THE EFFECT OF HEALTHY AGING ON TRUNK NEUROMUSCULAR ACTIVATION PATTERNS DURING A CONTROLLED FUNCTIONAL TRANSFER TASK

by

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The undersigned hereby certify that they have read and recommend to the Faculty of Graduate Studies for acceptance a thesis entitled “THE EFFECT OF HEALTHY AGING ON TRUNK NEUROMUSCULAR ACTIVATION PATTERNS DURING A CONTROLLED FUNCTIONAL TRANSFER TASK” by D. Adam Quirk in partial fulfilment of the requirements for the degree of Master of Science.

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In loving memory of my father, David Michael Quirk. It is difficult for me to know that you could not be with me to see this journey come to an end. However, I am grateful that you were able to watch my ambitions and personality growth during my masters. I know you are very proud of my accomplishments. Thank you for all the support, guidance, and advice you provided me through my life, and during this process.

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ABSTRACT

Healthy aging is associated with an increased incidence of injuries such as low back pain and falling. Age-related changes in trunk neuromuscular activation patterns could potentially explain uncontrolled trunk motion, a risk factor for these injuries. The purpose of this study was two-fold: i) to establish if trunk neuromuscular activation patterns differ between older and younger adults, and ii) to identify if increased cognitive load can alter the activation patterns used by older adults. Participants performed a controlled dynamic transfer task known to challenge trunk musculature by continuously changing the external moment experienced by the spine. In this thesis, the transfer task was altered by increasing the external moment and cognitive load to address purpose i) and ii) respectively. Three dimensional trunk and pelvis motion and surface electromyograms for 24 trunk muscle sites were collected in two experiments. In the first experiment 26 younger adults, and 17 older adults, performed the transfer task at two physical task intensities. In a second experiment, eight of these older adults performed the dynamic transfer task while performing a simultaneous cognitive dual-task. An analysis of amplitude and temporal characteristics of EMG waveforms, using principal component analysis, showed that older adults had altered trunk neuromuscular activation patterns compared to younger adults. In particular, after accounting for physical task intensity, older adults have increased antagonist co-activation, less temporal variation to changing external moments, and changes in the synergistic relationships between trunk muscle sites. Increasing the cognitive load had no influence on the trunk neuromuscular activation patterns used by healthy active older adults. However, as cognitive performance was not measured in the dual-task it cannot be excluded that i) the cognitive task did not represent enough challenge to result in competition of cortical resources, or ii) the cognitive resources necessary for the successful completion of the transfer task were prioritized in older adults.

LIST OF ABBREVIATIONS USED

ANOVA	Analysis of Variance
BMI	Body Mass Index
CoM	Center of Mass
EEG	Electroencephalography
EMG	Electromyography
EO1	External Oblique (Anterior Fibers)
EO2	External Oblique (Lateral Fibers)
EO3	External Oblique (Posterior Fibers)
FOB	Flock of Birds
HT	Hand Transition
Hz	Hertz
IO	Internal Oblique
kg	Kilogram
L	Left
L13	Superior Erector Spinae Longissimus Lumborum
L16	Superior Erector Spinae Iliocostalis Lumborum
L33	Inferior Erector Spinae Longissimus Lumborum
L36	Superior Erector Spinae Iliocostalis Lumborum
L48	Quadratus Lumborum
L52	Superficial Multifidus
LHT	Left Hand Transfer
LRA	Lower Rectus Abdominis
m	Meter
MC	Motor Cortex
MMSE	Mini-Mental State Examination
MRI	Magnetic Resonance Imaging
MVIC	Maximum Voluntary Isometric Contraction
Nm	Newton Meter
PC	Principal Component
PCA	Principal Component Analysis
PMA	Pre-Motor Area
R	Right
RHT	Right Hand Transfer
SD	Standard Deviation
SMA	Supplementary Motor Area
TMS	Transcranial Magnetic Stimulation
URA	Upper Rectus Abdominis

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CHAPTER 1 INTRODUCTION

Projections are that by 2030, that over 25% of Canadians will be older adults exceeding 65 years of age [Health Canada, 2002]. Eighty-eight percent of older adults complete activities of daily living independently [Health Canada, 2002], thus it is concerning that older adults are at an increased risk of experiencing musculoskeletal injuries such as low back pain [Bressler et al, 1999; Gourmelen et al, 2007; Plouvier et al, 2011], or experiencing a fall [Pijnappels et al, 2010a; Scott et al, 2005]. Nova Scotia, is particularly affected by this aging demographic having one of the oldest populations in Canada, and reporting an above national average number of falls per year [Scott et al, 2005]. Not all musculoskeletal injuries are serious, less than half of low back pain cases [Hicks et al, 2008], and 58% of fall cases in older adults require medical attention [Scott et al, 2005]. However incidence of these injuries are predictors of fall risk [Ganz et al, 2007; Hicks et al, 2008; Ruhe et al, 2011] that may necessitate medical attention. It is believed that the increased risk of injuries is a consequence of age-related changes in structures required to control motion generated by both voluntary and involuntary movements [Horak, 2006]. Considering that altered movement patterns emerge as early as the 6th decade [Vernazza-Martin et al, 2008], it is pertinent to identify changes as a consequence of healthy aging and determine if these are maladaptive or compensatory in nature.

Total body motion necessary to achieve a movement goal represents the summation of motion at individual joints [Madhavan et al, 2009]. Motion at an individual joint is controlled by the interaction of the musculoskeletal and neuromuscular systems balancing forces to permit movement while maintaining joint and total body stability. Panjabi defined joint stability as preventing excessive/ undesirable motion beyond a physiological range that may result in injury. He conceptualized that three subsystems (passive-bones, joints, ligaments, active-muscles, and neural-central and peripheral structures) work in combination to maintain joint stability [Panjabi, 1992a; Panjabi, 2003a]. Since healthy aging is known to alter the physiological structures within these systems (see Figure 1.1), the challenge to control joint motion and stability increases with age.

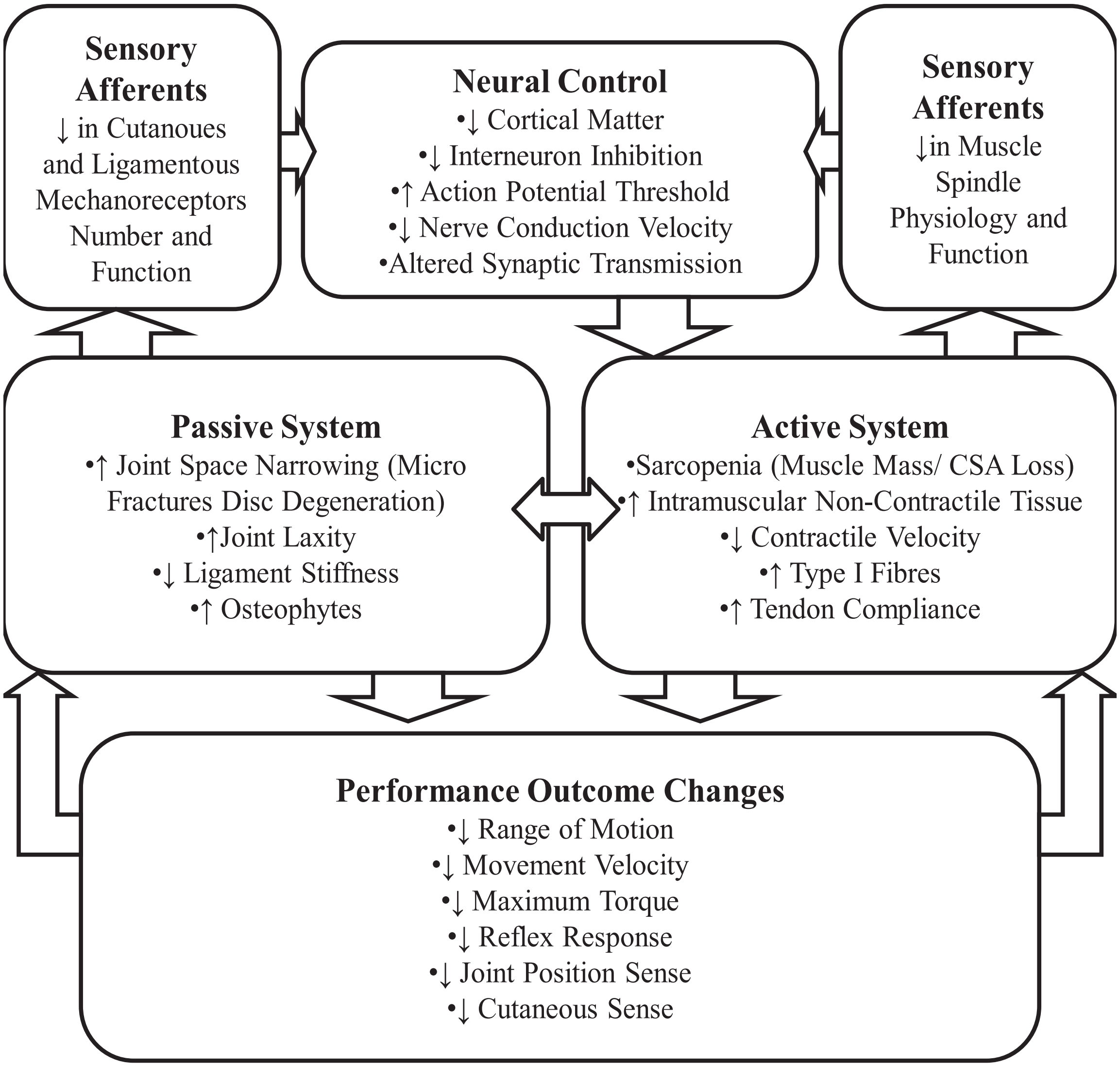

Figure 1.1: Conceptual framework of age-related changes in the three sub-systems involved in spinal stability and their effect on movement performance.

