS movement and interaction with the devices. In particular, device manuals provided by Uplift Technologies Inc. instruct the use of hand rests for stability and thus, may limit the applicability of these findings. Future work comparing constrained STS transfers and non-constrained STS transfers with self-selected foot position and/or arm position would help determine the generalizability of the results of this study.

3) Future work is needed on innovative approaches to alter the degree of trunk flexion to reduce either the demands at the knee joint or the hip joint. Previous studies have shown an important trade-off between hip and knee moments with trunk flexion (Fujimoto, Chou 2012, Papa, Cappozzo 2000, Scarborough, McGibbon & Krebs 2007). Similarly, in this study, greater trunk flexion angle with the Seat Assist™ led to greater peak hip flexion moments but lesser peak knee flexion moments than the Power Seat™. This trade-off highlights the importance of considering the kinematic strategy of the user in lifting-seat device design or device instructions. Based on this information, one lifting-seat device may not be appropriate for all individuals with STS impairments but rather different designs for those with knee pathologies and hip pathologies. A device similar to the Power Seat™ that promotes little trunk flexion will decrease hip moments, making it better for those with hip pathologies. Whereas, a device similar to the Seat-Assist that promotes more trunk flexion will decrease knee flexion moments, making it better for those with knee pathologies. Two innovative approaches are outlined below:

i) Developing a user training program specific for those with knee pathologies or specific to hip pathologies. Knee-specific device instructions would focus on greater trunk flexion at seat-off to shift the center of mass more anterior over the base of support to reduce knee flexion moments and demands on the knee extensor muscles. Hip-specific instructions would focus on having a more extended trunk to reduce the hip flexion moments and demands on the hip extensor muscles.

ii) A seat surface pressure sensor that detects the anterior-posterior center of pressure location could be implemented in an electric-elevator device. For a 'knee design', the pressure sensor would prevent the device from rising unless the trunk flexed forward causing the center of pressure to shift to the anterior portion of the seat surface.

For a ‘hip design’, the device would not rise unless the center of pressure was on the posterior portion of the seat surface from the trunk being more extended.

Future work is needed on both of these methods to determine their feasibility, safety, and effectiveness at reducing either knee flexion moments and quadriceps muscle activation or hip flexion moments and hip extensor muscle activation.

In conclusion the results from this thesis further our understanding of the effects of age, seat height, and lifting-seat devices on hip and knee kinematics, kinetics, and muscle activation of surrounding muscles. Both younger and older adults had similar trunk, hip, and knee biomechanics but older adults had greater activation of lower limb muscles. Raised seat height reduced the maximum and overall biomechanical and neuromuscular demands required during a STS transfer task, and the effect was similar in both groups. Both devices had different effects on trunk, hip, and knee motion, hip and knee moments and overall muscle activation compared to an unassisted rise from a raised seat height. The Seat Assist™ altered trunk, hip, and knee kinematics and kinetics and had greater muscle activation of all muscles. The Power Seat™ had greater biomechanical demands of the knee joint and greater muscle activation of the quadriceps only. All three seating conditions had similar effects on both younger and older adults.

REFERENCES

"Chapter 2: Physical Activity has many health benefits" 2008, in *Physical Activity Guidelines for Americans*.

Akram, S.B. & McIlroy, W.E. 2012, "Challenging Horizontal Movement Sit-to-Stand : Impact on Stability in the Young and Elderly", *Journal or Motor Behaviour,* vol. 43, no. 2, pp. 147-153.

Arborelius, U.P., Wretenberg, P. & Lindberg. 1992, "The effects of armrests and high seat heights on lower-limb joint load and muscular activity during sitting and rising", *Ergonomics,* vol. 35, no. 11, pp. 1377-1391.

Ashford, S. & Hospital, R. 2000, "A comparison of the timing of muscle activity during sitting down compared to standing up", *Physiotherapy Research International,* vol. 5, no. 2, pp. 111-128.

Bashford, G.M., Steele, J.R., Munro, B.J., Westcott, G., & Jones, M.E. 1998, "Ejector chairs: Do they work and are they safe?," *Australian Occupational Therapy Journal*, vol. 45, pp. 99-106. 126

Binkley, J.M., Stratford, P.W., Lott, S.A. & Riddle, D.L. 1999, "The Lower Extremity Functional Scale (LEFS): scale development, measurement properties, and clinical application. North American Orthopaedic Rehabilitation Research Network", *Physical Therapy,* vol. 79, no. 4, pp. 371-383.

Boonstra, M.C., Schwering, P.J.A., De Waal Malefijt, M.C., & Verdonschot, N. 2010, "Sit-to-stand movement as a performance-based measure for patients with total knee arthroplasty", *Physical Therapy,* vol. 90, no. 2, pp. 149-156.

Burdett, R.G., Habasevich, R., Pisciotta, J., & Simon, S.R. 1985, "Biomechanical comparison of rising from two types of chairs", *Physical Therapy*, vol. 65, no. 8, pp. 1177-1183.

Burnett, D.R., Campbell-Kyureghyan, N.H., Cerrito, P.B. & Quesada, P.M. 2011, "Symmetry of ground reaction forces and muscle activity in asymptomatic subjects during walking, sit-to-stand, and stand-to-sit tasks", *Journal of Electromyography and Kinesiology,* vol. 21, no. 4, pp. 610-615.

Burnfield, J.M., Shu, Y., Buster, T.W., Taylor, A.P., McBride, M.M. & Krause, M.E. 2012, "Kinematic and electromyographic analyses of normal and device-assisted sit-to-stand transfers", *Gait & Posture,* vol. 36, no. 3, pp. 516-522.

Camargos, A.C.R., Rodrigues-de-Paula-Goulart, F. & Teixeira-Salmela, L.F. 2009, "The effects of foot position on the performance of the sit-to-stand movement with chronic stroke subjects", *Archives of Physical Medicine and Rehabilitation,* vol. 90, no. 2, pp. 314-319.

Chen, S., Lee, Y., Chiou, W. & Chen, Y. 2010, "A pilot study examining seat heights and subjective ratings during rising and sitting", *International Journal of Industrial Ergonomics,* vol. 40, no. 1, pp. 41-46.

Christiansen, C.L., Bade, M.J., Judd, D.L. & Stevens-Lapsley, J.E. 2011, "Weight-bearing asymmetry during sit-stand transitions related to impairment and functional mobility after total knee arthroplasty", *Archives of Physical Medicine and Rehabilitation,* vol. 92, no. 10, pp. 1624-1629.

Christiansen, C.L. & Stevens-Lapsley, J.E. 2010, "Weight-bearing asymmetry in relation to measures of impairment and functional mobility for people with knee osteoarthritis", *Archives of Physical Medicine and Rehabilitation,* vol. 91, no. 10, pp. 1524-1528.

Costigan, P.A., Wyss, U.P., Deluzio, K.J. & Li, J. 1992, "Semiautomatic three-dimensional knee motion assessment system", *Medical & Biological Engineering & Computing,* vol. 30, no. 3, pp. 343-350.

Crockett, K., Lanovaz J. & Arnold C. 2012, "Kinematics of the 30-second chair stand in healthy older adults versus older adults at risk of falls", *Canadian Society of Biomechanics,* Burnaby, British Columbia, June 6-9, Conference Abstract.

de Souza, L.A.P.S., Curtarelli, M.D.B., Mukherjee, M. & Dionisio, V.C. 2011, "The effect of the partially restricted sit-to-stand task on biomechanical variables in subjects with and without Parkinson's disease", *Journal of Electromyography and Kinesiology,* vol. 21, no. 5, pp. 719-726.

Dehail, P., Bestaven, E., Muller, F., Mallet, A., Robert, B., Bourdel-Marchasson, I. & Petit, J. 2007, "Kinematic and electromyographic analysis of rising from a chair during a Sit-to-Walk task in elderly subjects: role of strength", *Clinical Biomechanics (Bristol, Avon),* vol. 22, no. 10, pp. 1096-1103.

Doorenbosch, C.A.M., Harlaar, J., Roebroeck, M.E. & Lankhorst J. 1994, "Two strategies of transferring from sit-to-stand the activation of monoarticular and biarticular", *Journal of Biomechanics,* vol. 27, no. 11, pp. 1299-1307.

Edlich, R.F., Heather, C.L., Galumbeck M.H., 2003, "Revolutionary advances in adaptive seating systems for the elderly and persons with disabilities that assist sit-to-stand tranfers", *Journal of Long-term Effects of Medical Implants,* vol. 13, no. 1, pp. 31-39.

Epifanio, I., Avila, C., Page, A. & Atienza, C. 2008, "Analysis of multiple waveforms by means of functional principal component analysis: normal versus pathological patterns in sit-to-stand movement", *Medical & Biological Engineering & Computing,* vol. 46, no. 6, pp. 551-561.

Etnyre, B. & Thomas, D.Q. 2007, "Event standardization of sit-to-stand movements", *Physical Therapy,* vol. 87, no. 12, pp. 1651-1666.

Farquhar, S.J., Kaufman, K.R. & Snyder-Mackler, L. 2009, "Sit-to-stand 3 months after unilateral total knee arthroplasty: comparison of self-selected and constrained conditions", *Gait & Posture,* vol. 30, no. 2, pp. 187-191.

Fujimoto, M. & Chou, L. 2012, "Dynamic balance control during sit-to-stand movement: an examination with the center of mass acceleration", *Journal of Biomechanics,* vol. 45, no. 3, pp. 543-548.

Gillette, J.C. & Stevermer, C.A. 2012, "The effects of symmetric and asymmetric foot placements on sit-to-stand joint moments", *Gait & Posture,* vol. 35, no. 1, pp. 78-82.

Goulart, R. & Valls-sole, J. 1999, "Patterned electromyographic activity in the sit-to-stand movement", *Clinical Neurophysiology*, vol. 110, pp. 1634-1640.

Grood, E.S. & Suntay, W.J. 1983, "A joint coordinate system for the clinical description of three-dimensional motions: application to the knee", *Journal of Biomechanical Engineering,* vol. 105, no. 2, pp. 136-144.

Gross, M.M., Stevenson, P.J., Charette, S.L., Pyka, G. & Marcus, R. 1998, "Effect of muscle strength and movement speed on the biomechanics of rising from a chair in healthy elderly and young women", *Gait & Posture,* vol. 8, no. 3, pp. 175-185.

Guralnik, J.M., Ferrucci, L., Simonsick, E.M., Salive, M.E. & Wallace, R.B. 1995, "Lower-extremity function in persons over the age of 70 years as a predictor of subsequent disability", *The New England Journal of Medicine,* vol. 332, no. 9, pp. 556-561.

Hirvensalo, M., Rantanen, T. & Heikkinen, E. 2000, "Mobility difficulties and physical activity as predictors of mortality and loss of independence in the community-living older population", *Journal of the American Geriatrics Society,* vol. 48, no. 5, pp. 493-498.

Hodge, W.A., Carlson, S.M., Fijan, R.S., Burgess, S.B., Riley, P.O., Harris, W.H. & Mann R.W. 1989, "Contact Pressures from an Instrumented Hip Endoprosthesis", *The Journal of Bone and Joint Surgery,* vol. 71-A, no. 9, pp. 1378-1386.

Hubley-Kozey, C.L., Deluzio, K.J., Landry, S.C., McNutt, J.S. & Stanish, W.D. 2006, "Neuromuscular alterations during walking in persons with moderate knee osteoarthritis", *Journal of Electromyography and Kinesiology,* vol. 16, no. 4, pp. 365-378.

Hughes, M.A., Myers, B.S. & Schenkman, M.L. 1996, "The role of strength in rising from a chair in the functionally impaired elderly", *Journal of Biomechanics,* vol. 29, no. 12, pp. 1509-1513.

Hughes, M.A., Myers, B.S. & Schenkman, M.L. 1996, "The role of strength in rising from a chair in the functionally impaired elderly", *Journal of Biomechanics,* vol. 29, no. 12, pp. 1509-1513.

Ikeda, E., Schenkman, M., Riley, P. & Hodge W.A. 1991, "Influence of Age on Dynamics of Rising from a Chair", *Physical Therapy*. vol. 71, no. 6., pp. 473-481.

Janssen, W., Bussmann, H. & Stam H.J., 2002, "Determinants of the sit-to-stand movement: a review", *Physical Therapy,* vol. 82, no. 9, pp. 866-879.

Jeyasurya, J. 2011, "Biomechanical analysis of assisted sit to stand", *Thesis, Faculty of Graduate Studies, University of British Columbia,* April.

Khemlani, M.M., Carr, J.H. & Crosbie, W.J. 1999, "Muscle synergies and joint linkages in sit-to-stand under two initial foot positions", *Clinical Biomechanics,* vol. 14, pp. 236-246.

Kuo, Y., Tully, E.A. & Galea, M.P. 2009, "Kinematics of sagittal spine and lower limb movement in healthy older adults during sit-to-stand from two seat heights", *Spine*, vol. 35, no. 1, pp. 1-7.

Landry, S.C., McKean, K.A., Hubley-Kozey, C.L., Stanish, W.D. & Deluzio, K.J. 2007, "Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed", *Journal of Biomechanics,* vol. 40, no. 8, pp. 1754-1761.

Lundin, T.M., Grabiner, M.D. & Jahnigen, D.W., 1995, "On the assumption of bilateral lower extremity joint moment symmetry during the sit-to-stand task", *Journal of Biomechanics,* vol. 28, no. 1, pp. 109-112.

Mak, M.K.Y., Levin, O., Mizrahi, J. & Hui-Chan, C.W.Y. 2003, "Joint torques during sit-to-stand in healthy subjects and people with Parkinson's disease", *Clinical Biomechanics,* vol. 18, no. 3, pp. 197-206.

McKean, K.A., Landry, S.C., Hubley-Kozey, C.L., Dunbar, M.J., Stanish, W.D. & Deluzio, K.J. 2007, "Gender differences exist in osteoarthritic gait", *Clinical Biomechanics (Bristol, Avon),* vol. 22, no. 4, pp. 400-409.

McLeod, P.C., Kettelkamp, D.B., Srinivasan, V. & Henderson, O.L. 1975, "Measurements of repetitive activities of the knee", *Journal of Biomechanics,* vol. 8, no. 6, pp. 369-373.

Mizner, R.L. & Snyder-Mackler, L. 2005, "Altered loading during walking and sit-to-stand is affected by quadriceps weakness after total knee arthroplasty", *Journal of Orthopaedic Research,* vol. 23, pp. 1083-1090.

Munro, B.J. & Steele, J.R. 2000, "Does using an ejector chair affect muscle activation patterns in rheumatoid arthritic patients ? A preliminary investigation", *Journal of Electromyography and Kinesiology,* vol. 10, pp. 25-32.

Munro, B.J., Steele, J.R., Bashford, G.M., Ryan, M. & Britten, N. 1998, "A kinematic and kinetic analysis of the sit-to-stand transfer using an ejector chair: implications for elderly rheumatoid arthritic patients", *Journal of Biomechanics,* vol. 31, pp 263-271.

Nagai, K., Yamada, M., Uemura, K., Yamada, Y., Ichihashi, N. & Tsuboyama, T. 2011, "Differences in muscle coactivation during postural control between healthy older and young adults", *Archives of Gerontology and Geriatrics,* vol. 53, no. 3, pp. 338-343.

Pai, Y. & Rogers M.W. 1991, "Speed variation and resultant joint torques during sit-to-stand", *Archives of Physical Medicine and Rehabilitation,* vol. 72, no. 11, pp. 881-885.

Papa, E. & Cappozzo, A. 2000, "Sit-to-stand motor strategies investigated in able-bodied young and elderly subjects", *Journal of Biomechanics,* vol. 33, no. 9, pp. 1113-1122.

Patsika, G., Kellis, E. & Amiridis, I.G. 2011, "Neuromuscular efficiency during sit to stand movement in women with knee osteoarthritis", *Journal of Electromyography and Kinesiology,* vol. 21, no. 5, pp. 689-694.

Quirk, A. & Hubley-Kozey, C.L. 2012, "Age-related changes in neuromuscular response of trunk muscles during a dynamic leg loading perturbation task", *Canadian Society of Biomechanics, Burnaby, British Columbia,* June 6-9, Conference Abstract.

Robbins, S.M., Birmingham, T.B., Jones, G.R., Callaghan, J.P., 2009, "Developing an estimate of daily cumulative loading for the knee: examining test-retest reliability", *Gait & Posture*, vol. 30, no. 4, pp 497-501.

Rodosky, M., Andriacchi, T. & Andersson G.B.J. 1989, "The influence of chair height on lower limb mechanics during rising", *Journal of Orthopaedic Research,* vol. 7, no. 2, pp. 266-271.

Rutherford, D.J., Hubley-Kozey, C.L., Deluzio, K.J., Stanish, W.D. & Dunbar, M. 2008, "Foot progression angle and the knee adduction moment: a cross-sectional investigation in knee osteoarthritis", *Osteoarthritis and Cartilage,* vol. 16, no. 8, pp. 883-889.

Rutherford, D.J., Hubley-Kozey, C.L. & Stanish, W.D. 2011, "Maximal voluntary isometric contraction exercises: a methodological investigation in moderate knee osteoarthritis", *Journal of Electromyography and Kinesiology,* vol. 21, no. 1, pp. 154-160.

Savelberg, H.H.C.M., Fastenau, A., Willems, P.J.B. & Meijer, K. 2007, "The load/capacity ratio affects the sit-to-stand movement strategy", *Clinical Biomechanics (Bristol, Avon),* vol. 22, no. 7, pp. 805-812.

Scarborough, D.M., McGibbon, C.A. & Krebs D.E. 2007, "Chair rise strategies in older adults with functional limitations", *Journal of Rehabilitation Research and Development,* vol. 44, no. 1, pp. 33-42.

Scarborough, D.M., Krebs, D.E. & Harris, B.A. 1999, "Quadriceps muscle strength and dynamic stability in elderly persons", *Gait & Posture,* vol. 10, no. 1, pp. 10-20.

Schenkman, M. & Riley, P. 1996, "Sit to stand from progressively lower seat heights - alterations in angular velocity", *Clinical Biomechanics,* vol. 11, no. 3, pp. 153-158.