Age-related changes to physiological structures contribute to altered joint control (Figure 1.1); however, the interaction between these stability systems makes it difficult to suggest any one change is the major risk factor. In the passive system, age-related decreases in joint space [Ferguson and Steffen, 2003; Gallagher et al, 2007; Hangai et al, 2008; Hicks et al, 2009; Papadakis et al, 2011], results in an increase in the ligament and joint capsule toe region. Thus, greater joint motion must occur for these structures to resist strain in older adults [Barros et al, 2002a; Iida et al, 2002b]. The age-related delay in response to movement is a multi-system problem. In the neural system older adults experience changes in sensory organs that affect the quality and quantity of sensory information [Goble, 2009; Shaffer and Harrison, 2007; Verschueren, 2002]. An age-related decline in nerve conduction velocity [Kido et al, 2004b; Rivner et al, 2001] slows the rate of transmission of this information. Once sensory information reaches the central nervous system, reduced number and integrity of interneurons [Doi et al, 2011a; Goble et al, 2010; Kido et al, 2010; Van Impe et al, 2011a] increases the demands of the system to generate a motor response [Ward, 2006]. Once a muscle fiber is excited, the active system takes longer to generate muscle tension due to age-related declines in contraction velocity [D'Antona et al, 2003; Dalton et al, 2010; Narici et al, 2008; Taube et al, 2008]. These changes suggest that older adults have impaired capacity to mitigate unexpected joint motion particularly when the joint is in a neutral position. However, if older adults increased activation of their trunk muscles, they could increase active stiffness minimizing trunk motion [Vera Garcia et al, 2007]. Suggesting that changes in neuromuscular activation patterns could compensate for these age-related differences.

Evidence suggests older adults have diffuse recruitment of multiple cortical structures during movement tasks, which may participate in the generation of muscle action potentials to control joint motion. In particular diffuse recruitment of the primary motor cortex may reflect increased recruitment of muscle synergies, or antagonist activation while performing a motion task [Goble et al, 2010; Hortobagyi and Devita, 2006; Seidler et al, 2010; Van Impe et al, 2011b]. In addition to more overall activation in the motor cortex, older adults have increased activation of additional cortical that participate in motor task. This increased cortical activity suggests that older adults may modify cortical activation for: i) planning prior to voluntary motion [Inuggi et al, 2011; Levin et al, 2011], ii) increased attention to feedback while performing movement task [Heuninckx et al, 2008; Van Impe et al, 2011b; Inuggi et al, 2011], and iii) a continuous activation of motor cortex and pre-motor cortex suggesting feed forward control of voluntary movement [Goble et al, 2010; Inuggi et al, 2011; Ward, 2006].

Collectively the age-related changes in the spinal stability subsystems (Figure 1.1) and inadequate compensation in the nervous system could participate in the age-related risk of musculoskeletal injuries. Since joint motion is influenced by the muscles surrounding a joint, any factor that alters the timing or amplitude of the muscle responses can influence the stability of the joint [McGill et al, 2003a]. The outcome of this interaction between physiological decline and altered neural control mechanism is the emergence of new movement patterns i.e. joint kinematics and kinetics, or their driving motor patterns i.e. muscle activations. Figure 1.2, provides a summary of the physiological changes that occur with aging, including changes in cortical activation, and how motor performance may be altered because of these changes.

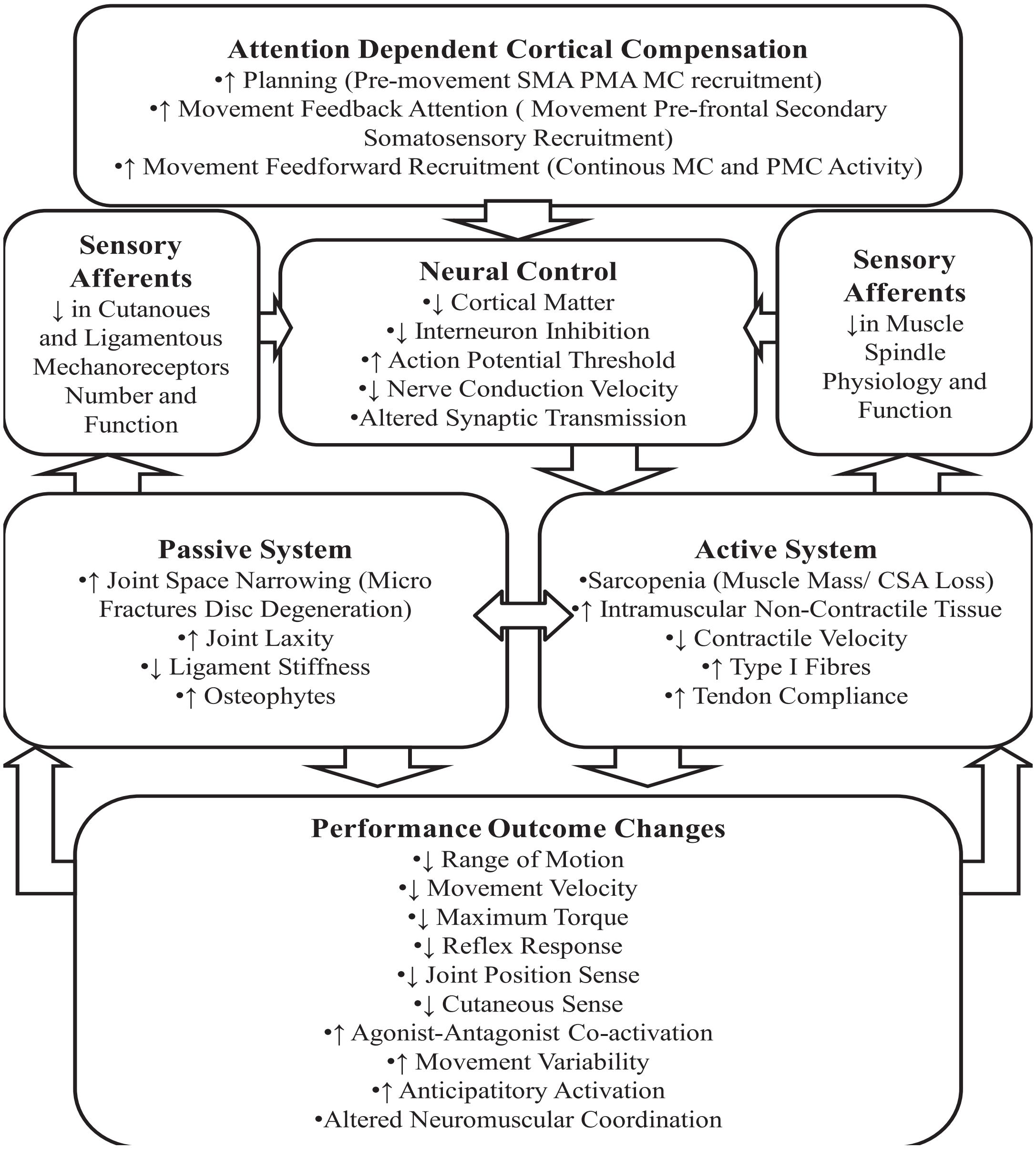

Figure 1.2: Conceptual model of how age-related changes in neural function interacts with the 3 subsystems of stability to modify movement performance. Abbreviations: SMA= supplementary motor area, PMA= pre-motor area, MC= motor cortex, CSA= cross sectional area

Cross-sectional and longitudinal studies have examined how motor and movement patterns change with healthy aging. Although these studies utilize different primary outcome measures (joint kinematics, kinetics, electromyography (EMG)) and movement tasks, most studies conclude that aging alters motor performance [Bleuse et al, 2006; Hortobagyi and DeVita, 2000; Hortobgyi et al, 2011; Kuo et al, 2011; McGill et al, 1999; Nagai et al, 2011a; Rankin et al, 2000; Seidler et al, 2002; Woollacott and Manchester, 1993]. Despite the multiple studies comparing motor performance between younger and older adults, few of these studies include measures to determine how the trunk muscles participate in trunk control. This is surprising given recent evidence supporting a link between trunk motion control and falling in the elderly [Doi et al, 2013; Grabiner et al, 2008b; Hicks et al, 2008]. This link can be explained since the trunk and head represents 60% of one's total body mass [Winter, 2009], therefore the location of this large mass has an influence on the location of the total body center of mass (Figure 1.3) within its base of support. The connection between trunk motion and balance is further supported by the relationship between mechanisms that disrupt joint control of the trunk, such as reduced muscle strength [Hicks et al, 2005a; Hicks et al, 2005b], induced muscle fatigue [Davidson et al, 2009; Lin et al, 2009; Suri et al, 2009], and reduced proprioception [Goldberg et al, 2005].

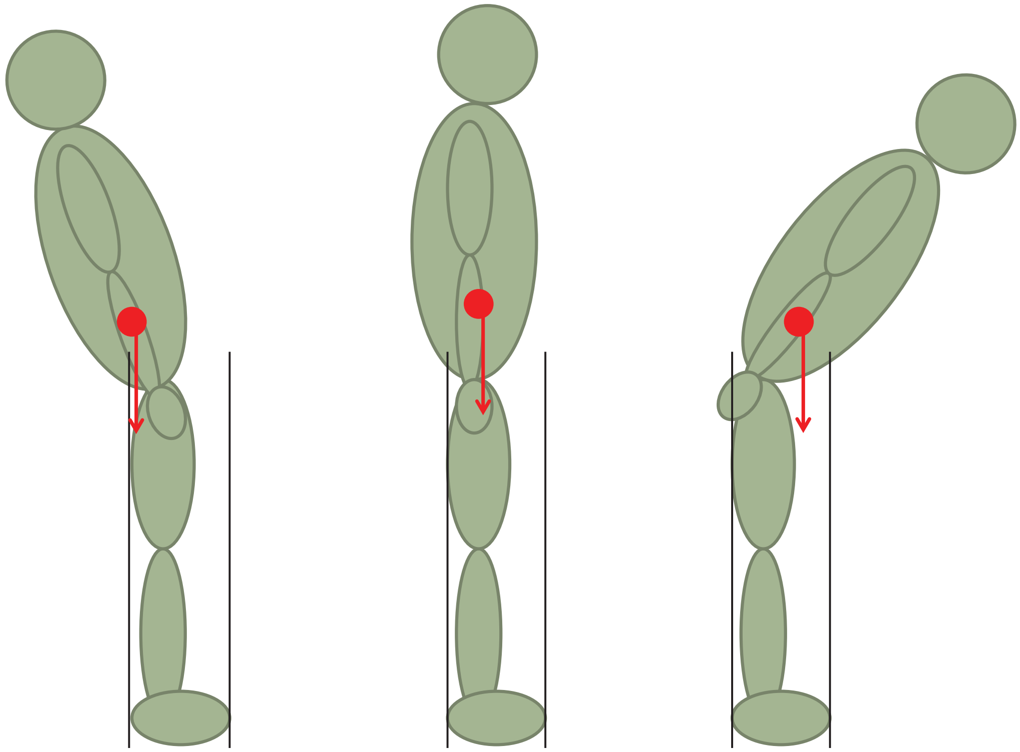

Figure 1.3: Representation of how trunk position (extension (left) to flexion (right)) contributes to total body center of mass (red dot). If the force vector of the center of mass (red arrow) lies close to or outside the base of support (depicted by the black lines) one would be at increased risk of fall.

Often studies of the trunk only analyze the kinematics or kinetics changes that occur with healthy aging [Burgess et al, 2009; Vernazza-Martin et al, 2008]. As a movement pattern is controlled by the summation of muscles acting around a joint [Arjmand et al, 2010; Rashedi et al, 2010] simply analyzing the kinematics ignores multiple combinations of muscle groups that produce these external moments [Rashedi et al, 2010]. In studies that analyzed trunk muscles, most are limited by the number of muscles (2-4) they quantify, assuming that the workload is evenly distributed between the muscle groups of the trunk [Asaka and Wang, 2008; Hwang et al, 2008; Kuo et al, 2011; Woollacott and Manchester, 1993]. This assumption has limitations since the trunk is composed of multiple muscles each with unique moment production capabilities. Trunk control is further complicated when one considers that some muscle groups have multiple fiber orientations, each with a unique moment arm length and physiological cross sectional area giving them moment production around the spine [Davis and Mirka, 2000; Ng et al, 1998; Urquhart et al, 2005]. Multiple studies confirm in younger adults that these muscle groups and their unique fibers are activated with respect to the external moments exerted by the movement task [Butler et al, 2010; Ceccato et al, 2009; Hu et al, 2011; Hubley-Kozey et al, 2009b; Talebian et al, 2010; Urquhart and Hodges, 2005; Vera-Garcia et al, 2010]. Therefore, analyzing too few muscles may not capture the complexity of the trunk, potentially masking age-related differences in the spatial-temporal control of movement [Hubley-Kozey et al, 2009b; McGill et al, 1999]. Finally, the outcome measures in trunk muscle EMG studies typically involve a form of data reduction that captures only one characteristic of the dynamic waveform. Typically only, the onset of muscle activity [Hwang et al, 2008; Inglin and Woollacott, 1988; Woollacott and Manchester, 1993], the peak or average activity amplitude across an entire phase [Kuo et al, 2011] are measured. Studies have attempted to elucidate the spatial-temporal response of trunk muscles by breaking down an entire movement in multiple phases [Butler et al, 2010; Kuo et al, 2011]. These strategies, while making data manageable, limit the ability to capture temporal features of muscle sites, which may further elucidate how muscle patterns vary instantaneously with a dynamic external challenge. Furthermore, these forms of data reduction hinder the analysis of temporal synchrony among multiple muscle sites where potential group differences could exist [Hubley-Kozey et al, 2012]. Thus, to determine how the neuromuscular system modulates its activity with healthy aging, there is a need to examine the time varying patterns of multiple muscle sites as shown qualitatively in previous studies using ensemble average waveforms [McGill et al, 1999].

Recent work in this laboratory captured a comprehensive view of the complex interactions among the trunk musculature by utilizing: i) multiple electrode placement and ii) pattern recognition techniques, to capture the time varying activation patterns of multiple trunk muscles while participants performed a dynamic exercise task [Hubley-Kozey et al, 2009]. This methodology quantified the common temporal responses of multiple muscle sites, allowing for the comparison of muscle sites and groups. This work showed that older adults utilize altered spatial and temporal patterns of the abdominal muscles to complete a dynamic controlled leg loading exercise task when compared to younger adults [Hubley-Kozey et al, 2009]. A follow up study established that older adults have higher back muscle activation, indicative of higher antagonist co-activation [Quirk and Hubley-Kozey, 2012]. This finding is consistent with other studies showing that older adults have increased agonist-antagonist co-activation around the knee and ankle joints [Hortobagyi and DeVita, 2000; Hortobagyi and Devita, 2006; Nagai et al, 2011a]. Co-activation is a strategy known to increase joint stiffness around these joints in older adults [Hortobagyi and DeVita, 2000], and increase trunk stiffness in younger adults [Brown and McGill, 2008a; Granata et al, 2001; Lee et al, 2006; Stokes et al, 2000a]. Whether trunk muscle activation patterns are altered in older adults while performing a functional task has not been studied.

In younger adults the trunk muscle activation patterns were modified in a systematic manner to the changing external moment while performing a functional dynamic lift and replace task [Butler et al, 2009a; Butler et al, 2010; Hubley-Kozey et al, 2012]. As older adults have a decrease in trunk muscle strength [Hasue et al, 1980; Kubo, 1994], one would expect that older adults would perform the same task at a higher percentage of physical intensity compared to younger adults. This theory is supported by the increased amplitude of normalized EMG in the trunk muscles of older adults [Kuo et al, 2011; McGill et al, 1999]. However increasing the physical intensity does not increase the activation of all trunk muscles proportionally in younger adults [Butler et al, 2010; Ng et al, 2003; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008]. Thus, age-related differences in trunk muscle activation may reflect the increase in physical task intensity. While non-uniform increases amongst muscle sites has been shown for older adults around other joints [Hoffren et al, 2007; Hortobgyi et al, 2011], this relationship has not been established in the trunk.

Finally, if age-related differences in trunk muscle activation patterns exist during a functional task, they could be explained by the multitude of changes in both the structures surrounding the joint, or a host of functional changes found in the cortex (Figure 1.2). To what extent cortical compensatory mechanisms contribute to these altered responses is not clear. Using a dual-task experiment, an individual is involved in a mental task while performing a movement (motor) task. This dual-task attempts to create a competition for cortical resources, between the mental and movement tasks.During dual-task conditions, older adults demonstrate deterioration in the motor performance relative to their single-task performance [Huxhold et al, 2006; Lacour et al, 2008]. Dual-task conditions have been shown to alter neuromuscular function in the ankle and knee joints of older adults [Rankin et al, 2000], as well as movement patterns in the trunk of older adults [Doi et al, 2011a]. However, to our knowledge no study has determined whether dual-task changes the neuromuscular activation pattern of trunk muscles in older adults during a dynamic movement task.

In summary, increasing age is associated with changes in the physiological structures and function that contribute to reduced joint control. The impact of these changes could be minimized by the use of altered cortical control. Neuromuscular patterns generated by the summation of spinal and cortical activity offer an ability to quantify how older adults utilize altered motor patterns to complete a movement task. There is evidence that trunk muscles of older adults have altered neuromuscular patterns for exercise tasks, but to date there are no comprehensive analysis of trunk muscle to determine the age-related change in trunk muscles during a controlled dynamic functional task. The latter would have direct relevance to most activities of daily living. Furthermore, to our knowledge no study has attempted to: i) elucidate the extent that age-related changes in trunk muscle activation patterns are explained by changing physical task intensity differences between older and younger adults, or ii) determine whether altered trunk neuromuscular patterns can be modified by increasing the cognitive load of a movement task in older adults.