Schenkman, M., Rlley, P. & Mann, R.W. 1990, "Whole-body movements during rising to standing from sitting", *Physical Therapy,* vol. 70, no. 10, pp. 638-651.

Seedhom, B. & Terayama, K. 1976, "Knee forces during the activity of getting out of a chair with and without the aid of arms", *Biomedical Engineering,* vol. 11, no. 8, pp. 278-282.

Shepherd, R. & Koh, H.P. 1996, "Some biomechanical consequences of varying foot placement in sit-to-stand in young women", *Scandinavian Journal of Rehabilitation Medicine,* vol. 28, no. 2, pp. 79-88.

Stegeman, D.F. & Hermens, H.J. 1999, "Standards for surface electromyography : the European project" *Surface EMG for Non-Invasive Assessment of Muscles (SENIAM)",* pp. 108-112.

Su, F.C., Lai, K.A. & Hong, W.H. 1998, "Rising from chair after total knee arthroplasty", *Clinical Biomechanics (Bristol, Avon),* vol. 13, no. 3, pp. 176-181.

Turcot, K., Armand, S., Fritschy, D., Hoffmeyer, P. & Domizio, S. 2012, "Sit-to-stand alterations in advanced knee osteoarthritis", *Gait & Posture,* vol. 36, no. 1, pp. 68-72.

Van der Heijden, M.M.P., Meijer, K., Willems, P.J.B. & Savelberg, H.H.C.M. 2009, "Muscles limiting the sit-to-stand movement: an experimental simulation of muscle weakness", *Gait & Posture,* vol. 30, no. 1, pp. 110-114.

Vaughan, C.L., Davis, B.L., & O'Connor, J.C. 1992, *Dynamics of Human Gait,* Human Kinetics Publishers, Champaign, Illinois, USA.

Winter, D.A., Fuglevand, A.J. & Archer, S.E. 1994, "Crosstalk in surface electromyography: theoretical and practical estimates", *Journal of Electromyography and Kinesiology,* vol. 4, no. 15., pp 15-26.

Winter, D.A. 2009. *Biomechanics and Motor Control of Human Movement*, *Fourth Edition*, John Wiley & Sons. New York City, New York, USA.

Wretenberg, P., Arborelius, U., Weidenhielm, L. & Lindberg, F. 1993, "Rising from a chair by a spring-loaded flap seat: A biomechanical analysis", *Scandinavian Journal of Rehabilitation Medicine,* vol. 25, pp. 153-159.

Yoshioka, S., Nagano, A., Hay, D.C. & Fukashiro, S. 2009, "Biomechanical analysis of the relation between movement time and joint moment development during a sit-to-stand task", *Biomedical Engineering Online,* vol. 8, no. 2, pp. 27-36.

Yoshioka, S., Nagano, A., Himeno, R. & Fukashiro, S. 2007, "Computation of the kinematics and the minimum peak joint moments of sit-to-stand movements", *Biomedical Engineering Online,* vol. 6, pp. 26-40.

Young, P. 1972, "Spring-booster chairs for patients with muscular disability", *Physical Therapy,* vol. 52, no. 12, pp. 1286-1289.

APPENDIX A PRELIMINARY STUDY FINDINGS: THE EFFECT OF LIFTING-SEAT DEVICES ON A SIT-TO-STAND TRANSFER COMPARED TO AN UNASSISTED STS FROM A RAISED SEAT HEIGHT

A.1 PURPOSE

To determine the effect of two lifting-seat devices with different lifting mechanisms compared to an unassisted STS transfer from a normal seat height and age on trunk, hip and knee biomechanics and muscle activation. Three different seating conditions were compared; an electric-elevator design lifting-seat device (Power Seat™), a pneumatic design lifting-seat device (Seat Assist™), and no device at a normal seat height.

A.2 METHODOLOGY

Study methodology has been previously described in Chapter 3 with respect to study participants, data collection, and data processing. Additional information on data processing and analysis unique to this preliminary study is presented below. Each participant performed five trials of three different seating conditions; 1) Seat-Assist (SA) (Seat Assist™ UPE-1, Uplift Technologies Inc., Dartmouth, NS, Canada), 2) Power Seat™ (PS) (Power Seat™ UPE-P100Ex 24VDC, Uplift Technologies Inc., Dartmouth, NS, Canada), 3) no device from a normal seat height (ND-normal). Before every trial, the participant was positioned fully on the chair with the knee angle positioned at 90° using a standard goniometer and the feet positioned parallel about shoulder width apart. The participant performed several trials to ensure they were performing each condition at a consistent self-selected speed and movement strategy. Participants were instructed to remain on the seat pan in an erect posture facing forward until a signal was given to begin the STS trial. In addition, they were instructed to fold their arms across their chest, keep feet on the floor, and face forward until trial completion. The ND-normal trials were performed first on the Power Seat™ in the closed position, followed by the two lifting-seat conditions in a random order. For the PS trials, participants pulled the lever to engage the seat when the signal was given. Participants were instructed to keep their feet planted on the ground and remain on the seat until maximum elevation and then stand up at self-selected pace after the Power Seat™ stops. For the SA trials, the participants were instructed to perform a comfortable STS movement with self-elected speed and movement strategy. For all conditions, once standing, participants remained standing for the duration of the trial recording.

Peak values were extracted from joint angles, moments, and EMG waveforms. For peaks values, the legs were collapsed. Group (young and old) and device (Power Seat™, Seat Assist™, ND-normal) differences were tested using a two-factor mixed model ANOVA that accounts for between and within group main effects and interactions (alpha=0.05). Participants were the only random factor in the ANOVA model. Post-hoc testing was employed for determining pair-wise significant findings using Bonferonni adjusted alpha levels. Statistical procedures were completed in Minitab™ Ver.16 (Minitab Inc. State College, PA, USA).

A.3 RESULTS

Key significant findings for device and group main effects (p<0.05) are presented in table A.1.

Table A.1. Preliminary study key significant findings (p<0.05).

	Device	Group
Peak Angles	**Trunk, Hip** ND-normal,SA > PS **Knee** ND-normal>SA>PS	**Trunk, Knee** No Group Differences **Hip** OA>YA
Peak Moments	**Hip** ND-normal>SA>PS **Knee** ND-normal,PS>SA	No group differences
Peak EMG	**VL, VM , RF** ND-normal>PS,SA **LH, GM** ND-normal,SA>PS	**VM, RF, GM, LH** OA>YA

A.4 CONCLUSIONS

In this preliminary study, both types of lifting-seat devices were compared with an unassisted rise at a normal seat height examining peak angles, moments, and muscle activity. Interestingly, not all key dependent measures were different between the two devices and normal seat height. The Seat Assist™ reduced peak knee flexion angle, peak hip and knee flexion moments, and peak quadriceps muscle activity compared to a normal seat height. The Power Seat™ reduced peak joint angles, peak hip moments, and peak activity of all muscles compared to a normal seat height.

APPENDIX B ABSTRACT: CANADIAN PHYSIOTHERAPY ASSOCIATION

This appendix contains a modified version of the abstract published for the Canadian Physiotherapy Association Congress, Montreal, Quebec, Canada. May 23-26, 2013.

Rutherford DJ, Hurley ST, Hubley-Kozey CL. Sit-to-stand transfer mechanics in healthy older adults: A comprehensive investigation of a portable lifting-seat device.

Relevance: Rising from a chair is a demanding motor task and an essential component for independent mobility. Physiotherapists prescribe lifting-seat devices, but objective evidence for their effectiveness is lacking.

Purpose: To evaluate lower extremity mechanics and muscle activation associated with the sit-to-stand transfer using a portable lifting-seat device and compare these data to an unassisted transfer in young and healthy older adults.

Materials & Methods: Using a cross-sectional, experimental design, bilateral lower extremity and low back musculature electromyography, three-dimensional leg and trunk motion and ground reaction forces were recorded from ten young (mean age =25) and ten healthy older (mean age =69) adults during 5 trials of i) no assist and ii) assisted transfers. Joint angles were derived and lower extremity moments of force were calculated using inverse dynamics. All data were time normalized to represent the period of seat-off to standing.

Analysis: Peak sagittal plane joint angles, moments of force and muscle activity were calculated. Analysis of variance models test for main effects and interactions ($\alpha = 0.05$).

Results: Trunk, hip and knee angles were significantly reduced and dorsiflexion increased with assisted transfer ($p<0.05$). Peak hip and ankle joint moments were reduced ($p<0.05$) and no change found in knee moments ($p>0.05$). Peak muscle activity was lower during the assisted transfer ($p<0.05$). Findings were similar between age groups.

Conclusions: In general, variables indicative of sit-to-stand functional demand were reduced with lifting-seat device use. Data provide a framework for future recommendations on product use by physiotherapists and research pertaining to the advancement of adaptive seating.

APPENDIX C DATA COLLECTION FORMS

C.1 Lower Extremity Functional Scale

LOWER EXTREMITY FUNCTIONAL SCALE

We are interested in knowing whether you are having any difficulty at all with the activities listed below because of your lower limb problem for which you are currently seeking attention. Please provide an answer for **each** activity.

Today, do you or would you have any difficulty at all with:

(Circle one number on each line)

ACTIVITIES	Extreme Difficulty or Unable to Perform Activity	Quite a bit of Difficulty	Moderate Difficulty	A Little bit of Difficulty	No Difficulty
a. Any of your usual work, housework or school activities.	0	1	2	3	4
b. Your usual hobbies, recreational or sporting activities.	0	1	2	3	4
c. Getting into or out of the bath.	0	1	2	3	4
d. Walking between rooms.	0	1	2	3	4
e. Putting on your shoes or socks.	0	1	2	3	4
f. Squatting.	0	1	2	3	4
g. Lifting an object, like a bag of groceries from the floor.	0	1	2	3	4
h. Performing light activities around your home.	0	1	2	3	4
I. Performing heavy activities around your home.	0	1	2	3	4
j. Getting into or out of a car.	0	1	2	3	4
k. Walking 2 blocks.	0	1	2	3	4
l. Walking a mile.	0	1	2	3	4
m. Going up or down 10 stairs (about 1 flight of stairs).	0	1	2	3	4
n. Standing for 1 hour.	0	1	2	3	4
o. Sitting for 1 hour.	0	1	2	3	4
p. Running on even ground.	0	1	2	3	4
q. Running on uneven ground.	0	1	2	3	4
r. Making sharp turns while running fast.	0	1	2	3	4
s. Hopping.	0	1	2	3	4
t. Rolling over in bed.	0	1	2	3	4
Column Totals:					

Score: ______ / 80

C.2 Data Collection Sheet

Data Collection Information

Subject Name:	DOB:	Height: Mass:	
Subject Extension (ZZZ):	R Thigh Cir (cm):	L Thigh Cir (cm):	
Kneenum:	R Calf Cir (cm):	L Calf Cir (cm):	
Date:	R Foot Width (cm):	L Foot Width (cm):	
Collectors:	R Knee Height (cm):	L Knee Height (cm):	
	Waist Circumf. (cm):	Hip Circumf. (cm):	

System Set up and Subject Calibration

Optotrak Fs: 50Hz	Siggi SWG 100uV P-P	RMS Cube: RMS Plate:
ODAU Fs: 500Hz	ODAU – Ch 1-6 FP 1	NI – Ch 0 = Torque /ES
EMG Fs: 2000Hz	ODAU – Ch 7-12 FP 2	NI – Ch 1-7 Left LE
Pt Unit Batteries < 8.8V	ODAU – Ch 13- Seat Switch	NI – Ch 9-15 Right LE

* **Note** : Channel 8 on BNC box is faulty (DO NOT USE) – Torque and ES on Channel 0

EMG Gains

		C1	C2	C3	C4	C5	C6	C7	C8
Amp 1	Left	TA (1)	MG (2)	VL (3)	VM (4)	RF (5)	LH (6)	Gmx (7)	ES (0)
Amp 2	Right	TA (9)	MG (10)	VL (11)	VM (12)	RF (13)	LH (14)	Gmx (15)	--------

Data Collection (Cal, VP and Sub Bias 1s, No Dev 5s, Pow_S 35s, Seat_A 10s, Maximums 3s)

Trial#	ID	Trial#	ID	Trial#	ID	Trial#	ID
001	Plate Calibration #1 (zero)	023	R Toe	045	S2S H seat	067	R PFsit max
002	Plate Calibration #2	024	R seated GT	046	Subject Bias	068	R DFsit max
003	Standing Calibration	025	L seated GT	047	L KE45 GC	069	R DFsit max
004	R Thoracic Spine	026	S2S No D	048	L KE45 max	070	L KF55P GC
005	L Thoracic Spine	027	S2S No D	049	L KE45 max	071	L KF55P max
006	R PSIS	028	S2S No D	050	L KEHF max	072	L KF55P max
007	L PSIS	029	S2S No D	051	L KEHF max	073	L Stand PF
008	R ASIS	030	S2S No D	052	L PFsit GC	074	L Stand PF
009	L ASIS		S2S D 1	053	L PFsit max	075	R Stand PF
010	L Medial Epicondyle		S2S D 1	054	L PFsit max	076	R Stand PF
011	L Fibular Head		S2S D 1	055	L DFsit max	077	R Glutmax
012	L Tibial Tuberosity		S2S D 1	056	L DFsit max	078	R Glutmax
013	L Medial Malleolus		S2S D 1	057	R KF55P GC	079	L Glutmax
014	L 2^{nd} Metatarsal Head		S2S D 2	058	R KF55P max	080	L Glutmax
015	L Heel		S2S D 2	059	R KF55P max	081	ES
016	L Toe (most distal)		S2S D 2	060	R KE45 GC	082	ES
017	R Medial Epicondyle		S2S D 2	061	R KE45 max	083	Sine wave
018	R Fibular Head		S2S D 2	062	R KE45 max	084	Grnd
019	R Tibial Tuberosity	041	S2S H seat	063	R KEHF max	085	Mass 18kg
020	R Medial Malleolus	042	S2S H seat	064	R KEHF max	086	No Mass
021	R 2^{nd} Metatarsal Head	043	S2S H seat	065	R PFsit GC	Moment Arm Length (cm)	
022	R Heel	044	S2S H seat	066	R PFsit max		

EMG set up

Left Leg (Amp 1, DOHM lead, DOHM PU)

DOHM	Sit to stand
LG	TA
MG	MG
VL	VL
VM	VM
RF	RF
LH	LH
MH	Glut max
GM	Erector spinea L33

Right Leg (Amp 2, Spare lead, SparePU)

DOHM	Sit to Stand
LG	TA
MG	MG
VL	VL
VM	VM
RF	RF
LH	LH
MH	Glut max

Electrode Placements

TA – 1/3 of the distance between fibular head and medial malleolus

MG – 35% of the distance between medial knee joint line and calcaneal tubercle

VL – 25% of the distance between lateral knee joint line and ASIS

VM – 20% of the distance between medial knee joint line and ASIS

RF –50% of the distance between patellar base and ASIS

LH –50% of the distance between fibular head and ischial tuberosity

Glut max – 50% of the distance between S2 and greater trochanter. Corresponds to the greatest prominence of the middle of the buttocks.

L33 – 3cm lateral to L3 (Left)

Motion Capture Set up

Left Leg (First string of leads – Marker set STS 1)

	DOHM	Sit to Stand
Mkr 1	SH	LM
Mkr 2	Fin1	Foot1
Mkr 3	Fin2	Foot2
Mkr 4	Fin3	Foot3
Mkr 5	GT `	GT
Mkr 6	Thigh1	Tibia1
Mkr 7	Thigh2	Tibia2
Mkr 8	Thigh3	Tibia3
Mkr 9	LE	LE
Mkr 10	Tibia1	Thigh1
Mkr 11	Tibia2	Thigh2
Mkr 12	Tibia3	Thigh3
Mkr 13	LM	SH
Mkr 14	Foot1	Tsp1
Mkr 15	Foot2	Tsp2
Mkr 16	Foot3	Tsp3

Right Leg (Second string of leads – Marker set STS 2)

	DOHM	Sit to Stand
Mkr 17	SH	LM
Mkr 18	Fin1	Foot1
Mkr 19	Fin2	Foot2
Mkr 20	Fin3	Foot3
Mkr 21	GT `	GT
Mkr 22	Thigh1	Tibia1
Mkr 23	Thigh2	Tibia2
Mkr 24	Thigh3	Tibia3
Mkr 25	LE	LE
Mkr 26	Tibia1	Thigh1
Mkr 27	Tibia2	Thigh2
Mkr 28	Tibia3	Thigh3
Mkr 29	LM	SH
Mkr 30	Foot1	Pelvis1
Mkr 31	Foot2	Pelvis 2
Mkr 32	Foot3	Pelvis 3

Seat Markers (third string of leads – Marker set STS 3)

Mkr 33,34,35	Digitizer/Seat triad
Mkr 36	Digitizer/main axis of rotation
Mkr 37	Digitizer/Seat pan axis of rotation
Mkr 38	Digitizer/Back of seat

APPENDIX D STANDING EVENT METHOD DETERMINATION

D.1 INTRODUCTION

Sit-to-stand (STS) events are used to separate a STS transfer task into different phases of interest. The phase most commonly analyzed is from seat-off to standing, which is the period where the greatest lower limb muscle activity and joint moments occur as the center of mass moves from a stable three-point sitting position to a more unstable two-point standing position (Pai, Rogers 1991, Doorenbosch et al. 1994, Jeyasurya 2011, Goulart, Valls-sole 1999). As a result, determining appropriate seat-off and standing events is required prior to analyzing biomechanical dependent variables needed to address study objectives.

Seat switches used to determine when buttocks lose contact with the seat are easy to employ and widely used to determine seat-off in STS studies (Munro et al. 1998, Goulart, Valls-sole 1999, Turcot et al. 2012, Christiansen, Stevens-Lapsley 2010, Papa, Cappozzo 2000). However, standing event determination is much more variable. Standing event methods include; i) when knee extension angular velocity equals zero (Epifanio et al. 2008), ii) when hip extension angular velocity equals zeroc(Akram, McIlroy 2012, Schenkman, Rlley & Mann 1990, Ikeda et al. 1991), iii) when hip marker linear horizontal velocity or center of mass momentum equals zero (Khemlani, Carr & Crosbie 1999, Pai, Rogers 1991, Munro et al. 1998, Munro, Steele 2000, Shepherd, Koh 1996), iv) when the vertical velocity of a kinematic marker or segment equals zero (Kuo, Tully & Galea 2009, Turcot et al. 2012), v) when the vertical ground reaction force rate of change equals zero (Etnyre, Thomas 2007, Jeyasurya 2011). Of the proposed standing event methods, it is important to use one that consistently and accurately identifies standing for all study conditions and groups, particularly when calculating dependent variables that account for duration, such as integrated EMG and moment impulse.

The purpose of this appendix is to determine which standing event method(s) is the most appropriate for describing standing for four different seating conditions, unassisted STS at normal and raised seat heights, assisted STS using two different types of lifting-seat devices (the Power Seat™ and the Seat Assist™) for both healthy younger and healthy older adults. The ideal standing event method would 1) be suitable for all conditions and groups, 2) work for all trials, and 3) accurately select a time of standing. In order to determine the accuracy, key biomechanical measures at the instant of each standing event will be measured. They will be compared with the timing and key biomechanical measures at the time of the gold standard standing event, which is the time when the knee and hip are extended and the shoulder marker has reached its peak vertical position. We hypothesize that all standing event methods and the gold standard standing event will be similar in timing and key biomechanical variables.

D.2 METHODOLOGY

Study methodology has been previously described in Chapter 3 with respect to study participants, data collection, and data processing. Additional information on data processing and analysis unique to this sub-objective study is presented below. All data processing was completed using custom programs written in MATLAB version 7.4 (Mathworks, Natick, MA, USA)

D.2.1 Gold Standard Standing Event

The gold standard standing event (GOLD) was determined when all three conditions were met; 1) the hip flexion angle (HA) reached 5% of the angular excursion from the minimum flexion angle (Equation D.1), 2) the knee flexion angle (KA) reached 5% of the angular excursion from the minimum flexion angle (Equation D.2), and 3) the shoulder vertical position (SH) reached 95% of the positional excursion from the minimum position (Equation D.3).

[D.1] $Hip\ Standing = \min(HA) + 0.05(\max(HA) - min(HA))$

[D.2] $Knee\ Standing = \min(KA) + 0.05(\max(KA) - min(KA))$

[D.3] $Shoulder\ Standing = \min(SH) + 0.95(\max(SH) - min(SH))$

[D.4] $GOLD = \max(time(Hip\ Standing), time(Knee\ Standing), time\ (Shoulder\ Standing))$

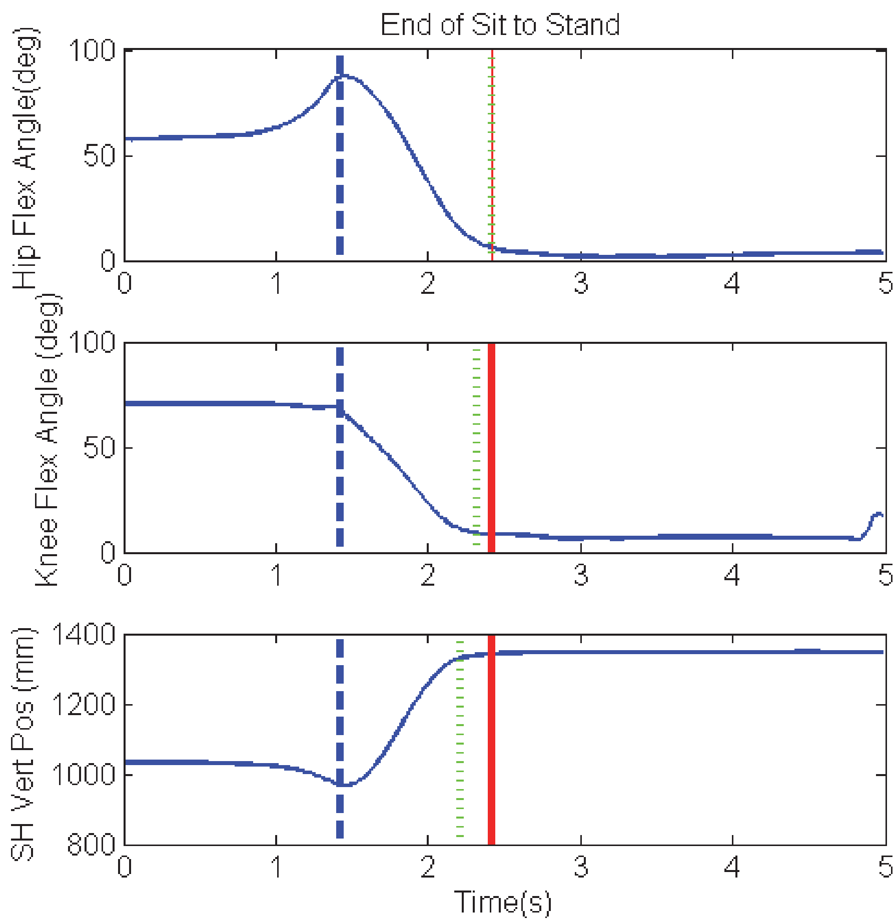


Figure D.1. Example of gold standing determination. First dashed line indicates time of seat-off. Solid line indicates time of gold standing when all three standing events (second dashed lines) have been met.

D.2.2 Standing Event methods

Five methods were used to determine standing event. 1) when the thoracic Spine vertical velocity equals zero (TSP), 2) when the hip extension angular velocity equals zero (HipExt), 3) when the knee extension angular velocity equals zero (KneeExt), 4) vertical ground reaction force rate of change equals zero after maximum and local minimum (vGRF), 5) when vertical velocity of the shoulder marker equals zero (SH). Velocities and rates of changes were calculated using finite differences (Equation D.5).

[D.5] $Velocity(n) = \frac{[position(n+1) - position(n-1)]*(sampling\ rate)}{2}$

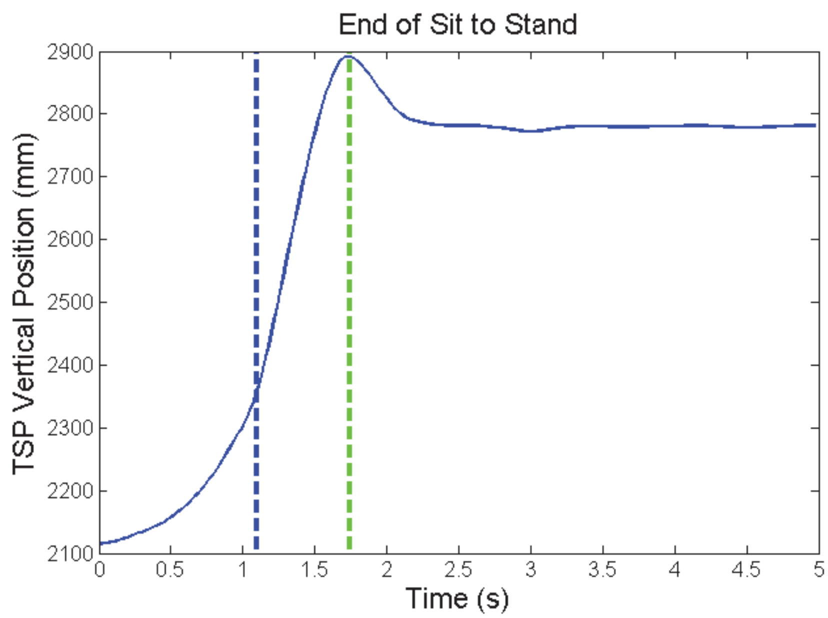


Figure D.2. Example of thoracic spine (TSP) standing determination. First dashed line indicates time of seat-off and second dashed line indicates time when TSP vertical velocity = 0.

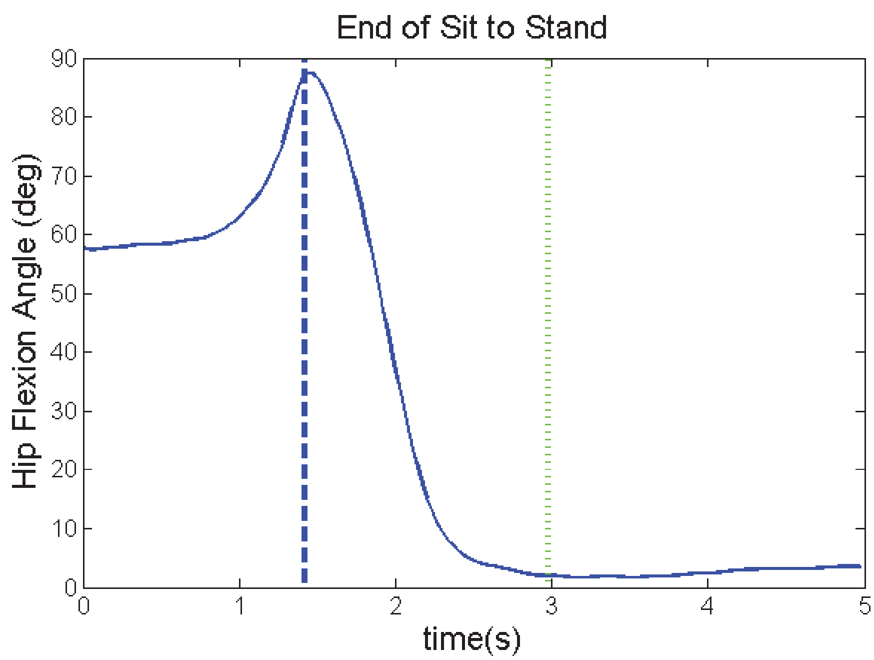


Figure D.3. Example of hip extension (HipExt) standing determination. First dashed line indicates time of seat-off and second dashed line indicates time when hip extension angular velocity = 0.

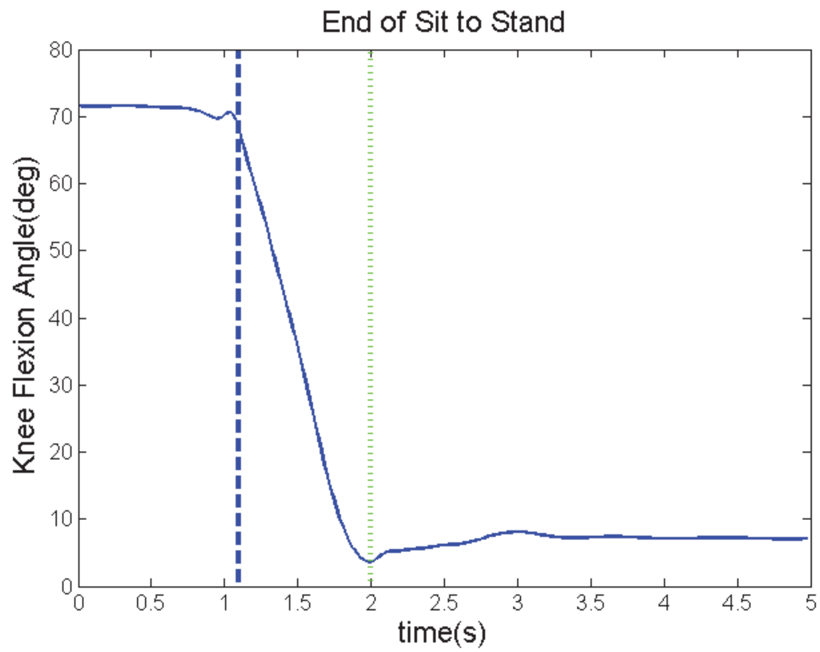


Figure D.4.Example of knee extension (KneeExt) standing determination. First dashed line indicates time of seat-off and second dashed line indicates time when knee extension angular velocity = 0.

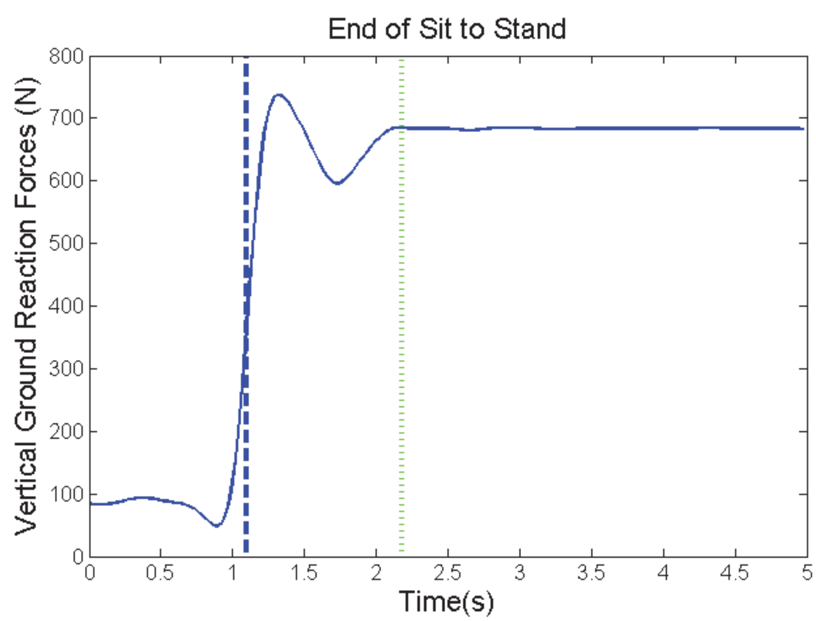


Figure D.5.Example of vertical ground reaction (vGRF) standing determination. First dashed line indicates time of seat-off and second dashed line indicates time when vertical ground reaction force rate of change = 0 after maximum and local minimum.

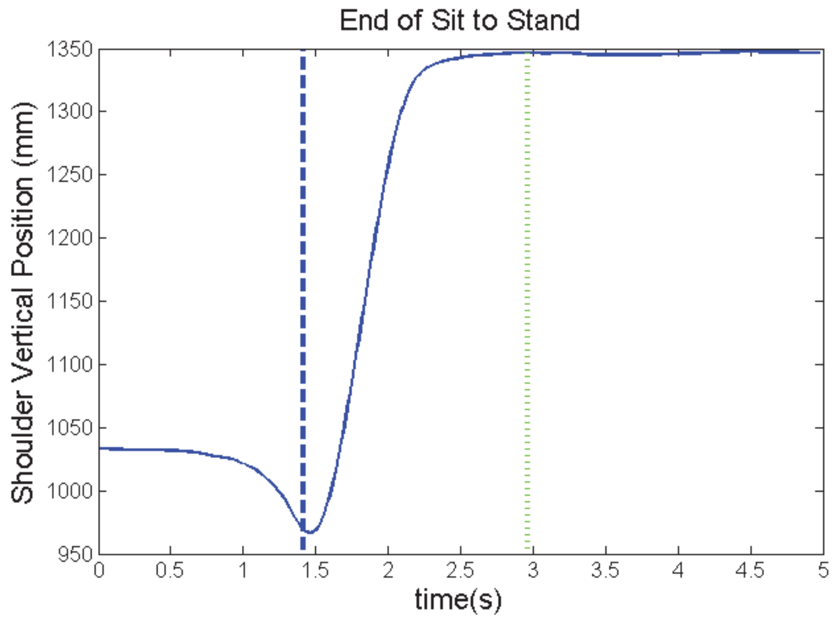


Figure D.6 Example of shoulder (SH) standing determination. First dashed line indicates time of seat-off and second dashed line indicates time when shoulder marker vertical velocity = 0.

D.2.3 Dependent Variables

If the event was unable to be accurately determined based on graphical observation, the trial was recorded as an error trial and removed from the analysis. For each method, means and standard deviations for dependent variables (Table D.1) for each group and seating condition were calculated at the time of standing event. Descriptive statistics of dependent variables will be presented. No statistical analysis was performed.

Table D. 1. Dependent variables at the time of standing events.

Category	**Variables**
Time	Seat-off to Standing
Sagittal Plane Angles	(Hip, Knee)
Sagittal Plane Moments	(Hip, Knee)
Vertical position	Shoulder marker

D.3 RESULTS

D.3.1 Event Order

Hip, knee, and shoulder event order when determining Gold standing event is presented in table D.2. The most common order of events for all seating conditions and both groups is shoulder reaching highest vertical position first, knee reaching full extension second, and hip reaching full extension third. Second most common is shoulder first, then hip second, and knee third.

D.3.2 Event Errors

Standing event error trials that were removed from the analysis are presented in Table D.3. 31 trials were removed for older adults and 10 trials were removed for the younger adults. The most trials were removed for the Power Seat™ (17) and the least were removed for the Seat Assist™ (5). For the standing event methods, 19 trials were removed for the vertical ground reaction force trials, 12 of which were PS trials. No trials were removed using the thoracic spine or shoulder standing methods.

Table D.2. Hip (H), knee (K), and shoulder (S) event order when determining Gold standing event. Example: H>K>S represents the shoulder reaching highest vertical position first, knee reaching full extension second, and hip reaching full extension third.

Older	**H>K>S**	**K>H>S**	**H>S>K**	**K>S>H**	**S>H>K**	**S>K>H**	**H=K=S**	**H=K>S**	**S>H=K**	**K=S>H**	**H>K=S**	**H=S>K**	**K>H=S**
ND-normal	20	9	2	0	0	0	0	2	0	0	1	0	0
Power Seat	19	12	2	0	2	0	0	4	0	0	0	0	0
Seat Assist	25	8	3	0	0	0	0	1	1	0	0	0	0
ND-raised	18	11	2	0	2	0	0	1	0	0	0	0	0
Younger	**H>K>S**	**K>H>S**	**H>S>K**	**K>S>H**	**S>H>K**	**S>K>H**	**H=K=S**	**H=K>S**	**S>H=K**	**K=S>H**	**H>K=S**	**H=S>K**	**K>H=S**
ND-normal	24	12	0	0	0	0	0	6	0	0	0	0	0
Power Seat	23	11	0	0	0	1	0	3	0	0	1	0	0
Seat Assist	20	17	0	0	0	0	0	3	0	0	0	0	0
ND-raised	23	14	0	0	0	0	0	5	0	0	0	0	0

Table D.3. Standing event error trials that were removed from the analysis. 400 total trials.