1.1 PURPOSE AND OBJECTIVES OF THE THESIS

The overall purpose of this research was to investigate the age-related changes in trunk muscle activation patterns while performing a controlled dynamic right to left lift, transfer, and replace task designed to challenge trunk stability. Secondly, this research attempted to determine if these changes are the consequence of age-related compensatory mechanisms in neuromuscular control, or a product of the increased physical demand, or increased movement attention. One primary objective and two sub objectives were examined.

The primary objective was to determine if healthy aging results in altered trunk muscle activation patterns while performing a dynamic functional lift and replace task known to challenge the trunk musculature.

The two sub-objectives are:

1. To determine if altered trunk muscle activation patterns in older adults are accounted for by differences in relative task intensity by changing the external moment applied to the trunk.
2. To determine if trunk muscle activation patterns used by older adults can be changed by increasing the cognitive load of a movement task through the addition of a cognitive dual-task.

To address the primary objective and two sub-objectives, two experiments were conducted with participants performing a modified dynamic right-to-left lift, transfer, and replace task while attempting to maintain a neutral spinal posture and minimize trunk motion. This task challenged the neural and active sub systems, since the passive sub system has minimal contribution to spinal stability in a neutral spine position [Cholewicki et al, 1997]. During these tasks muscle activation patterns were quantified using surface EMG electrodes placed on 24 muscle sites including agonist (back extensor) and antagonist (abdominal wall) muscles providing a comprehensive assessment of main muscle fibers associated with control of the lumbar and pelvis segments. The ability to minimize lumbar–pelvic motion was monitored by an electromagnetic motion tracker. As outlined in Figure 1.4, the primary objective and sub-objective one were addressed in Chapter 4 (Experiment 1), where sub-objective two, was addressed in Chapter 5 (Experiment 2).

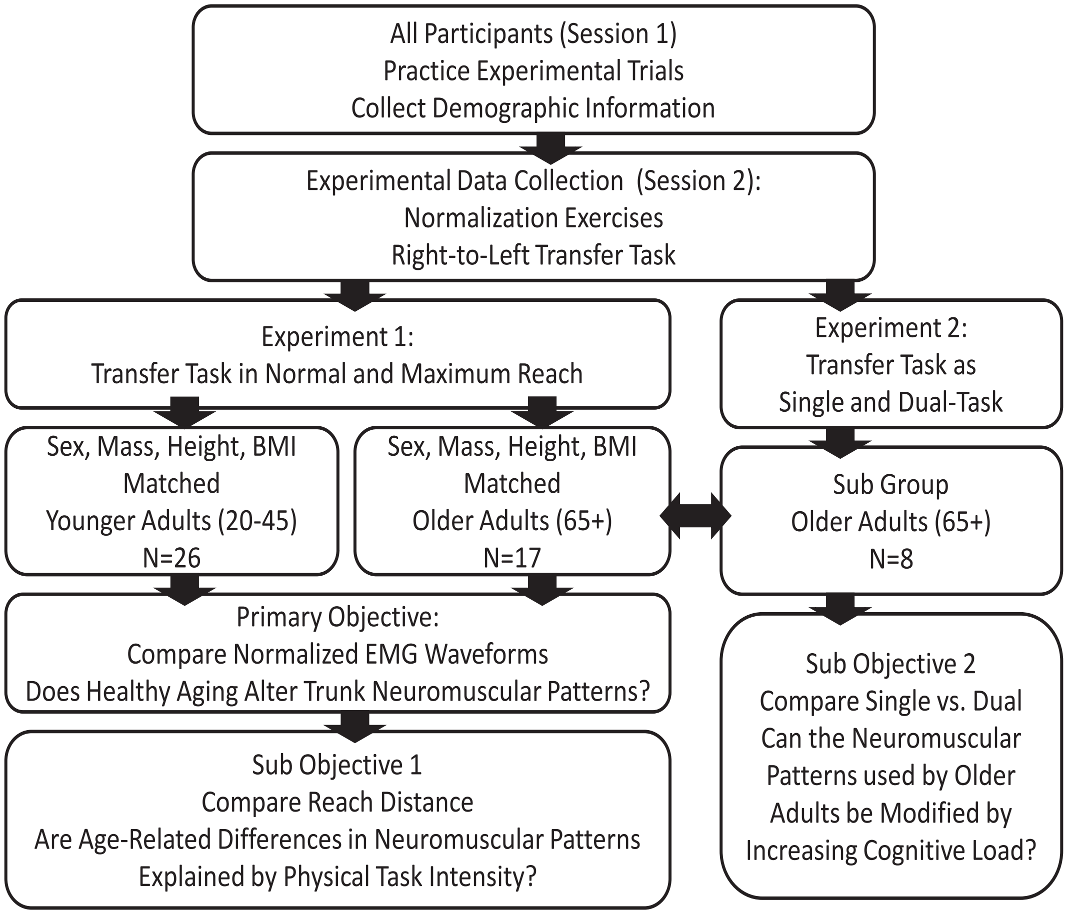

Figure 1.4: Flow chart of the experimental design for the primary objective and two sub objectives.

1.2 HYPOTHESIS

Age-related changes in the structures that control joint motion (active, passive, and neural), and cortical activity for planning and execution of voluntary movement will alter neuromuscular activation patterns used by older adults during a dynamic voluntary lift and replace task. Specific hypothesis related to the objectives include

1) Compared to young adults, older adults will have altered trunk muscle activation patterns including increased activation of both agonists and antagonist trunk muscles, less change muscle activation in response to changing external moments, and changes in the temporal synergies shared by multiple trunk muscle sites.

2) Age-related differences in trunk muscle activation patterns between older and younger adults will not be explained by the change in physical task intensity.
3) Older adults will have an alteration in trunk muscle activation patterns while attempting to perform a simultaneous dual-task compared to a single task.

1.3 ASSUMPTIONS

The major assumptions of this study are:

- That minimal movement occurred between the electrodes and the underlying muscles during the EMG recordings (controlled by the use of a predominantly an isometric task to assess participants)

- That no learning effect occurred between trials (controlled for by using a familiarization session, practice trials and randomization of conditions).

- That fatigue was not present during the testing (by giving periodic rest breaks between block conditions and MVIC)

- That the participants were able to exert maximal voluntary isometric contraction during the normalization exercises (improved by practice contractions in the familiarization session).

- That surface electrodes were placed above the muscle fibers of interest, and there would be minimal cross talk (controlled by literature-based guidelines, palpation of electrode sites, and validation exercises to test unique muscle fiber recruitment).

1.4 THESIS STRUCTURE

Chapter 1 introduces the thesis, the topic of healthy aging and the rational for studying age-related differences in trunk neuromuscular patterns. One primary objective, two sub objectives, and hypothesis are stated. A brief overview of experimental design is presented with key assumptions.

Chapter 2 provides a comprehensive overview of literature relevant to this thesis. The literature establishes the theoretical framework of how age-related changes in specific tissues can alter the ability to maintain joint stability. It further establishes a relationship on how healthy aging alters movement and motor patterns, and methodologies necessary to quantify these alterations.

Chapter 3 provides a comprehensive overview of the general methodology employed to carry out the objectives of the thesis. Specifically this section was divided into three sub-sections. Section 3.1 described the common methodology used for participant recruitment and setup and collection of electromyographic and kinematic data. Section 3.2 explained the experimental task design. Finally, Section 3.3 explained data processing including pattern recognition methods and statistical analysis.

Chapter 4 addresses the primary objective and first sub objective of the thesis. This experiment was written as an original manuscript intended for scientific publication titled “Are age-related changes in trunk neuromuscular activation patterns during a controlled functional transfer task explained by increasing physical task intensity?” to be submitted as a manuscript to a scientific journal.

Chapter 5 contains experiment two related to sub-objective 2 of the thesis. This preliminary data was written in a mini scientific paper format titled “Does a dual-task paradigm during a controlled functional transfer task performed by healthy older adults alter trunk neuromuscular activation patterns?”.

Chapter 6 contains the conclusion of this thesis, providing a summary of key results and their implications. It identifies the limitations of these studies, and gives directions for future studies to understand how healthy aging may alter joint control.

Appendix A contains a published abstract, and figures, of a manuscript intended for scientific publication titled “Age-related changes in neuromuscular activation patterns of trunk muscles during a dynamic leg loading perturbation task”. This work is cited in the thesis and was presented as a poster at the Canadian Society for Biomechanics, Vancouver, British Columbia, Canada, June 6-9, 2012.

CHAPTER 2 REVIEW OF THE LITERATURE

Aging is associated with deterioration in structures that control joint motion. However, changes in tissue do not fully explain age-related differences in motor performance. Alterations in motor control are also controlled by the central nervous system that can modify neuromuscular function to maintain the joint stability. The purpose of this literature review is to analyze the physiological changes associated with aging that disrupt the control of joint motion, and how the nervous system can modify its function to compensate for these changes. After considering the theoretical construct to why older adults have altered motor performance, this review of the literature will evaluate how healthy aging altered motor performance, along with potential confounders known to alter motor performance. Special attention will be given to studies focusing on control of the trunk in older adults. As there are many potential ways to quantify age-related differences in motor performance, a secondary purpose of this literature review will be to assess methodologies on how to quantify and design an experiment to capture age-related changes in motor control, while minimizing the risk of false negative conclusions.