Older	**GOLD**	**Thoracic Spine**	**Knee Extension**	**Hip Extension**	**vGRF**	**Shoulder**
ND-normal	3	0	5	0	2	0
Power Seat	2	0	0	1	9	0
Seat Assist	1	0	1	0	3	0
ND-raised	1	0	1	0	2	0
Younger	**GOLD**	**Thoracic Spine**	**Knee Extension**	**Hip Extension**	**vGRF**	**Shoulder**
ND-normal	0	0	0	1	0	0
Power Seat	0	0	0	2	3	0
Seat Assist	2	0	0	1	0	0
ND-raised	0	0	0	1	0	0

Table D.4. Differences between dependent variables at the time of the gold standard event and at the time of different standing event methods for

		Thoracic Spine		Knee Extension		Hip Extension		vGRF		Shoulder	
Older Adults		**Mean**	Std	**Mean**	Std	**Mean**	Std	**Mean**	Std	**Mean**	Std
ND-normal	**Time(s)**	**-0.43**	0.07	**0.16**	0.11	**-0.36**	0.14	**0.05**	0.10	**-0.05**	0.15
	Hip Angle(°)	**24.80**	3.33	**-1.34**	1.21	**3.14**	0.85	**-0.40**	1.55	**3.36**	4.77
	Knee Angle(°)	**19.50**	6.78	**-1.54**	0.68	**1.37**	1.18	**0.19**	1.03	**2.26**	2.61
	Hip Moment(Nm/kg)	**0.20**	0.06	**0.00**	0.02	**0.03**	0.02	**0.00**	0.02	**0.02**	0.04
	Knee Moment(Nm/kg)	**0.15**	0.07	**-0.03**	0.04	**0.03**	0.04	**0.00**	0.02	**0.04**	0.05
	Shoulder position(mm)	**-31.62**	9.93	**-0.91**	0.88	**1.96**	2.12	**-0.83**	2.15	**3.04**	2.26
Power Seat	**Time(s)**	**-0.62**	0.13	**-0.01**	0.16	**-0.27**	0.22	**-0.17**	0.21	**-0.13**	0.20
	Hip Angle(°)	**21.15**	2.57	**-0.48**	1.45	**2.06**	0.82	**2.63**	3.92	**2.45**	2.90
	Knee Angle(°)	**16.00**	3.73	**-0.94**	0.53	**0.96**	0.63	**2.10**	3.31	**1.87**	1.76
	Hip Moment(Nm/kg)	**0.18**	0.04	**0.00**	0.02	**0.02**	0.01	**0.02**	0.04	**0.02**	0.03
	Knee Moment(Nm/kg)	**0.15**	0.08	**-0.01**	0.01	**0.01**	0.03	**0.02**	0.03	**0.02**	0.03
	Shoulder position(mm)	**-26.99**	9.74	**-1.11**	2.04	**0.66**	1.28	**-4.51**	6.58	**1.57**	0.59
Seat Assist	**Time(s)**	**-0.52**	0.11	**0.17**	0.10	**-0.43**	0.21	**0.04**	0.31	**-0.06**	0.16
	Hip Angle(°)	**25.59**	3.32	**-1.89**	1.25	**3.47**	1.00	**2.21**	4.54	**2.89**	3.64
	Knee Angle(°)	**16.40**	3.74	**-1.44**	0.67	**1.22**	1.26	**2.10**	3.02	**1.93**	1.94
	Hip Moment(Nm/kg)	**0.21**	0.06	**-0.01**	0.02	**0.04**	0.02	**0.01**	0.03	**0.02**	0.03
	Knee Moment(Nm/kg)	**0.15**	0.08	**-0.02**	0.03	**0.00**	0.04	**0.03**	0.04	**0.02**	0.02
	Shoulder position(mm)	**-37.52**	12.01	**-1.12**	1.26	**1.30**	3.16	**-3.25**	6.31	**2.38**	2.21
ND-raised	**Time(s)**	**-0.61**	0.13	**-0.05**	0.28	**-0.24**	0.24	**-0.12**	0.13	**-0.16**	0.21
	Hip Angle(°)	**23.76**	5.43	**0.13**	1.66	**2.30**	0.80	**1.16**	2.04	**2.48**	3.85
	Knee Angle(°)	**16.02**	4.49	**-0.85**	0.83	**0.52**	1.22	**1.36**	1.16	**1.33**	2.01
	Hip Moment(Nm/kg)	**0.19**	0.05	**0.01**	0.03	**0.02**	0.01	**0.01**	0.02	**0.02**	0.03
	Knee Moment(Nm/kg)	**0.16**	0.07	**-0.01**	0.03	**0.01**	0.02	**0.01**	0.03	**0.02**	0.03
	Shoulder position(mm)	**-32.54**	13.24	**-0.72**	2.10	**1.11**	1.03	**-2.18**	3.11	**1.60**	2.22

Negative values indicate a greater value for the Gold Standard Event.

Table D.5. Differences between dependent variables at the time of the gold standard event and at the time of different standing event methods in younger adults

Younger Adults		**Thoracic Spine** Mean	Std	**Knee Extension** Mean	Std	**Hip Extension** Mean	Std	**vGRF** Mean	Std	**Shoulder** Mean	Std
ND-normal	**Time(s)**	**-0.41**	0.12	**0.21**	0.08	**0.35**	0.11	**0.05**	0.10	**-0.12**	0.13
	Hip Angle(°)	**20.05**	5.49	**-2.10**	0.80	**-3.24**	0.97	**-0.99**	1.73	**3.61**	3.28
	Knee Angle(°)	**15.97**	3.56	**-2.04**	0.72	**-2.27**	0.70	**-0.65**	1.11	**2.73**	2.45
	Hip Moment(Nm/kg)	**0.15**	0.06	**-0.01**	0.02	**-0.03**	0.02	**-0.01**	0.02	**0.02**	0.03
	Knee Moment(Nm/kg)	**0.18**	0.07	**-0.03**	0.02	**-0.03**	0.03	**0.00**	0.02	**0.04**	0.04
	Shoulder position(mm)	**20.41**	7.99	**-1.95**	1.82	**-2.90**	1.96	**-2.31**	2.18	**2.47**	2.51
Power Seat	**Time(s)**	**-0.49**	0.11	**0.19**	0.15	**0.40**	0.13	**-0.01**	0.10	**-0.17**	0.12
	Hip Angle(°)	**17.83**	4.14	**-1.00**	1.37	**-2.26**	0.63	**-0.03**	1.24	**3.82**	2.11
	Knee Angle(°)	**15.53**	4.03	**-1.38**	1.11	**-1.82**	0.76	**0.20**	1.14	**3.00**	2.12
	Hip Moment(Nm/kg)	**0.15**	0.05	**-0.01**	0.02	**-0.02**	0.02	**-0.01**	0.02	**0.03**	0.02
	Knee Moment(Nm/kg)	**0.19**	0.08	**-0.01**	0.02	**-0.01**	0.02	**0.01**	0.02	**0.05**	0.04
	Shoulder position(mm)	**16.93**	8.02	**-0.90**	0.99	**-1.37**	1.52	**-1.09**	1.09	**1.54**	1.42
Seat Assist	**Time(s)**	**-0.48**	0.19	**0.17**	0.09	**0.35**	0.08	**0.08**	0.21	**-0.19**	0.16
	Hip Angle(°)	**20.02**	5.69	**-1.65**	0.86	**-2.76**	1.00	**-0.62**	2.53	**3.99**	2.08
	Knee Angle(°)	**16.34**	4.80	**-1.68**	0.60	**-2.02**	0.61	**-0.08**	2.17	**3.10**	2.54
	Hip Moment(Nm/kg)	**0.17**	0.07	**-0.01**	0.01	**-0.03**	0.02	**-0.01**	0.03	**0.03**	0.02
	Knee Moment(Nm/kg)	**0.18**	0.07	**-0.02**	0.01	**-0.02**	0.02	**0.01**	0.03	**0.04**	0.04
	Shoulder position(mm)	**22.16**	12.35	**-1.67**	0.94	**-2.48**	2.01	**-2.35**	2.36	**2.06**	1.08
ND-raised	**Time(s)**	**-0.41**	0.15	**0.16**	0.12	**0.28**	0.10	**0.07**	0.14	**-0.13**	0.12
	Hip Angle(°)	**17.92**	6.51	**-1.45**	1.10	**-2.36**	0.76	**-0.89**	1.73	**2.92**	1.89
	Knee Angle(°)	**16.06**	4.25	**-1.69**	0.89	**-1.95**	0.59	**-0.71**	1.76	**2.48**	2.04
	Hip Moment(Nm/kg)	**0.16**	0.08	**-0.01**	0.02	**-0.03**	0.02	**-0.02**	0.02	**0.03**	0.02
	Knee Moment(Nm/kg)	**0.18**	0.06	**-0.01**	0.02	**-0.01**	0.01	**0.00**	0.02	**0.03**	0.03
	Shoulder position(mm)	**20.40**	13.20	**-1.37**	1.13	**-2.22**	1.23	**-1.76**	0.97	**1.65**	1.21

Negative values indicate a greater value for the Gold Standard Event.

D.3.3 Dependent Variables

All standing event methods with the exception of the TSP method had similar angles (<4°), moments (<0.05Nm/kg) and shoulder position (<4.5mm) compared to the gold standard method in both younger and older adults. The TSP method determined a standing time that was earlier (0.41-0.62 seconds), with the knee in 15.5-19.5° more flexion, the hip in 17.8-25.6° more flexion, the shoulder was 16.9-37.3mm lower.

D.4 DISCUSSION

The purpose of this appendix was to determine which standing event method(s) is the preferred method for describing standing for four different seating conditions (ND-normal, SA, PS, ND-raised) and for both healthy younger and healthy older adults. The shoulder marker method was selected as the best standing event method based on our criteria; 1) suitable for all conditions and both groups, 2) works for all trials, 3) denotes a accurate time of standing.

In order to determine the accuracy of the standing event, timing and key biomechanical measures at the instant of each standing event were compared with those at the time of the gold standard standing event. The gold standard standing event was a novel approach that selected the time when the knee and hip were fully extended and a superior anatomical location, represented by the shoulder marker in its maximum vertical position. The sequence varied between trials with the most common being the shoulder reaching its maximum vertical position first, then the knee reaching full extension, and lastly, the hip reaching full extension. The sequences of events were similar for older and younger adults as well as between conditions. There was some variation in event order, which suggests a combination of events might be preferable compared to singular event methods for accurately determining standing. However, there were nine trials where the gold standard method did not work properly, which forced the removal of these trials. Nine trials out of four hundred is a small percentage but ideally, the standing event method would work for every trial. The question remaining; is there a method that works for every STS trial and selects a time point that is similar to the gold standing event?

Five commonly used standing event methods were selected and compared in this study; 1) when the thoracic spine vertical velocity equals zero, 2) when the hip extension angular velocity equals zero (Akram, McIlroy 2012, Schenkman, Rlley & Mann 1990, Ikeda et al. 1991), 3) when the knee extension angular velocity equals zero (Epifanio et al. 2008), 4) vertical ground reaction force rate of change equals zero after maximum and local minimum (Etnyre, Thomas 2007, Jeyasurya 2011), and 5) when vertical velocity of kinematic marker, such as the shoulder marker equals zero (Kuo, Tully & Galea 2009, Turcot et al. 2012).

The TSP method determined a standing time that was earlier (0.41-0.62 seconds) and as a result, the knee was in 15.5-19.5° more flexion, the hip was in 17.8-25.6° more flexion, the shoulder was 16.9-37.3mm lower compared to the gold standard event. Based on these findings the TSP method was considered a poor standing event method. All other methods had similar angles (<4°), moments (<0.05N•m/kg) and shoulder position differences (<4.5mm) compared to the gold standard method. Therefore, the knee extension method, the hip extension method, the vertical ground reaction force method, and the shoulder method were all accurately selected the standing event. As previously mentioned, the ideal standing event method would not only be accurate but also be suitable for all conditions and groups and work for all trials. The shoulder method was the only one of these methods that did not have any error trials, and thus, is the preferred standing event method based on the findings of this study. The vertical ground reaction method was particularly susceptible to error trials with 19 trials (12 PS trials), which refutes the recommendations by Etnyre et al. (2007) of using vertical ground reaction force STS events. Their study compared STS with varying arm positions. Therefore, different methods for STS event selection may be required for different research questions, such as the effect of seat height or lifting seat devices.

D.5 Summary

Several different methods accurately selected the standing event. Of these, the shoulder vertical velocity method was the only method absent of any error trials, making it the preferred standing event method for the different seating conditions used in this study involving healthy older and younger adults.

APPENDIX E CHAPTER 4: ENSEMBLE AVERAGE WAVEFORMS AND INTERACTION PLOTS

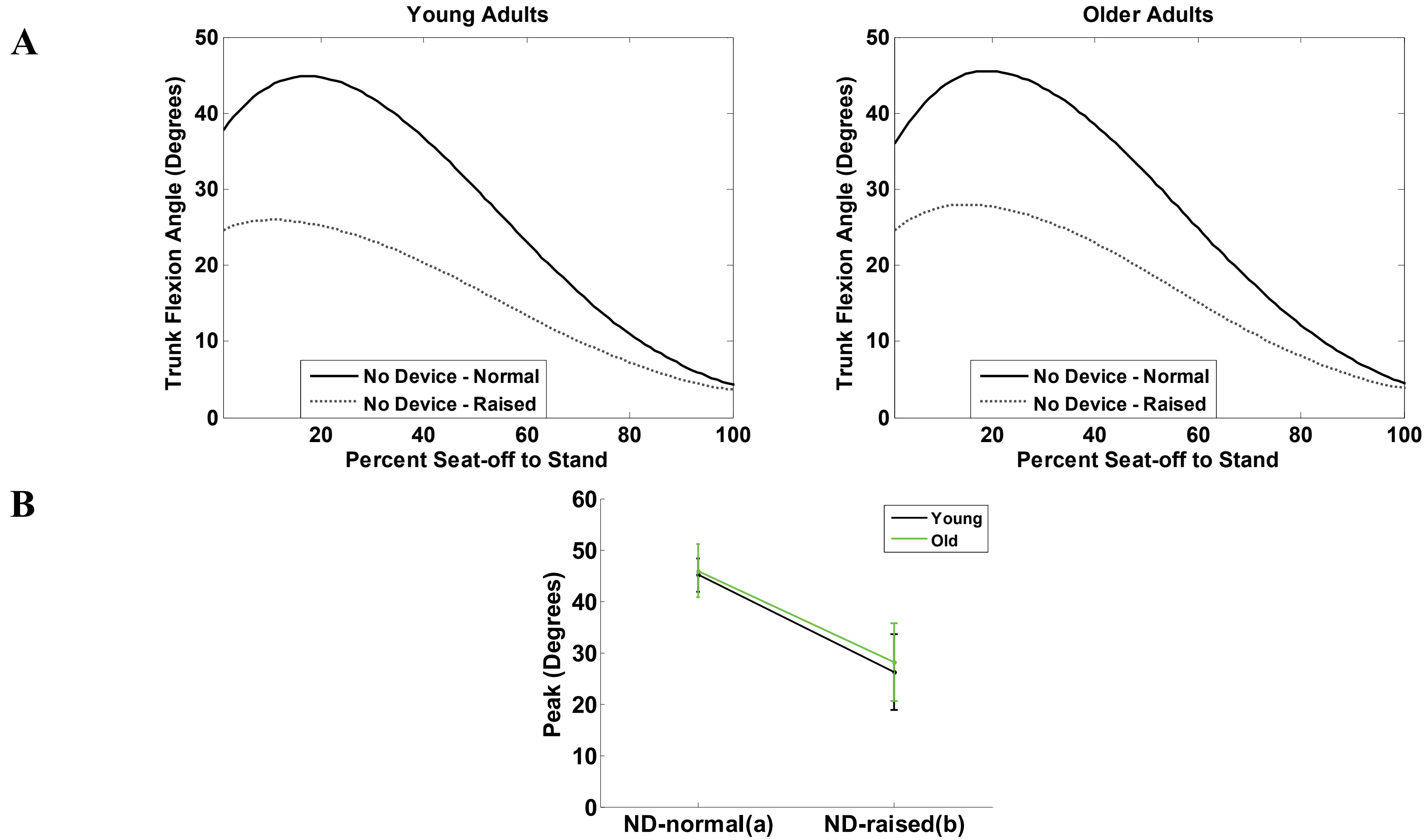


Figure E.1 Ensemble average trunk flexion angle waveforms for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