2.1 SYSTEMS INVOLVED IN JOINT STABILITY AND THEIR AGE-RELATED CHANGES

The maintenance of controlling joint motion is a complex problem that is solved by all structures surrounding a joint, combined with additional sensory information provided by somatosensory, visual, and vestibular systems [Horak, 2006]. According to the work of Panjabi et al., these structures can grouped into 3 subsystems (passive, active, and neural), each system being co-dependent but providing unique abilities to maintain joint stability [Panjabi, 1992a; Panjabi, 1992b; Panjabi, 2003a; Panjabi, 2003b]. With healthy aging structural changes in these systems, contribute to decreased joint control. This section will analyze the role of these various systems and their age-related changes.

2.1.1 The Passive System

The passive system represents stiffness generated by joint structures, such as the bones, ligaments, and joint capsules, to resist joint motion [Goodworth and Peterka, 2009]. These structures share common properties where if in a neutral position, the application of stress (force per area) will result in large strain (tissue displacement or deformation) in a region known as the "slack zone" or toe region [Panjabi, 1992b; Panjabi, 2003b]. Once strain reaches a threshold these passive tissues counter strain by applying a force relative to their cross sectional area, known as stiffness. This relation remains linear until permanent deformation or failure occurs [Panjabi, 1992b; Panjabi, 2006]. At the level of a joint, this relationship is more complex. The mixed composition of biological tissues that make up a joint each have a unique toe region and stiffness values resulting in a non-linear stress-strain relationship [Solomonow, 2011]. Further complexity exist in the passive system when one considers that the stiffness properties of a tissue depends on the rate of strain [Solomonow, 2009], and previous loading events from which a tissue has not fully recovered, known as tissue fatigue (creep) [Ben-Masaud et al, 2009; Sanchez-Zuriaga et al, 2010; Solomonow, 2011]. Therefore, each joint loading event does not have a consistent response from the passive system.

With aging there are numerous changes in the tissues of the passive system. Increasing age is associated with a deceased bone mineral density and bone mineral content [Guadalupe-Grau et al, 2009; Syed and Ng, 2010], predisposing older adult to an increased risk of vertebral end plate micro fractures [Ferguson and Steffen, 2003; Gallagher et al, 2007; Wang et al, 2012]. End plate fractures increase the compression experienced by the spinal disc of older adults increasing the probability of degenerative disc disease relative to younger adults [Hangai et al, 2008; Hicks et al, 2009]. The combined effect of end plate fractures and disc degeneration is joint space narrowing, decreased distance between two vertebral bodies [Hangai et al, 2008; Papadakis et al, 2011]. Age-related joint space narrowing effectively increases "toe region" of older adults ligaments and joint capsules resulting in these tissues requiring a larger displacement from neutral zone to occur before these tissues can apply stress [Solomonow, 2009; Solomonow, 2011]. Once these tissues are lengthened to threshold, their stiffness properties are different from younger adults. The spinous ligaments in older adults have increased calcification with different thickness and layering of elastin and collagen fibers compared to younger adults [Barros et al, 2002]. This results in older adults having a more compliant ligament that requires less loading before the tissue achieve permanent tissue deformation (failure) [Gallagher et al, 2007; Iida et al, 2002b]. Reduced elastic properties of ligaments in older adults have additional implications for these tissues have a slower return to resting length after loading [Iida et al, 2002b], suggesting that older adults relative to younger adults have slower recovery from an acute loading event [Solomonow, 2011]. Thus, if spinal stability were controlled for by only connective tissue the vertebral bodies of older adults would experience a greater displacement from neutral zone under loading conditions until these tissues reached failure.

Fortunately, bone remodeling that occurs with old age reduces the probability of ligament damage. Increased boney growths (osteophytes) at the borders of the facet joints and lateral surfaces of the vertebral body increase with age at all spinal levels [Hicks et al, 2009]. Functionally age-related changes in the connective tissue result in a less stable spine while the joint is in a neutral position, however since boney structures prevent vertebral end range of motion [Panjabi, 2003a; Panjabi, 2003b], older adults have decreased range of motion for vertebral bodies in all planes of motion [Intolo, 2009].

2.1.2 The Active System

Joint motion is not controlled by just the passive system. The musculature that surrounds the joint can generate tension to increase the stiffness of the spine to control joint motion. This system has an advantage relative to the passive system whereby it can modulate joint stiffness even in the neutral zone [Cholewicki et al, 1997]. Furthermore, the tension generated by muscles can influence joint stability by either producing a counter moment to oppose an external force, or by increasing tension of multiple muscles which will increases the stability a joint regardless of the direction of an external force [Brown and McGill, 2008a; Cholewicki et al, 1997; Cholewicki and VanVliet, 2002; Stokes et al, 2000a; Stokes et al, 2000b]. For this reason in joints that would be considered otherwise "unstable", such as the spine [McGill et al, 2003a], can be stabilized by tension generated by the trunk muscles [Cholewicki et al, 1997; Granata and Marras, 2000].

Older adults experience a 45-60% decrease in back extensor [Ajit Singh et al, 2011; Kubo, 1994; Sinaki et al, 2001; Yassierli et al, 2007] and a 35% reduction in trunk flexor strength [Kubo, 1994]. A decrease in strength is partially explained by less muscle mass and cross sectional area with advancing age. Cross sectional studies have identified that a 10% decline in muscle mass is experienced by the age of 50 compared to young adults in their 20s [Doherty, 2001; Frontera et al, 2000] with the rate of lean tissue loss accelerated beyond the age of 60 to a rate of .2-.3 kg/year [Koster et al, 2010]. Decline in muscle mass and cross sectional area explain strength loss, but this is not the only factor.

Age-related strength loss is partially independent of muscle tissue finding that older adults have reductions in muscle quality, force per cross sectional area [Delmonico et al, 2009; Frontera et al, 2000]. An age-related decline in muscle quality is explained by multiple factors. Within both the trunk and other joints, increasing age results in a change in muscle fibers type. With increasing age fast twitch (type II) muscle fibers become less prevalent with a shift towards type I or slow twitch muscle fibers [Callahan et al, 2009; D'Antona et al, 2007; Delbono, 2011; Fling et al, 2009; Kent-Braun, 2009; Yassierli et al, 2007]. Type I muscle fibers produces less force per cross sectional area [D'Antona et al, 2003] relative to type II, or fast twitch muscle fibers. However, despite fiber type changes when older adults' muscles fibers are corrected for fiber type, cross sectional area and extracellular connective tissue they still have reduced force production properties compared to muscle fibers in young adults [Frontera et al, 2000].

Additional factors that contribute to age-related decrease in muscle quality include changes in the cellular structure and organelle function in older adults' muscle fibers. Sarcolemma in older adults have more non-contractile material including intramuscular fat and connective tissue relative to young adults [Anderson et al, 2012b; Delmonico et al, 2009; Doherty, 2001a; Hicks et al, 2005a]. Muscle tension generation depends on the sarcoplasmic reticulum to release intramuscular calcium allowing actin-myosin cross bridge formation to generate a power stroke, if sufficient ATP is present to drive this process. In older adults the release and requisition of calcium in the sarcoplasmic reticulum is slowed [Delbono, 2011; Klass et al, 2005]. Myosin heavy chain protein type is also known to change with increasing age reducing force production abilities relative to the myosin chain type found in younger adults [D'Antona et al, 2003; Frontera et al, 2000].

While the above mechanisms partially explain the decline in strength associated with aging, they also contribute to strength differences being dependent on the contraction type [Callahan et al, 2009]. Older adults have higher eccentric strength than young adults when matched for muscle cross sectional area [Roig et al, 2010]. This is believed to occur from slower contractile properties of actin myosin cross bridges detachment [Roig et al, 2010], and increased intramuscular connective tissue increasing the passive stiffness of muscle fibers [Doherty, 2001a]. In concentric contractions, older adults have further reductions in strength and power, force per velocity [Clark and Fielding, 2011]. Changes in muscle fiber type and myosin chain type result in older adults having slower contractile properties (half contraction time) than younger adults. Reduced half contraction time is explained by slower cross bridge cycling [D'Antona et al, 2003; Delbono, 2011], and slower release of calcium required for actin-myosin cross-bridge formation [Callahan et al, 2009; Chung et al, 2005; Dalton et al, 2010; Klass et al, 2005]. When muscles begin to exert force in older adults' initial force is lost into increased tendon strain resulting from less tendon stiffness [Dalton et al, 2010; Hoffren et al, 2007; Narici et al, 2008].

With increasing physical task intensity, muscular fatigue is more likely to occur in both younger and older adults [Ajit Singh et al, 2011]. However, an interesting age-related phenomenon exists when if physical task intensity is matched for older and younger adults as a percentage of the maximum strength older adults experience less fatigue, quantified as a decline in force over time, relative to their younger counter parts in isometric contractions [Kent-Braun, 2009; Yassierli et al, 2007]. This would assist older adults by reducing the fatigue that is experienced by using greater relative force to complete task similar to young adults.