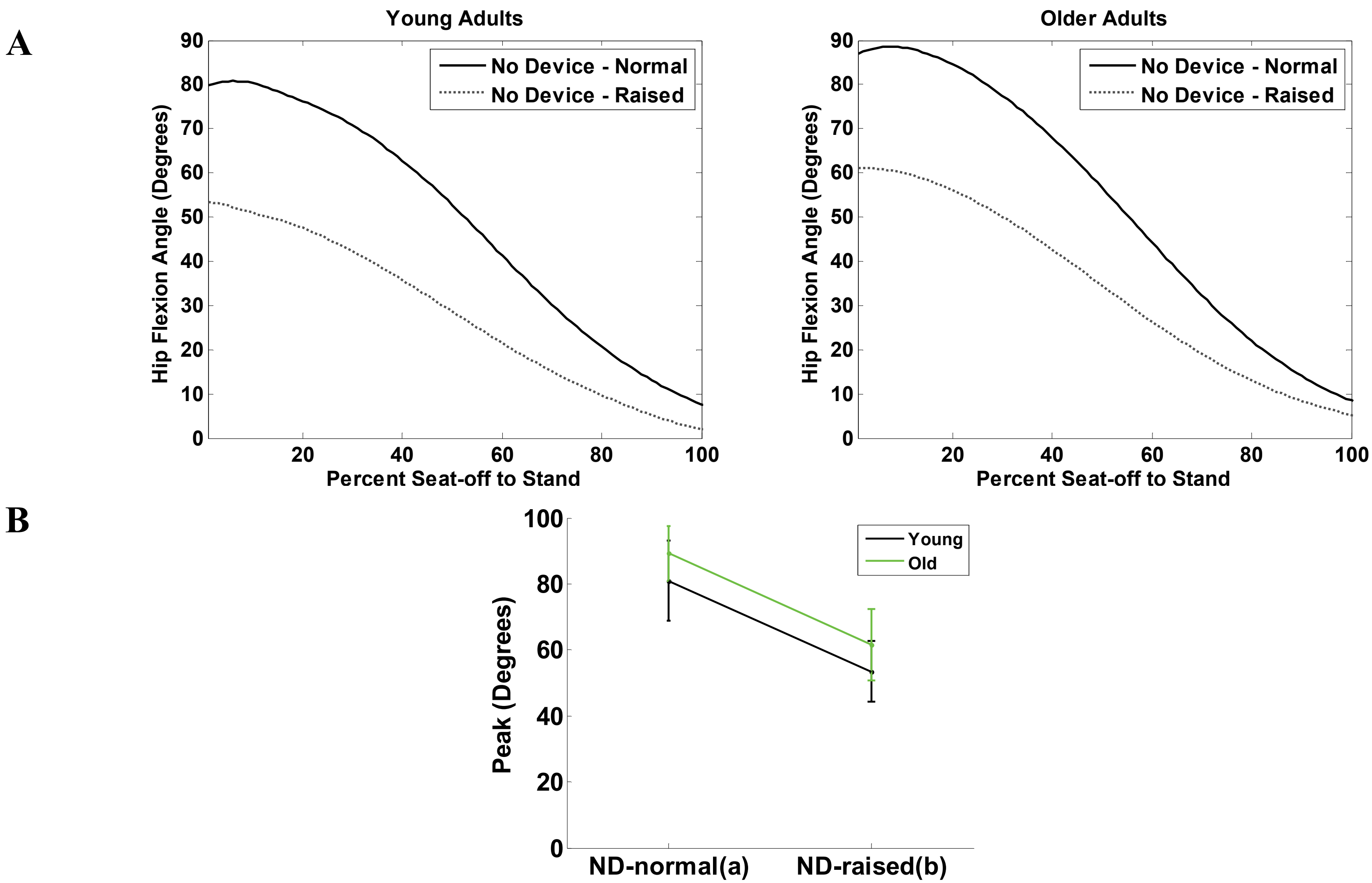


Figure E.2. Ensemble average hip flexion angle waveforms for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

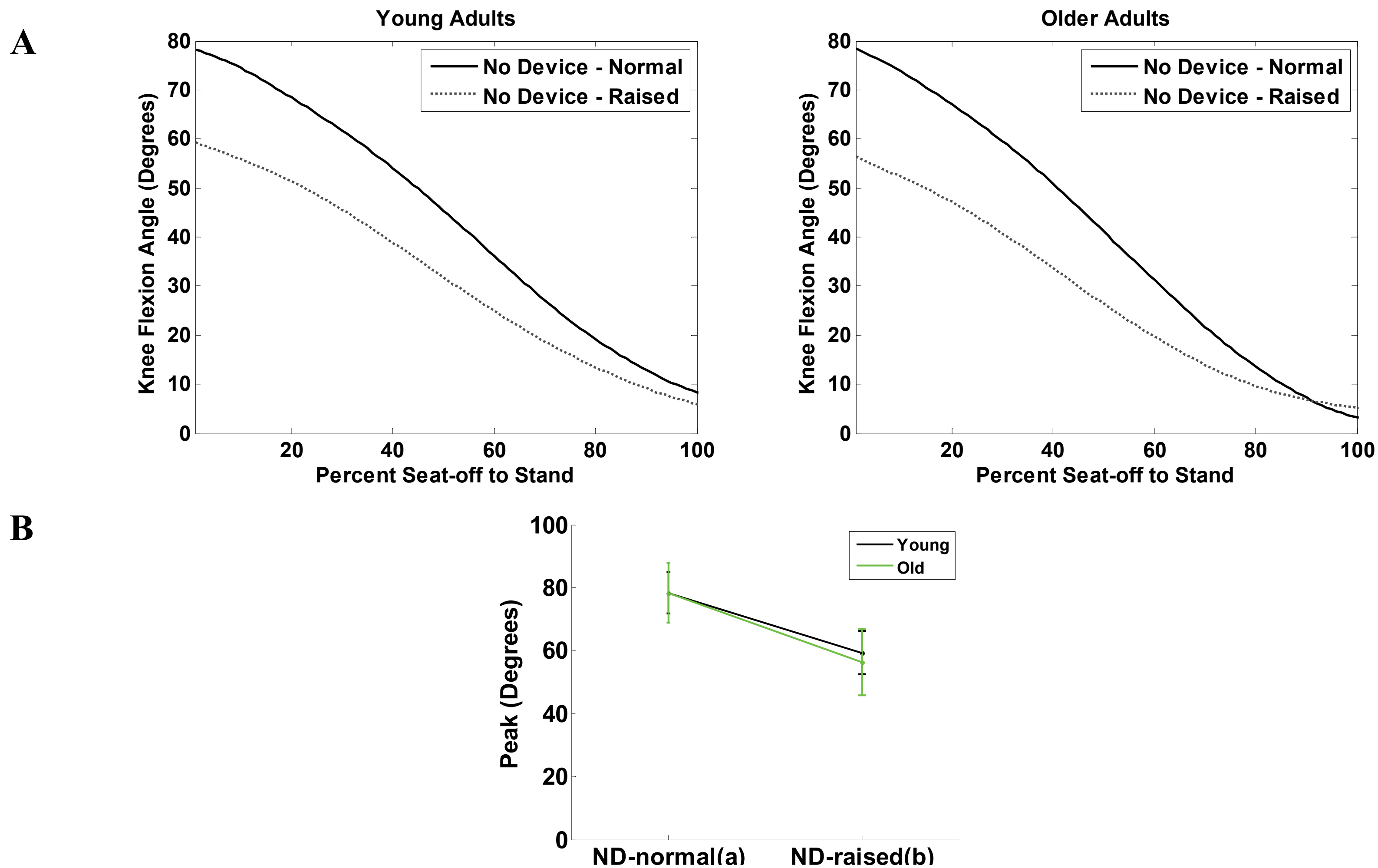


Figure E.3. Ensemble average knee flexion angle waveforms for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B). * indicates a significant main group effect (p<0.05) and different letters on the x-axis indicate a significant condition main effect (p<0.05).

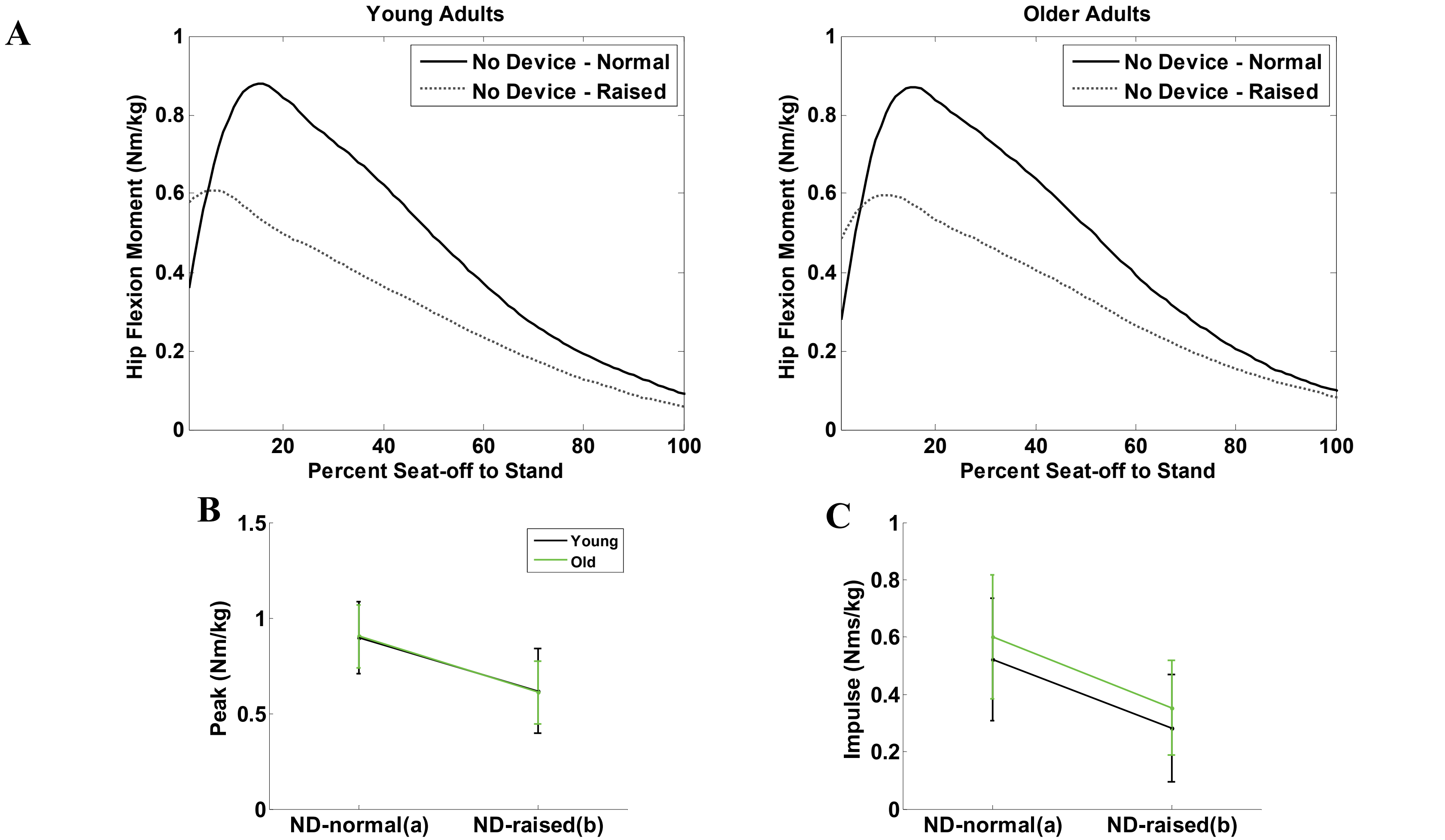


Figure E.4. Ensemble average hip flexion moment waveforms for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B), and impulse values (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

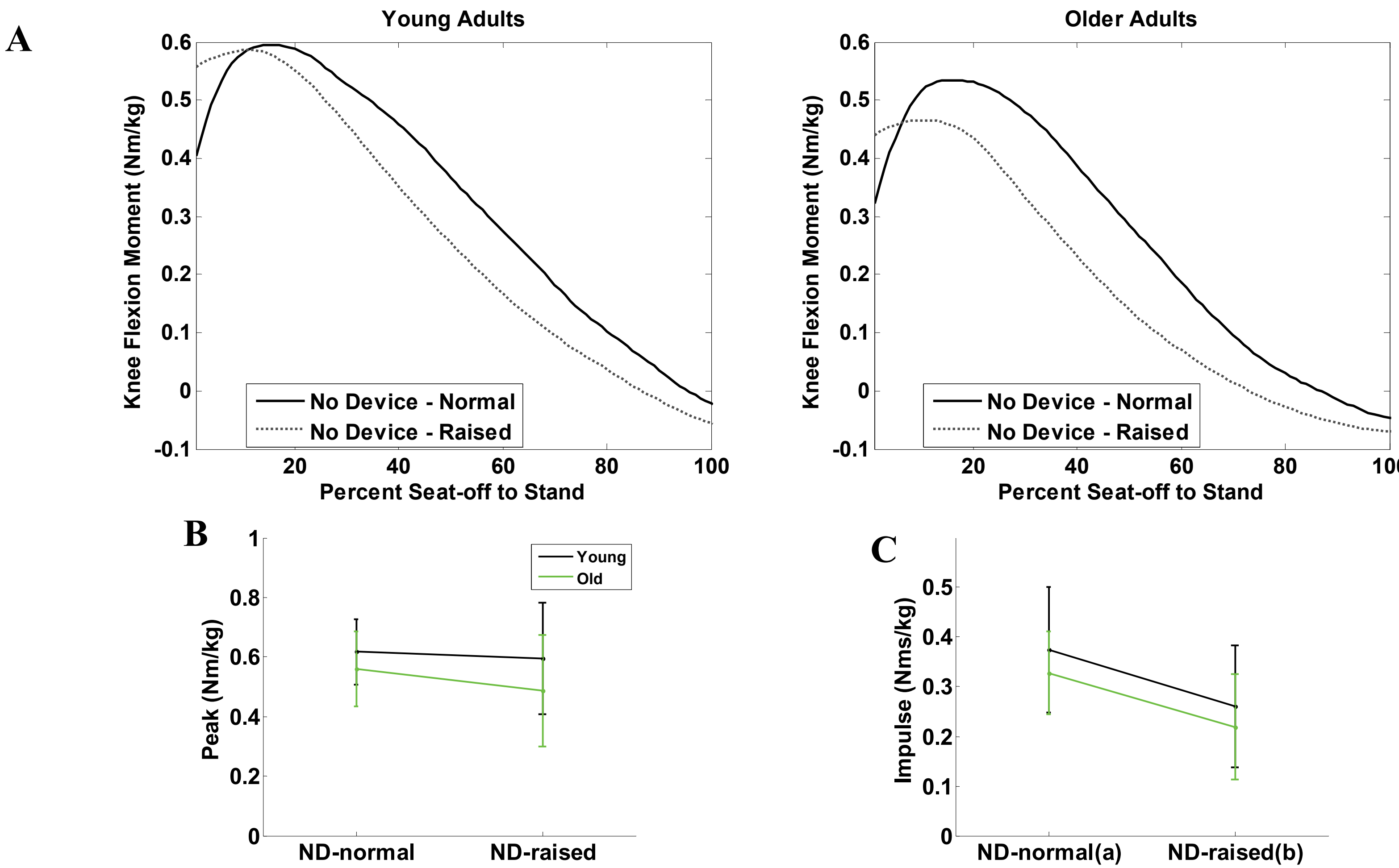


Figure E.5. Ensemble average knee flexion moment waveforms for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and impulse values (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

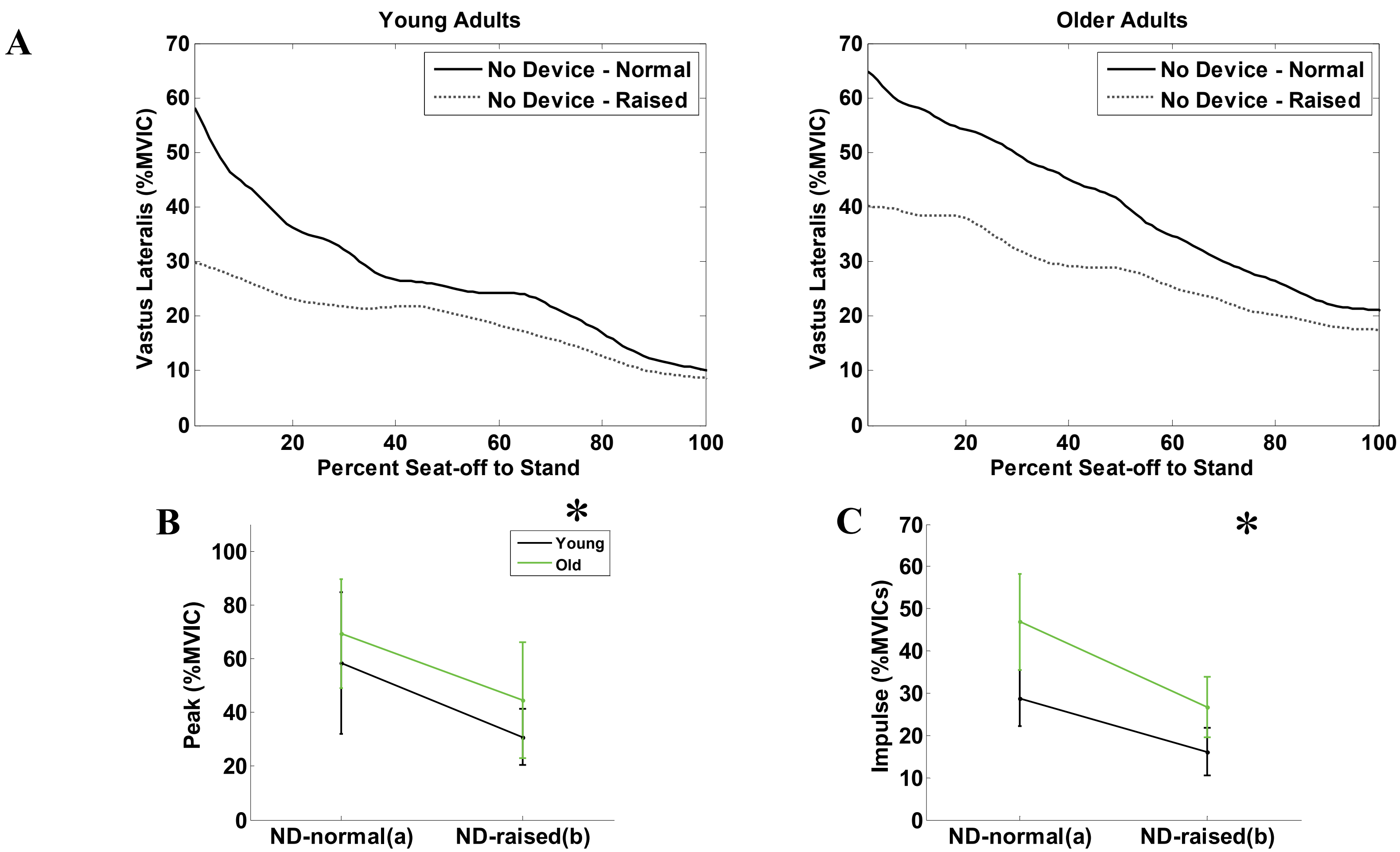


Figure E.6. Ensemble average EMG activity waveforms of the vastus lateralis for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