Healthy aging is associated with a reduction in the maximum capacity of structures to minimize motion through muscle force. With further decrements when muscles must contract quickly to make rapid adjustments for controlling undesired movements. It is important to acknowledge age-related differences in contraction type, or fatigue properties of muscles for they can confound the interpretation of results. When reviewing studies one should consider the contraction type and express load as a percentage of maximum force to allow for analysis on the demands of the task. The implications of these properties on movement performance will be considered in section 2.3.1 in this literature review.

2.1.3 The Neural System

Muscle fibers do not contract randomly, they require excitation (in the form of action potentials) generated by their respective motor neuron. The neural control system represents the numerous structures that have direct or indirect innervations with motor neurons that function to coordinate the level of tension of various muscles fibers at a specific point in time [Diedrichsen et al, 2010; McGill et al, 2003b]. Motor units are controlled by a combination of subcortical reflexes and cortical control working together to minimize the impact of external and internal forces applied to a joint [Goodworth and Peterka, 2009; Horak, 2006; Taube et al, 2008]. When a perturbation is applied to the body, receptors in muscles [Brown and McGill, 2009; Goodworth and Peterka, 2009; Granata et al, 2001; Granata and Rogers, 2007], and passive tissue (ligaments, joint capsules, and the annulus fiber of the vertebral disc) [Mohapatra et al, 2012; Sanchez-Zuriaga et al, 2010; Solomonow, 2009] can generate an inhibitory or excitatory reflex response to the motor units. Reflexes of muscles can also be generated from the visual or vestibular system [Goodworth and Peterka, 2009; Serrador et al, 2009]. When cortical structures detect undesired motion they can use weighted sensory afferent information (cutaneous, proprioceptive, vestibular, and visual) to recruit motor units with an appropriate timing and magnitude to achieve a movement goal [Goodworth and Peterka, 2009] or restore body position [Mohapatra et al, 2012; Santos et al, 2010]. These strategies to stabilize body position after the application of a perturbation are termed compensatory postural adjustments [Santos et al, 2010; Silfies et al, 2009a]. However the neural control system has a second strategy, known as anticipatory postural adjustments [Hodges and Richardson, 1999; Hodges, 2001; Jacobs et al, 2010; Mohapatra et al, 2012; Santos et al, 2010; Silfies et al, 2009a], where sensory information is used counter balance externally [Mohapatra et al, 2012; Santos et al, 2010], or internally generated forces [Hodges and Richardson, 1999; Hodges, 2001; Inglin and Woollacott, 1988; Silfies et al, 2009b; Woollacott and Manchester, 1993]. Using this information to adjust muscle tension prior to the application of force assist in minimizing joint motion [Ebenbichler et al, 2001; Goodworth and Peterka, 2009; Hodges, 2001]. This role of the nervous system to predict and respond to perturbations has a large implication related to joint control. For at any given time if the tension of the multiple muscles surrounding a joint is insufficient to counter an external moment or generate an undistributed joint load there is a risk of injury [Ebenbichler et al, 2001; McGill et al, 2003a].

The ability of the neural system to control the stability around a joint suggests that this system could potentially compensate for age-related changes in the active and passive systems. However, the neural control system also experiences age-related changes that may compromise its function. These changes are identified in the sensory afferents (visual [Black and Wood, 2005], vestibular [Serrador et al, 2009], and somatosensory system [Lacour et al, 2008]), the nerves that conduct action potentials, and the interneurons that participate in motor unit recruitment.

The cutanous system, part of the somatosensory system, represents sensors that identify touch and vibration applied to the skin. This information can be utilized to determine point of force application, joint movement [Verschueren, 2002] and provide feedback for fine motor skills [Kalisch et al, 2008]. Cutaneous sense declines with increasing age [Van Ooteghem et al, 2009; Woollacott and Manchester, 1993]. This decline in function occurs with a simultaneous reduction in the number and quality of mechanoreceptors in the skin responsible for detecting touch and vibration in older adults [Shaffer, 2007].

Proprioception, the knowledge of body position, and kinesthesia, knowledge of joint movement, both decline with aging. Although the cutaneous system contributes to these senses [Verschueren, 2002], other structures such as mechanoreceptors in the ligaments, joint capsules, tendons, and muscle spindles provide additional information [Ribeiro, 2007; Shaffer, 2007]. In older adults, there is a reduction in the number of mechanoreceptors in the joint capsule and ligaments compared to young adults [Goble, 2009]. However, age-related increases in ligament slack from joint space narrowing, also disrupts mechanoreceptor function as older adults would require greater strain before these structures experience similar tension to activate the mechanoreceptors. Muscle spindles change with aging, where increasing age is associated with lower number and quality of intrafusal and nuclear chain fibers [Shaffer, 2007]. Furthermore, older adults are believed to have reduced muscle spindle function within the trunk [Brumagne et al, 2004]. These changes account for the decreased ability for older adults to identify when movement begins, static position sense [Goble, 2009; Pijnappels et al, 2010b; Verschueren, 2002], and detect when a joint reaches a target angle, dynamic position sense [Madhavan and Shields, 2005]. However, dynamic position sense relies on both the quality of sensory receptors, and the conduction velocity of nerves that conduct these signals to allow individuals to perceive a target angle [Verschueren, 2002].

With increasing age there is a reduction in the conduction velocity of mixed nerve bundles [Fujimaki et al, 2009; Kido et al, 2004; Rivner et al, 2001]. Decreased nerve conduction velocity occurs with an age-related reduction in myelin thickness primarily in large nerves [Kalisch et al, 2008; Shaffer, 2007]. The diminished speed and quality of information provided by the somatosensory system (proprioceptive and cutaneous) result in greater variability in feedback provided to older adults making performance more variable, and potentially slowing motor learning in older adults [Christou, 2011; Van Ooteghem et al, 2009]. Combined these changes may contribute to the reduced reflex response and reduced position sense exhibited by older adults [Granacher et al, 2006; Kido et al, 2004a; Taube et al, 2008]. Reductions in joint position sense have been recognized in the trunk of older adults quantified as greater error in a repositioning to a target angle [Goldberg et al, 2005]. However, the ability to reproduce a target angle or the ability to detect when a target angle is reached depends on the cognitive recollection abilities of an individual [Goble et al, 2012].

With increasing age, neurons have smaller cell bodies and reduced dendrite length [Levin et al, 2011]. A change in neuron size and number partially account for the loss of cortical grey matter in older adults [Goble et al, 2010; Seidler et al, 2010]. Loss of grey matter occurs in all structures but is widespread in the frontal and parietal cortex [Gunning-Dixon et al, 2009; Kido et al, 2010]. In remaining neurons, changes in neurotransmitter secretion [Seidler et al, 2010], and altered neurotransmitter receptors result in reduced excitation and inhibition from synaptic transmission [Levin et al, 2011;

Yordanova et al, 2004]. Loss of action potential size and velocity occurs from a decline in white matter density and integrity in both the central [Gunning-Dixon et al, 2009; Van Impe et al, 2011a; Ward, 2006], and peripheral nervous system older adults [Shaffer, 2007]. These changes contribute to an impaired ability to inhibit spinal [Hortobagyi and Devita, 2006; Kido et al, 2004a; Obata et al, 2010] and cortical areas in older adults [Bernard and Seidler, 2012; Hortobagyi et al, 2006; McGregor et al, 2011]. Functionally all of the age-related changes in the neural system would result in a system that is slower to respond to less accurate feedback information. However, older adults have altered recruitment of cortical resources which may compensate for these age-related changes.

2.2 AGE-RELATED CHANGES IN NEURAL FUNCTION

The neural control system can compensate for age-related changes to the joint stability system by changing the recruitment of cortical structures. Functional magnetic resonance imaging (fMRI) and electroencephalography (EEG) studies confirm that older adults recruit more neurons in the motor cortex, pre-motor cortex and supplementary motor area when performing movement tasks [Hortobagyi and Devita, 2006; Seidler et al, 2010; Van Impe et al, 2011b; Yordanova et al, 2004]. This age-related “overactivation” of additional cortical areas is not just contralateral to the limb performing the movement but is also observed in the ipsilateral cortex [Goble et al, 2011; Yordanova et al, 2004]. Increased activation of the motor cortex may reflect a lack of inhibition of additional motor neurons [Bernard and Seidler, 2012; Hortobagyi et al, 2006; McGregor et al, 2011], and/or increased cortical contribution to activate motor units [Seidler et al, 2010]. Because of the somatotopic organization of the motor cortex [Kandel et al, 2000], diffuse recruitment of motor cortex could suggest increased activation of muscle synergist, and antagonist [Hortobagyi and Devita, 2006; Inuggi et al, 2011]. During movement, older adults have increased bilateral activation of the somatosensory cortex compared to young adults [Goble et al, 2010; Heuninckx et al, 2008; Van Impe et al, 2011b; Yordanova et al, 2004]. Increased activation of somatosensory cortex compensates for the decline in somatosensory information provided by sensory afferents. However, increased activity might also suggest that older adults shift their attention towards feedback information while performing movement task [Goble et al, 2010; Goble et al, 2011; Goble et al, 2012].