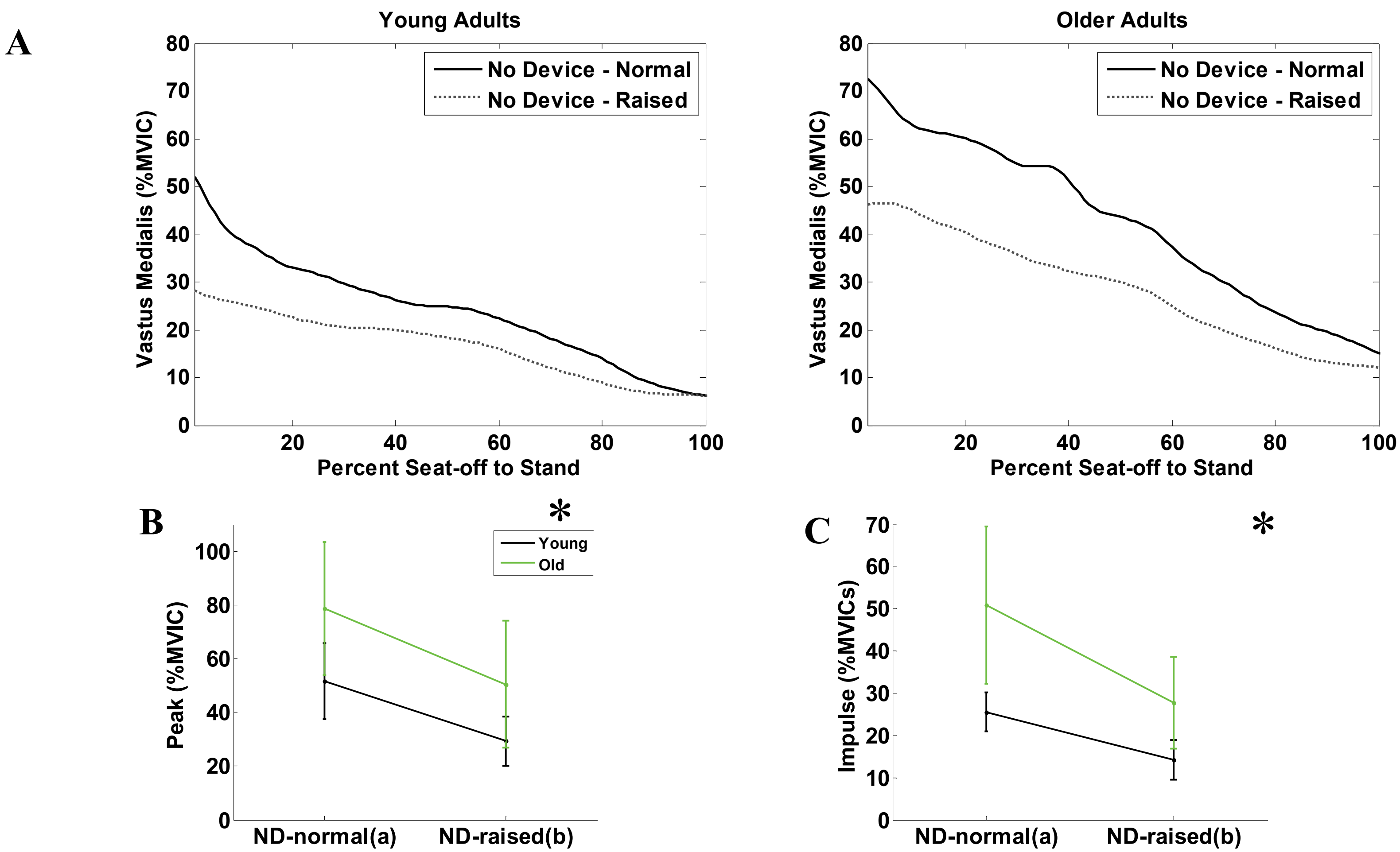


Figure E.7. Ensemble average EMG activity waveforms of the vastus medialis for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

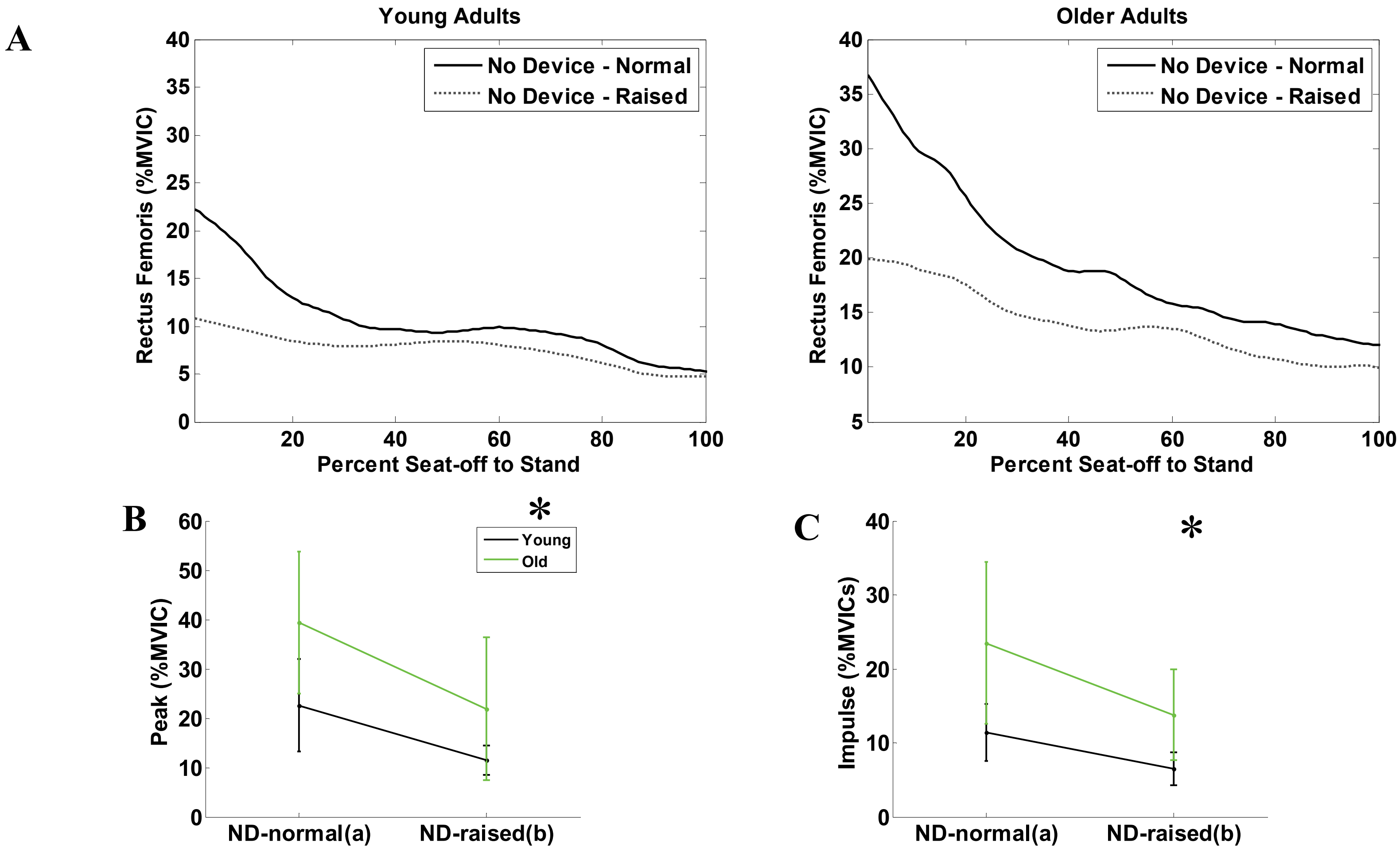


Figure E.8. Ensemble average EMG activity waveforms of the rectus femoris for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect (p<0.05) and different letters on the x-axis indicate a significant condition main effect (p<0.05).

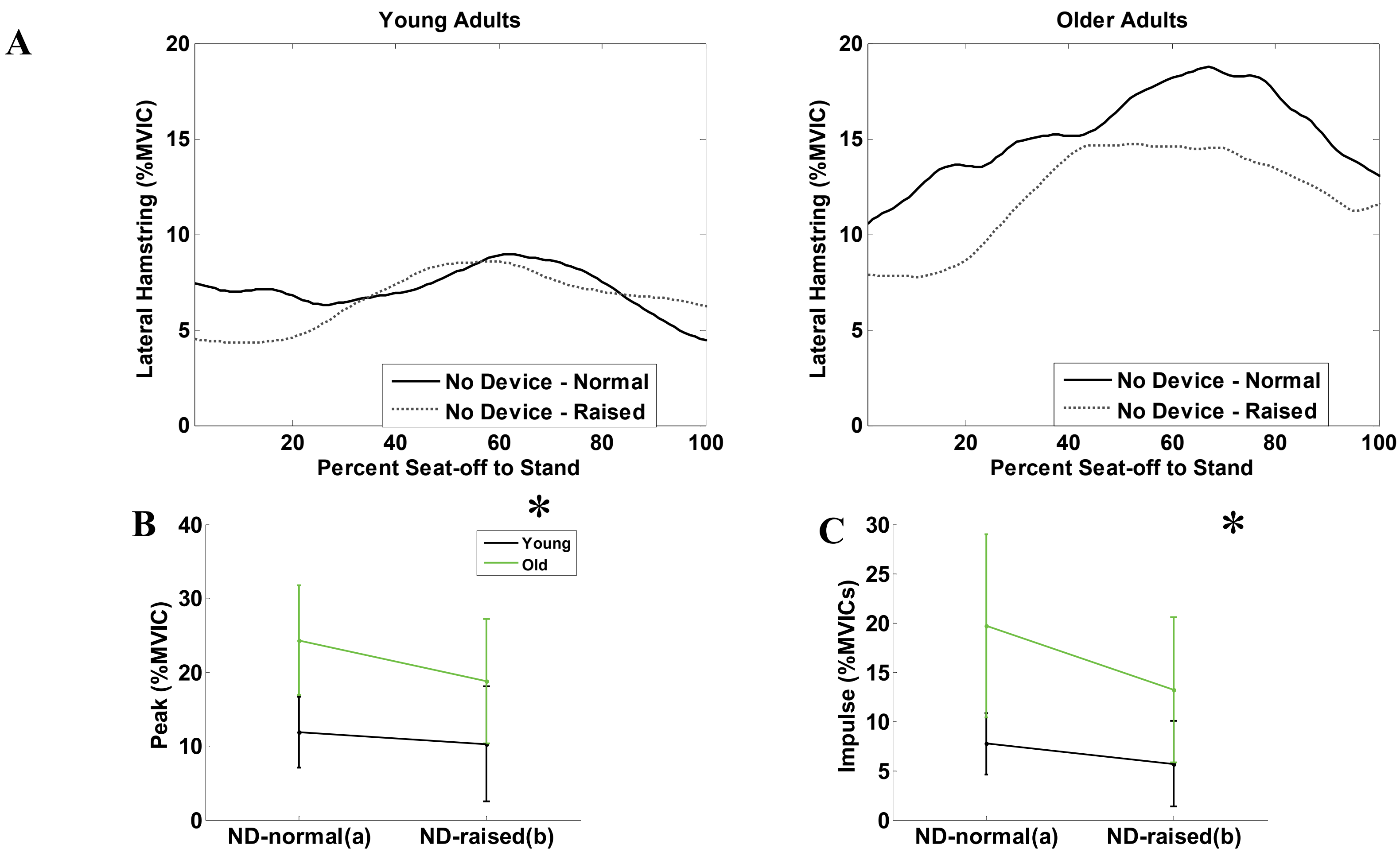


Figure E.9. Ensemble average EMG activity waveforms of the lateral hamstring for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

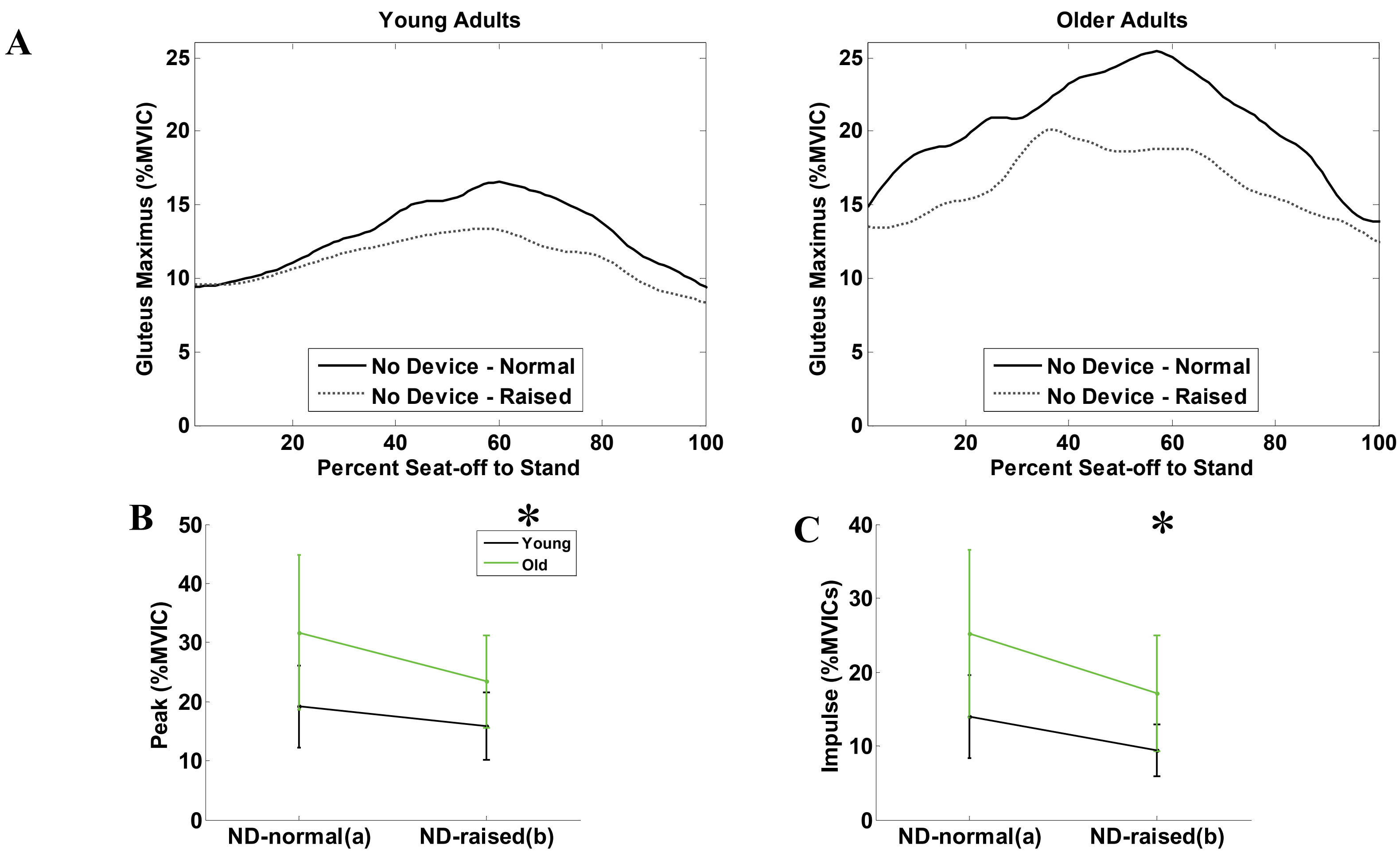
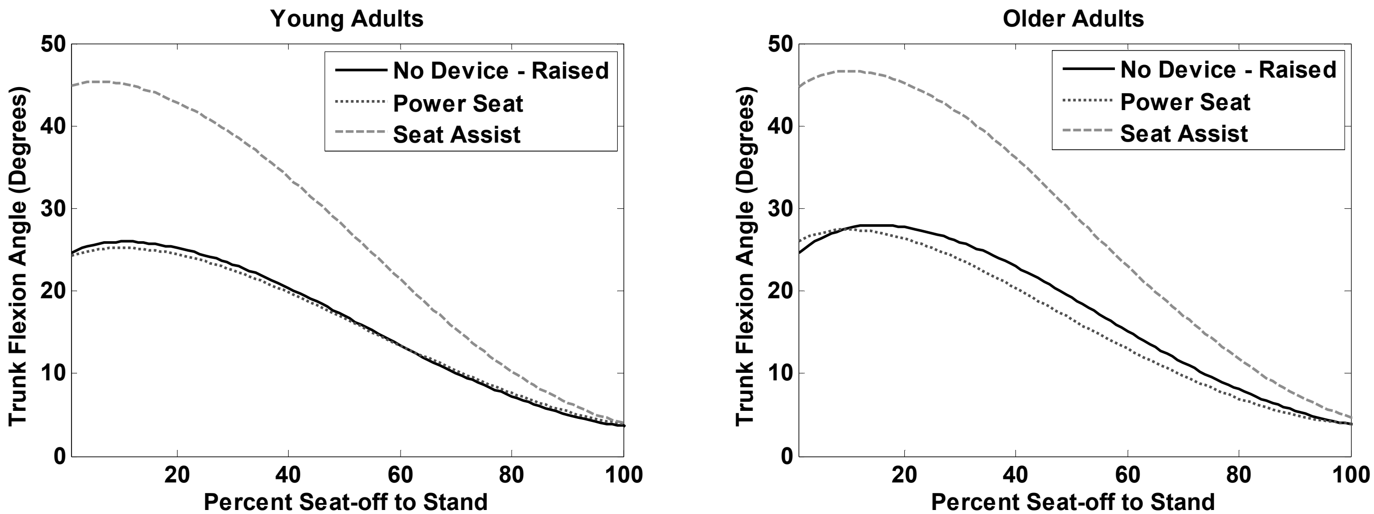


Figure E.10. Ensemble average EMG activity waveforms of the gluteus maximus for two different seat heights between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

APPENDIX F CHAPTER 5: ENSEMBLE AVERAGE WAVEFORMS AND INTERACTION PLOTS

A



B

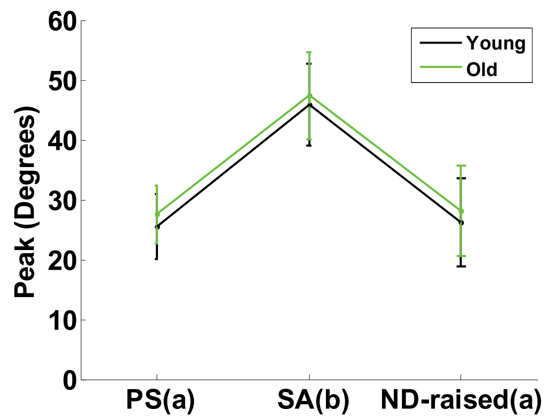


Figure F.1. Ensemble average trunk flexion angle waveforms for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for the peak values (B). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

A

B

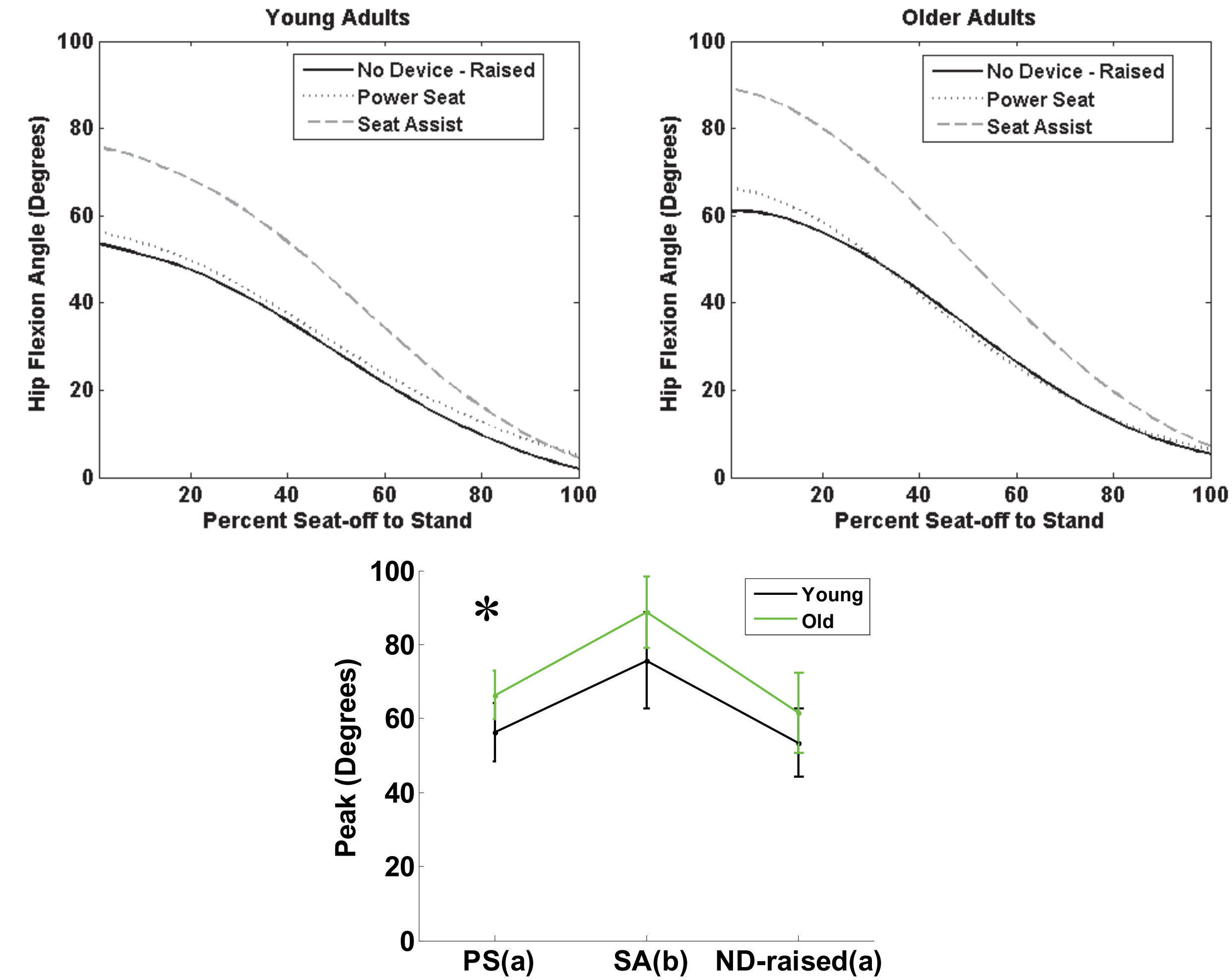


Figure F.2. Ensemble average hip flexion angle waveforms for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

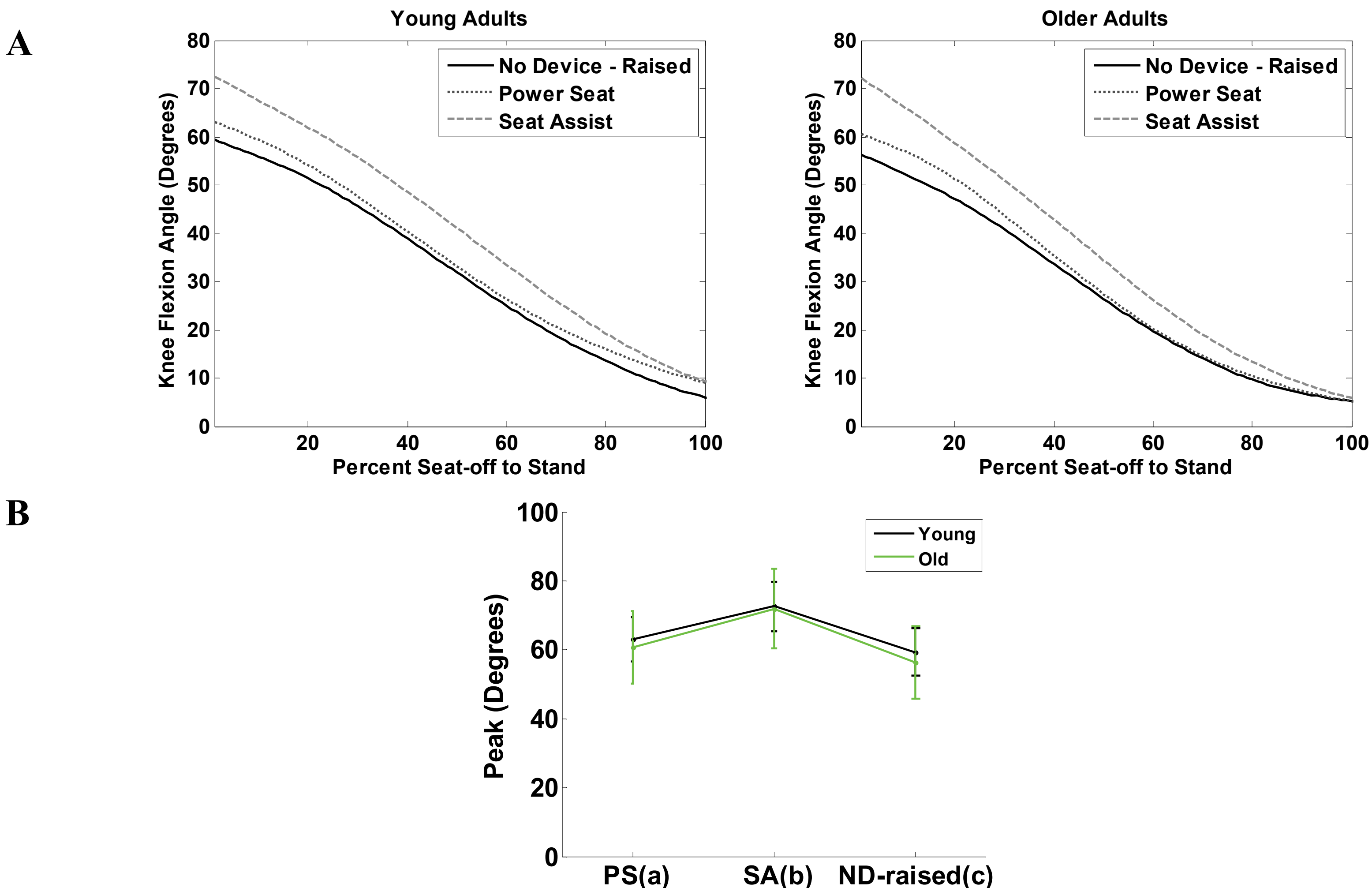


Figure F.3. Ensemble average knee flexion angle waveforms for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for the peak values (B). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

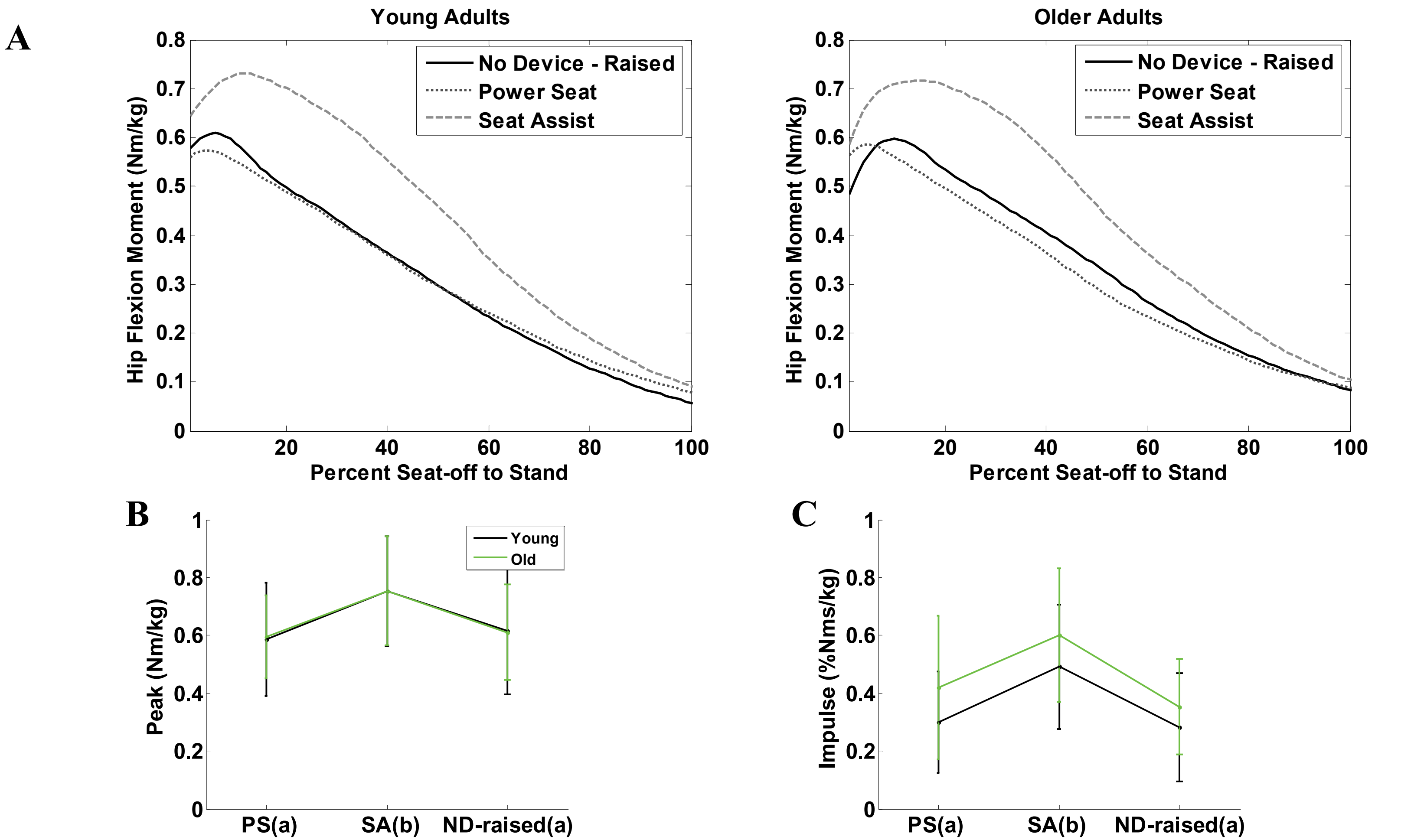


Figure F.4. Ensemble average hip flexion moment waveforms for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and impulse values (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

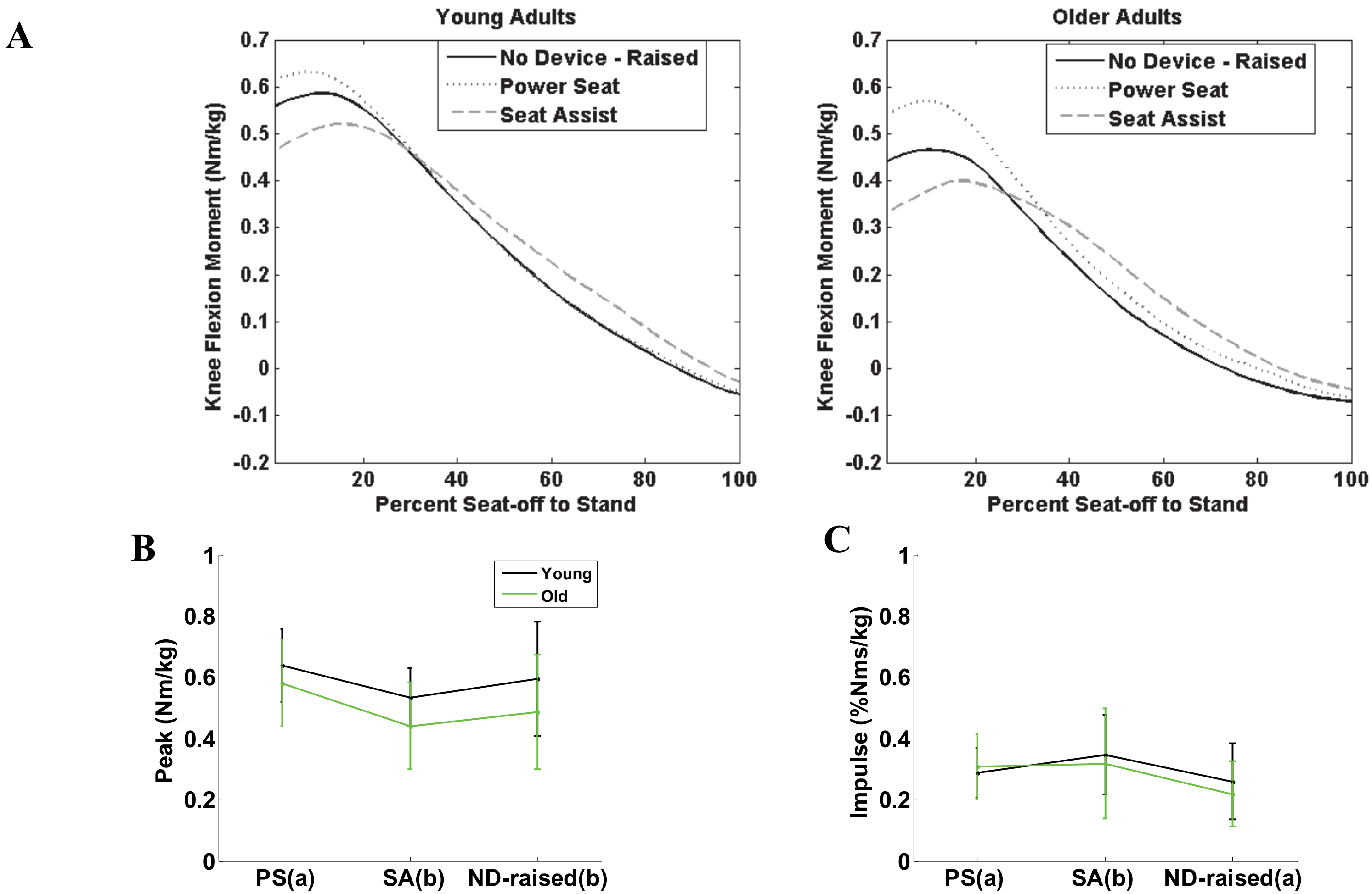


Figure F.5. Ensemble average knee flexion moment waveforms for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and impulse values (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