The activation of somatosensory and pre-motor cortex areas reflects an increase in cognitive control for planning movement in older adults [Inuggi et al, 2011]. This theory is supported by older adults having increased activation of areas in the frontal cortex, that are typically utilized for attention and planning [Goble et al, 2010; Goble et al, 2012; Heuninckx et al, 2008; Van Impe et al, 2011b]. Increased planning allows older adults to pre-activate the primary motor cortex to a greater degree than young adults [Inuggi et al, 2011]. This strategy is advantageous for older adult as it can gain motor cortex excitability before a voluntary movement speeding up simple reaction times [Inuggi et al, 2011; Levin et al, 2011; Yordanova et al, 2004].

Increased attention required by older adults to plan and perform movements is supported by dual-task experiments where one performs a motor task and a secondary mental operation simultaneously. In these experiments, older adults have reduced performance for either/both the cognitive and motor task [Goble et al, 2012; Van Impe et al, 2011b]. With the aforementioned decrease in cortical tissue with aging, older adults are faced with a predicament where increased cognitive demand must be met by dwindling cortical resources [Goble et al, 2010]. In task with increasing complexity older adults may not have enough cognitive resources to perform a movement to the same ability of younger adults [Lacour et al, 2008; Ward, 2006]. This theory is by a study that brain atrophy was associated with balance performance during a dual-task [Doi et al, 2011a].

2.3 AGE-RELATED CHANGES IN MOTOR AND MOVEMENT PERFORMANCE

The previous sections describe how age-related changes in tissues required for joint stability, and neural function can alter movement performance. The outcome of this interaction between physiological decline and altered neural control is the emergence of new motor patterns i.e. muscle activations, and altered movement patterns ie. joint kinematics and kinetics. Figure 1.2, provides a summary of the physiological and cortical function changes that occur with aging, which may alter motor performance. Though human motion tends to be highly individualized [Winter, 2009], systematic differences exist between younger and older adults.

In order to compare motor performance between younger and older adults a brief introduction to human movement is necessary. The forces used to generate human motion come from internal forces/ moments generated by muscles and other passive tissue surrounding a joint and external forces/ moments generated by: objects in contact with the body, body segment weight, and inter-segmental forces and moments [Winter, 2009]. Combined, these forces interact to produce and resist motion at a joint. Thus, human motion is partially controlled by the coordinated effort of muscles to generate internal forces to achieve a movement goal (including no motion) [Diedrichsen et al, 2010; McGill et al, 2003a]. Movement goals are achieved by using feedback information at both the cortical and spinal level, to plan, activate, monitor and make necessary adjustments to motor patterns with changing task demands [Brown and McGill, 2009; Goble et al, 2012; Granata and Rogers, 2007; Jacobs et al, 2010; Levin et al, 2011; Silfies et al, 2009b; Van Impe et al, 2011b].

Because of change in both physiological structures and cortical recruitment, one would expect age-related differences in coordination. Coordination has many definitions, however, two important types include: i) intra-joint coordination; how multiple muscles are recruited to control motion at a single joint [Arjmand et al, 2010; Rashedi et al, 2010], and ii) inter-joint coordination; how multiple joints interact in order to achieve a movement goal by complementing, resisting, or performing an unrelated task, to another joints during movement [Hodges, 2001; Silfies et al, 2009a; Woollacott and Manchester, 1993]. These types of coordination are dependent on one another. For example, intra-joint control is necessary to control the motion of the multiple joints used for inter-joint coordination. The primary focus of this literature review is how healthy aging contributes to altered motion control of the trunk; coordination will be used through this document representing intra-joint coordination, with any exceptions noted.

2.3.1 Confounders to Age-Related Changes in Movement and Motor Performance

When attempting to quantify group differences in coordination it is important to acknowledge numerous studies have identified that changing the intensity of a task has the capacity to modify movement performance. This section will analyze how movement performance changes with aging, but also how these parameters may be confounded by differences of relative task complexity between an old and young population.

2.3.1.1 Physical Task Intensity

Many studies compare older and younger adults performing movement tasks with similar external moments. When body weight is used as an external force, if two groups have similar weight and height distribution and perform the movement with similar range of motion and acceleration then external moments should be similar. However, age-related decreases in strength result in older adults using a higher percentage of their maximum force capacity to complete a task [Kuo et al, 2011; Madhavan et al, 2009; McGill et al, 1999; Nagai et al, 2011a;]. This represents a potential confounder when comparing muscle activity between older and younger adults, as intra-joint coordination of multiple muscles are depend on the relative intensity of the task demands [Brown and McGill, 2008b; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008; Song, 2004]. This is particularly true for antagonist that do not increase activity proportionally with agonist muscles [Brown and McGill, 2008b; Song, 2004]. Butler et al., identified that the amplitude of individual trunk muscles alter with an increasing external moment demands for both symmetric and asymmetric lift and replace tasks in younger adults [Butler et al, 2009a; Butler et al, 2010; Butler et al, 2010]. This altered muscle coordination also changes the onset times of multiple muscles (including the trunk muscles) in both younger and older adults [Bleuse et al, 2006; Brown et al, 1994; Carman et al, 1994]. It is important to acknowledge that increased task demands can change coordination when performing movements. Thus, attempts should be made to control relative task intensity as it can confound the interpretation of group differences in muscular coordination. Secondly, one should consider the long-term implications of increased task demands. For example a higher percent of relative muscle strength would result in localized muscle [Yassierli et al, 2007], and aerobic fatigue [Hortobgyi et al, 2011]. Just as a single loading even can alter trunk muscle coordination, fatigue is also recognized to alter the recruitment of trunk muscles [Granata et al, 2001; Sanchez-Zuriaga et al, 2010].

2.3.1.2 Movement Velocity

The speed of a movement effects the coordination both within and between joints. Compared to younger adults, older adults have a lower self-selected velocity for most movements [Gomez et al, 1991; Paquette et al, 2006]. Lower movement velocity is the result slower muscle contraction velocity in older adults [D'Antona et al, 2003]. But it may also represent a beneficial change, since movement at a lower velocity would minimize the perturbation applied to linked segments [Paquette et al, 2006] and increase time to react to feedback information [Madhavan et al, 2009; Verschueren, 2002]. When older and younger adults move at a speed that is faster than their self-selected velocity the timing and amplitude response of muscles surrounding a joint is altered [Bleuse et al, 2006; Hodges and Richardson, 1999; Woollacott and Manchester, 1993]. Movement above self selected velocity also increases the kinematic and kinetic variability in both older and younger adults [Barry et al, 2005]. Knowing that changes in coordination occur because of movements speeds that deviate from one’s self-selected velocity, any attempts to constrain movement speed may resulting in a confounding effect on muscle coordination as both groups would be moving at a different percentage of their self selected velocity.

2.3.1.3 Total Body Stability

Movement analysis studies have compared older and younger adults while performing both highly stable, for example producing ankle flexion in a seated position with all other joints constrained, and less stable (example standing) experimental task. The amount of stability that one has influences the coordination of a movement. In a highly stabilized seated ankle plantar flexion task older and younger adults had similar agonist-antagonist co-activation [Simoneau et al, 2009]. However, this relationship is dependent on the muscle tested for even in a stabilized task some muscles have increased agonist-antagonist co-activation in older adults [Bautmans et al, 2011; Klass et al, 2005;

Klass et al, 2007]. Most activities of daily living do not involve using a single joint in a seated position, in these task older adults have increased muscle co-activation relative to younger adults [Nagai et al, 2011a; Simoneau et al, 2008]. Furthermore, in tasks that are considered destabilizing such as stepping down and landing older adults increase agonist-antagonist co-activation to stiffen the joints in preparation to landing [Hoffren et al, 2007; Hortobagyi and DeVita, 2000]. While stability of an experimental task has a relationship with motor performance, this relationship may be confounded when participants have a perception of being unstable, which alters agonist-antagonist co-activation [Nagai et al, 2011b]. Regardless of the mechanism, it is important to consider the environment in which one performs movement, for if it is destabilizing it can influence the coordination differences observed between younger and older adults.

2.3.1.4 Number of Joints Necessary for Movement Goals

The number of joints involved to complete motor task effects coordination differences observed between younger and older adults. In a simple reaching task, using just the elbow, older adults had similar performance to younger adults. However, kinematic performance differed between these groups when multiple joints (elbow and shoulder) were necessary to reach to a target [Seidler et al, 2002]. The author concluded that increased agonist-antagonist co-activation in the single joint condition smoothed the reaching motion in older adults giving them similar kinematics as younger adults. However, in multiple joint movements older adults had a decrease in co-activation suggesting that they had difficulty coordinating additional joints. This reduction in control in multiple joint movements provides an explanation for studies that find older adults attempting to simplify a motor task by minimizing motion at specific joints typically by increasing agonist-antagonist co-activation around these “non mobile joints” [Carvalho et al, 2010; Kuo et al, 2011; Vernazza-Martin et al, 2008]. This feature would minimize the cortical demands required to plan for motion at multiple joints, and may be advantageous given that increasing age is associated with reduced control of joint motion through sub cortical reflexes [Granacher et al, 2006; Taube et al, 2008], which are used by younger adults [Kuo et al, 2011; Vernazza-Martin et al, 2008].