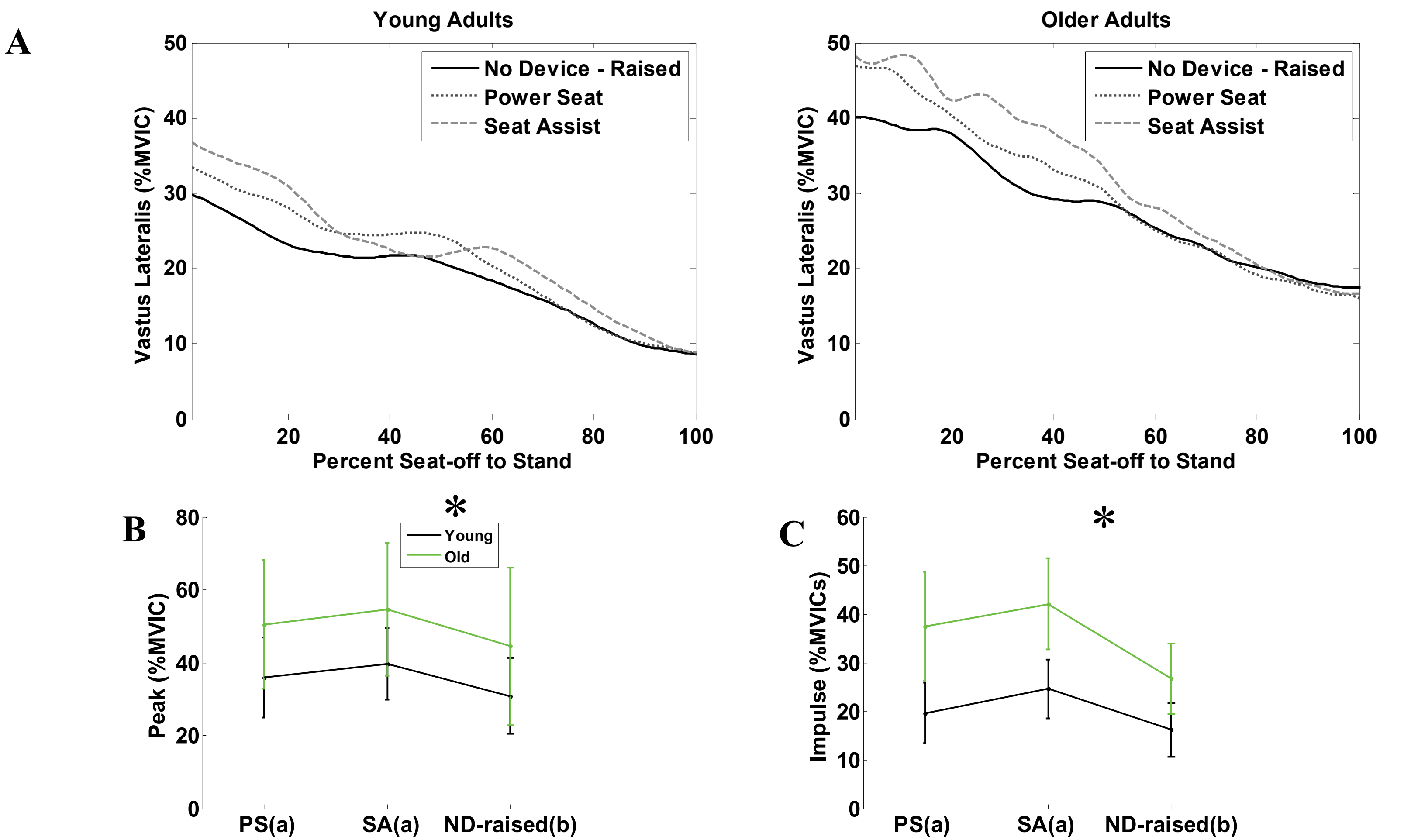


Figure F.6. Ensemble average EMG activity waveforms of the vastus lateralis for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for the peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

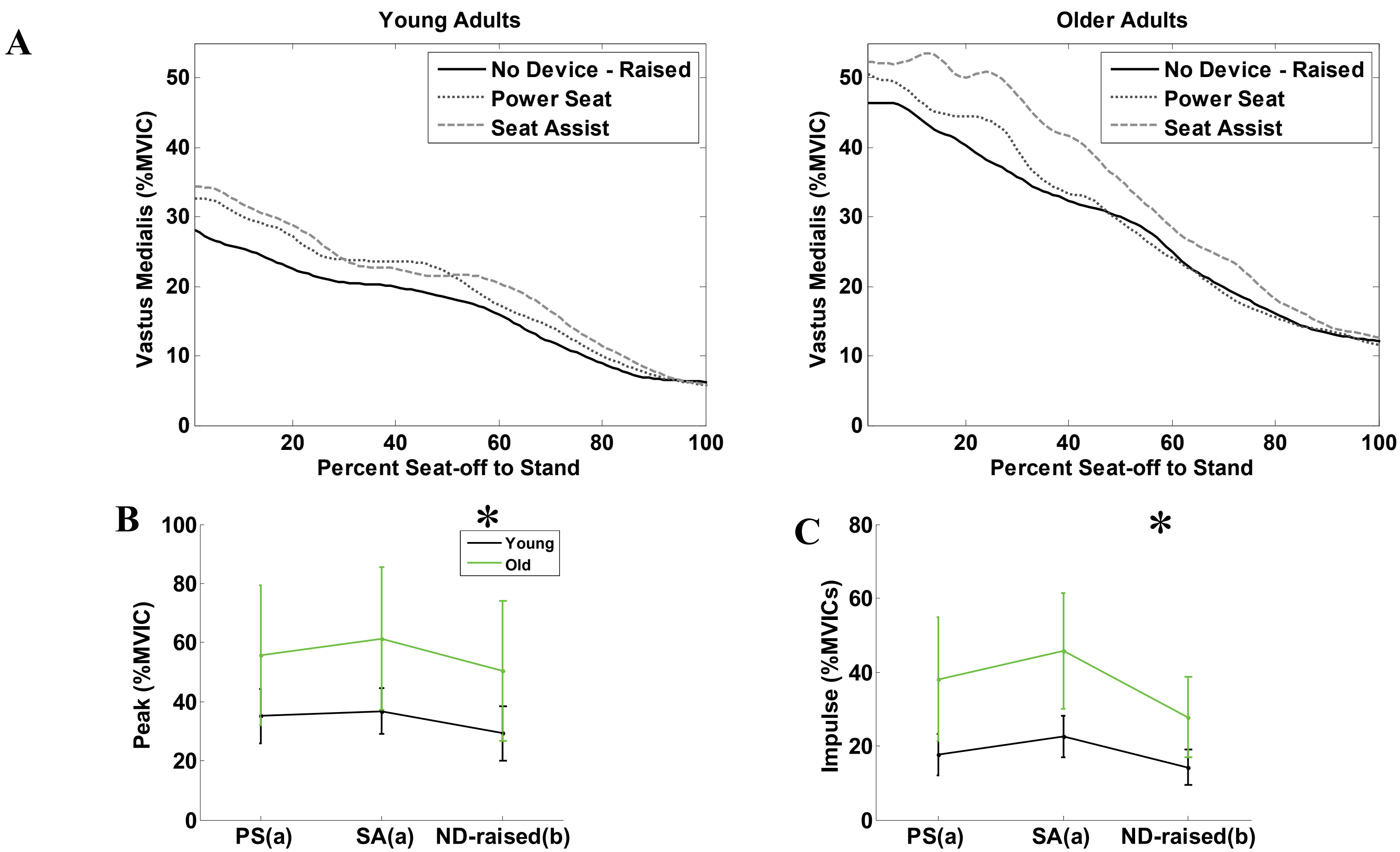


Figure F.7. Ensemble average EMG activity waveforms of the vastus medialis for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect (p<0.05) and different letters on the x-axis indicate a significant condition main effect (p<0.05).

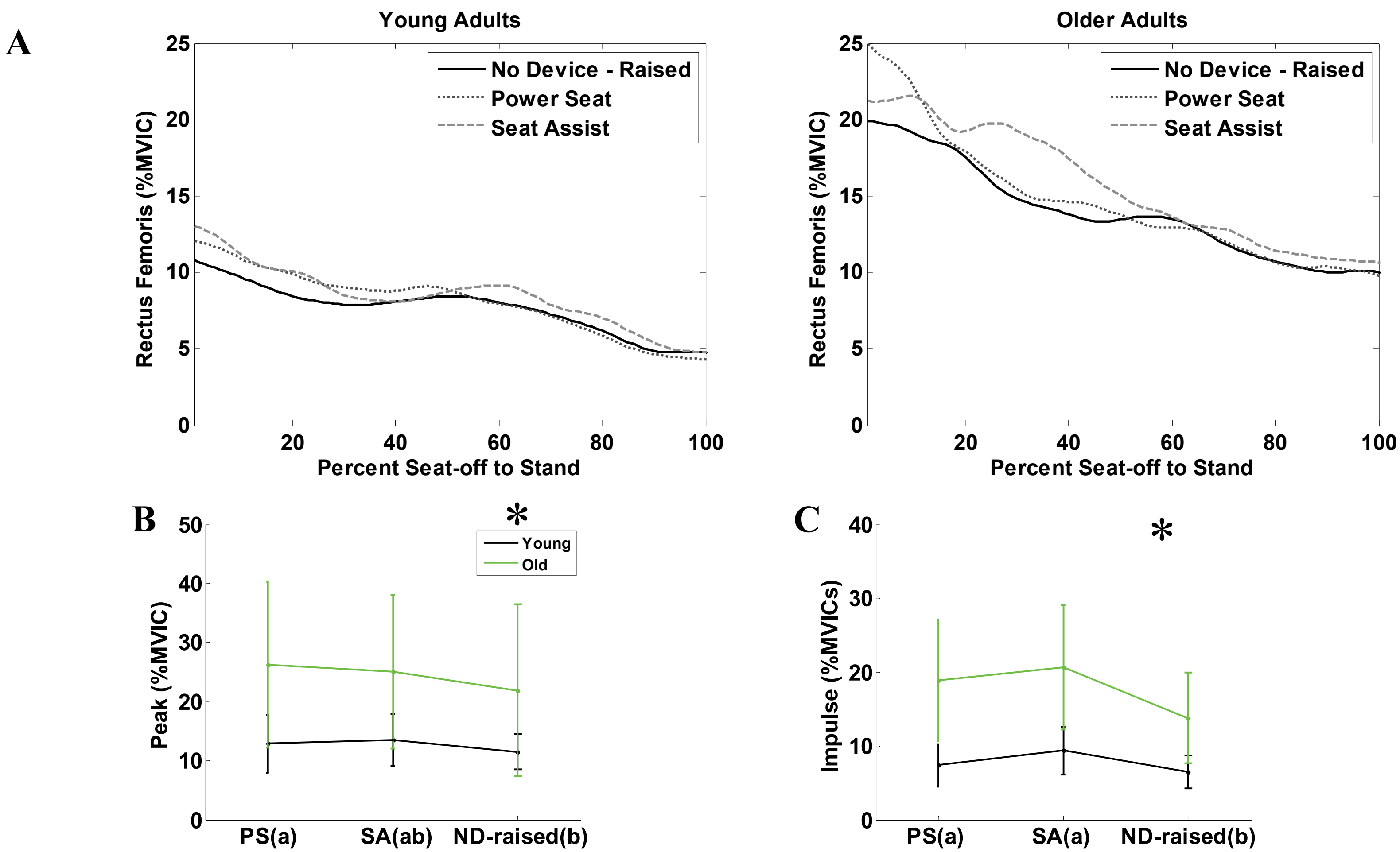


Figure F.8. Ensemble average EMG activity waveforms of the rectus femoris for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).

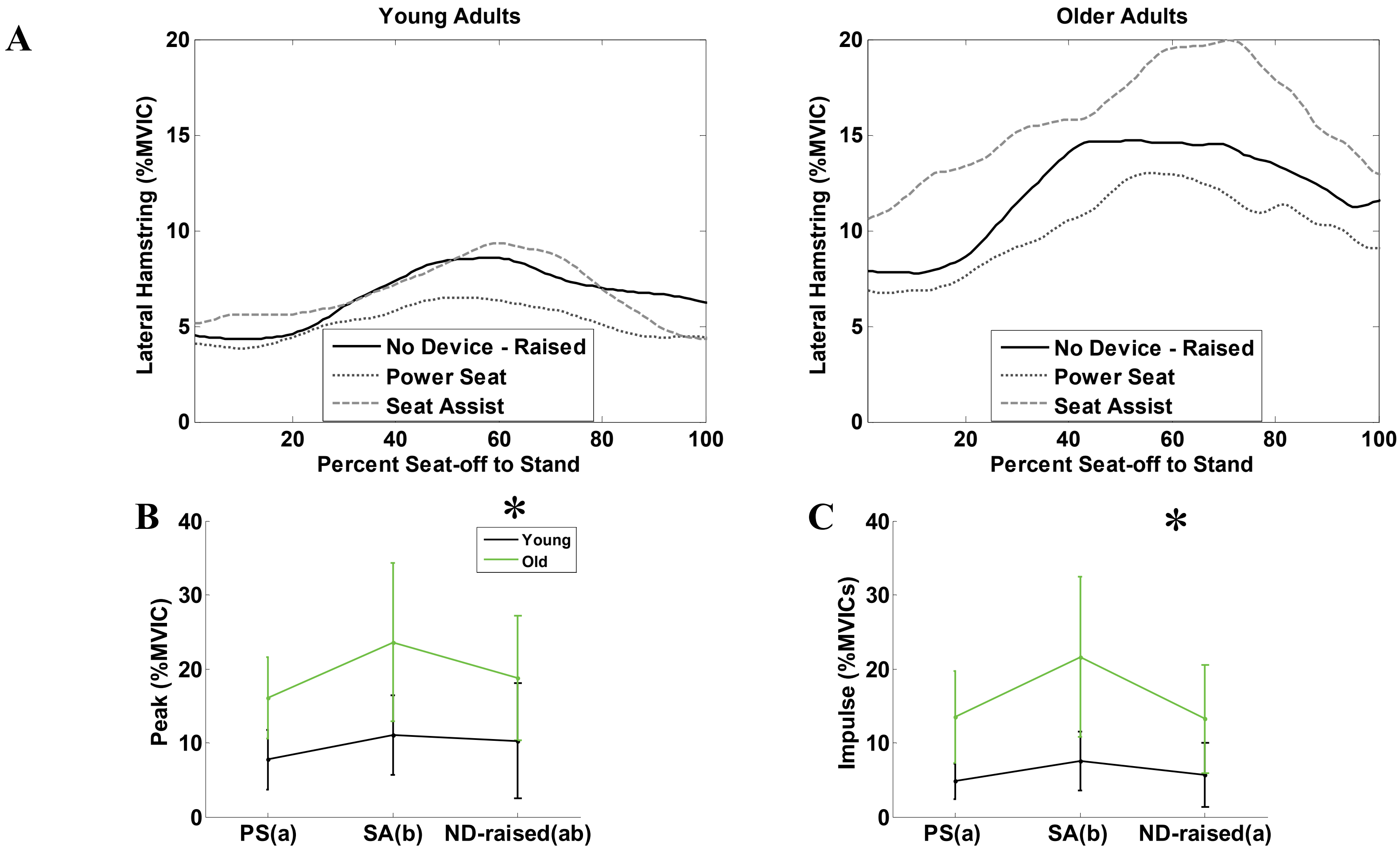


Figure F.9. Ensemble average EMG activity waveforms of the lateral hamstings for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect (p<0.05) and different letters on the x-axis indicate a significant condition main effect (p<0.05).

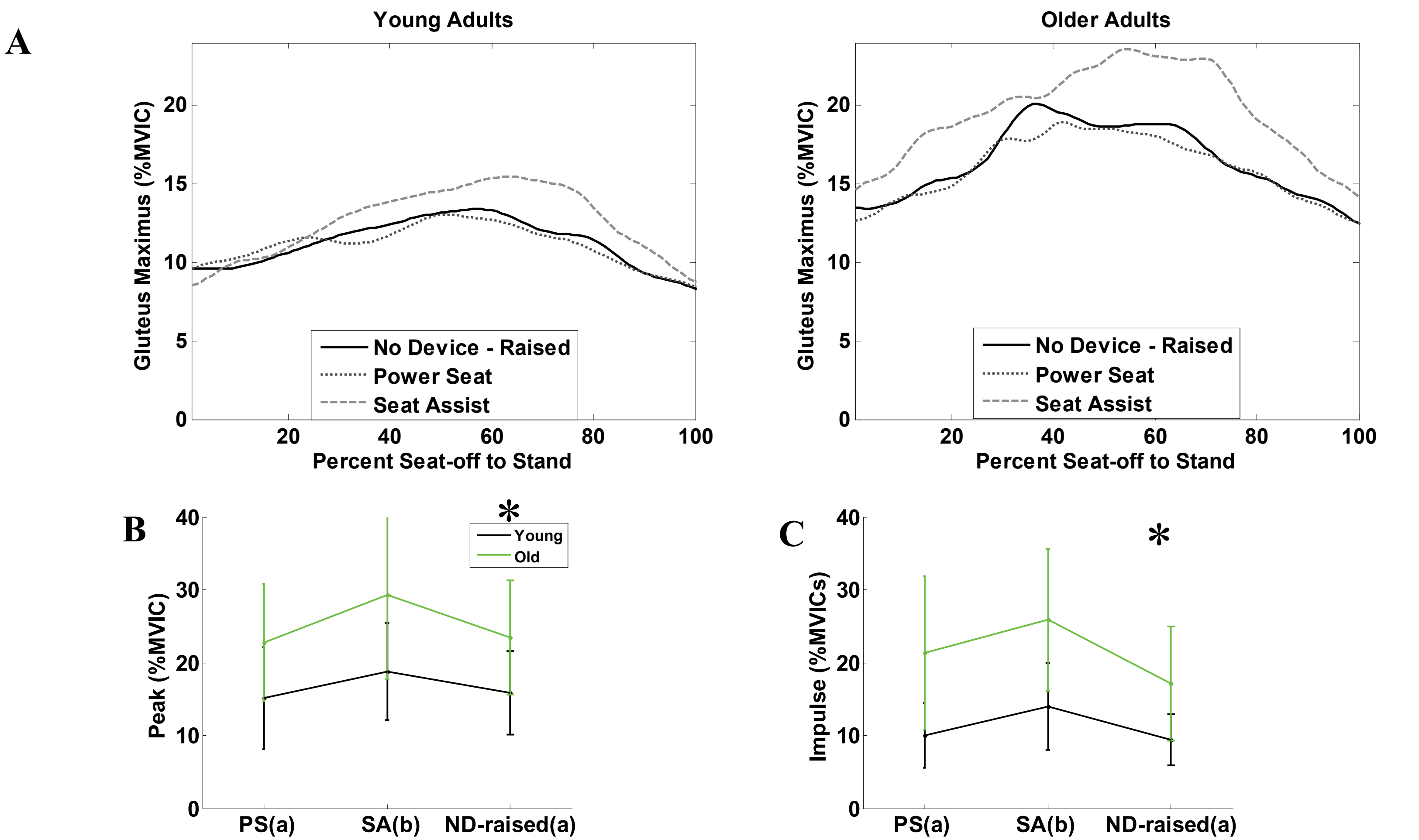


Figure F.10. Ensemble average EMG activity waveforms of the gluteus maximus for the three different seating conditions between the younger adults (A-left) and older adults (A-right). Interaction plots for peak values (B) and integrated EMG (C). * indicates a significant main group effect ($p<0.05$) and different letters on the x-axis indicate a significant condition main effect ($p<0.05$).