A secondary mechanism that older adults utilize to reduce the complexity of inter-joint coordination is moving joints with a similar pattern [Van Emmerik et al, 2005]. A study comparing older and younger adults performing a rapid head turn during gait found older adults coupled head turn rotation with trunk and pelvis rotation [Paquette et al, 2006]. Younger adults had out of phase pelvis and trunk rotation, rather coupling the pelvis with swing leg motion allowing for uninterrupted gait [Paquette et al, 2006]. This mechanism suggests that older adults attempt to move multiple joints in synchrony to minimize the complication of planning independent motion at multiple joints, but it may also reflect increased stiffness provided by agonist-antagonist co-activation causing older adult non mobile segments to move as a rigid body [Allum et al, 2002; Hatzitaki et al, 2005].

2.3.1.5 Heterogeneity in of Older Adults

The previous sections have highlighted that aging is associated with the decline in structures that make up the passive, active, and neural subsystems that are necessary to maintaining joint control. However, the rate of decline of these tissues is not a homogeneous process. Genetic and environmental factors interact with the rate of age-related changes in structures to attenuate or reverse some of these changes. For example, physical activity has been identified to increase bone mineral density [Guadalupe-Grau et al, 2009; Syed and Ng, 2010], ligament stiffness [Solomonow, 2009], muscle mass [Doherty, 2001a; Lang et al, 2010; Roig et al, 2010], muscle contractile properties [D'Antona et al, 2007], fatigue resistance [Kent-Braun, 2009], tendon stiffness [Narici et al, 2008], proprioceptive function [Ribeiro, 2007; Ribeiro and Oliveira, 2010], grey matter volume [Doi et al, 2011a; Gow et al, 2012], and white matter volume and integrity [Gunning-Dixon et al, 2009; McGregor et al, 2011]. Interestingly consistent with the theory that older adults use cortical mechanism to compensate for the decline in the structures involved in joint control fit older adults have similar cortical recruitment to younger adults exhibiting less frontal lobe and ipsilateral cortex involvement [McGregor et al, 2011] and more reflexive involvement [Granacher et al, 2006]. These features diverge from what is observed in sedentary older adults performing the same movement task. The combined effect of improved structures involved in joint control and reduced cortical compensation may explain why movement and motor performance of fit older adults are closer to those young adults than sedentary older adults [Carvalho et al, 2010; Clark et al, 2011; Hortobagyi and Devita, 2006]. Suggesting that studies comparing older and younger adults should attempt to quantify the physical activity levels of older adults.

2.3.2 Themes of Age-Related Changes in Motor and Movement Performance

The previous section has highlighted the numerous confounders exist when comparing movement performance between young and old adults. One must acknowledge age-related changes in coordination are partially explained by altered tissues, and cortical function but altered coordination also represents changes partially explained by differences in the relative complexity of a motor task in older adult participants. Despite the inclusion of the confounding differences from physical task intensity the literature identifies major themes on how motor performance changes with increasing age. These include increased agonist-antagonist co-activation, altered anticipatory postural adjustments, and changes in the spatial-temporal recruitment of multiple muscles. These changes may be the consequence of adaptive processes of how older adults utilize the neural control system to complete a task.

2.3.2.1 Antagonist Co-Activation

To complete a movement task many studies find that older adults have increased agonist-antagonist co-activation [Asaka and Wang, 2008; Carvalho et al, 2010; McGill et al, 1999; Nagai et al, 2011a]. Despite studies that claim controversy reporting, lower co-activation methodological differences including: non-normalized amplitude of EMG [Hoffren et al, 2007], different muscles tested [Bautmans et al, 2011; Klass et al, 2007], and increased stability with use of constrained joints [Simoneau et al, 2009] may account for these findings. Co-activation may result from deficits in the ability to inhibit antagonist activity at both the spinal and cortical level [Hortobagyi et al, 2006; Kido et al, 2004a; McGregor et al, 2011]. However, co-activation may also occur from diffusion in the motor cortex [Bernard and Seidler, 2012; Hortobagyi et al, 2009; Inuggi et al, 2011; Seidler et al, 2010] resulting in increased motor unit activation of desired as undesired muscles. Functionally increased agonist-antagonist co-activation can increase the stiffness at a joint to minimize angular displacement to external perturbations [Brown et al, 2006; Brown and McGill, 2009; Gardner-Morse and Stokes, 2001; Stokes et al, 2000a; Vera Garcia et al, 2007]. Increased agonist-antagonist co-activation may serve a secondary purpose to compensate for reduced proprioceptive sense [Benjuya et al, 2004]. In older adults, greater co-activation has been correlated with improved performance on an ankle dynamic position sense task [Madhavan and Shields, 2005].

2.3.2.2 Anticipatory Postural Adjustments

When performing a voluntary movement the body can anticipate destabilizing forces to pre-contract muscles to minimize the motion caused these forces [Bleuse et al, 2006; Cohen et al, 2011; Santos et al, 2010; Silfies et al, 2009a]. Studies comparing the response of the neural control of older and young adults identify that more cortical areas become active in older adults prior to the initiation of voluntary movement [Inuggi et al, 2011; Yordanova et al, 2004]. This would lead to the assumption that older adults have increased activation of multiple muscles in anticipation to destabilizing forces. The results of this hypothesis are mixed, studies that only compare the temporal aspect of anticipation postural adjustments find that older adults have a delayed onset latency of postural muscles relative to younger adults [Bleuse et al, 2006; Hwang et al, 2008; Woollacott and Manchester, 1993]. Yet other studies find when older adults anticipate an external force from foot strike, they increase the amplitude of muscle activity compared to younger adults to minimize motion induced from this perturbation [Hoffren et al, 2007; Hortobagyi and DeVita, 2000; Tsang et al, 2008]. The use of anticipatory co-activation prior to a voluntary movement has also been observed in the trunk of older adults [Asaka and Wang, 2008]. This gives an important implication in the outcome measures used to quantify coordination, if one were to only analyze temporal features (muscle onset) they would not capture the magnitude of muscle response. These studies suggest additional cortical resources used by older adults may result from increased for planning to generate anticipatory postural adjustments to stabilize a joint, rather than having to rely on short latency reflexes and rapid forces generation that declines with advancing age [Granacher et al, 2006; Kido et al, 2004a].

2.3.2.3 Spatial Changes in Motor Recruitment and Accessory (Out of Plane) Motions

While performing a movement task older adults have different motor and movement patterns compared to young adults. Two patterns have that have been discussed include increased co-activation, and altered anticipatory response of muscles. However, other altered patterns are noted when comparing coordination quantified by joint kinetics, kinematics, and electromyography. Older adults have an altered response of trunk muscles while performing the same movement task as younger adults [Hubley-Kozey et al, 2009b; McGill et al, 1999; Quirk and Hubley-Kozey, 2012]. In a study by McGill et al., older adults performed a dynamic range of motion task involving the trunk finding that older adults utilized different qualitative muscles activation patterns in both lateral bend and axial rotation compared to young adults. For the axial rotation task in particular older adults had increased activation of the Rectus Abdominus resulting in a coupled (flexion/rotation) motion compared to younger adults, who performed pure rotation [McGill et al, 1999]. The emergence of coupled movements have also been identified in gait of older adults suggesting older adults have more frontal and transverse plane trunk motion relative to younger adults [Doi et al, 2013; Van Emmerik et al, 2005]. These coupled motions are the result of additional motor activity required to meet task demands resulting in additional muscle recruitment, or an inability to inhibit muscle activation [Hortobagyi and Devita, 2006; Hortobagyi et al, 2006], the consequence of both result in the production of movements in undesired planes. However, the possibility exist that additional motor unit recruitment may represent a change in feedforward strategies or an inability to change ones motor pattern in response to a changing external moments.

Studies suggest that older adults utilize feedforward control to compensate for deteriorated feedback information, whereas younger adults that may set thresholds for reflex activity to assist in the stabilization of multiple joints during motion [Vernazza-Martin et al, 2008]. For example, a qualitative assessment of EMG showed that older adults adults utilized increased abdominal activity near the end of a deep flexion task to increase forward flexion whereas younger adults deactivated their back extensor muscles allowing gravity to move their trunk mass [McGill et al, 1999]. Older adults using feedforward control during voluntary movements is further supported by evidence that older adults utilize more primary and pre-motor cortex relative to young adults [Van Impe et al, 2011b] with more constant activation of these areas during the entire movement task [Inuggi et al, 2011], this cortical activation suggesting that older adults use more cortical control to activate motor neurons. However, increased feedforward control may result in inefficient movement patterns that do not accurately respond to the external moment demands of the task. Using a controlled dynamic leg loading exercise Hubley-Kozey et al. found that while younger adults utilized muscle activation patterns that corresponded with changing external moments, older adults had more constant activation (amplitude over time) of specific abdominal sites during this same time period [Hubley-Kozey et al, 2009a]. The authors believed that this increased activation was the result of an inability to relax abdominal muscles during the movement to maintain trunk stability rather than relying on proprioceptive feedback to alter muscle activation patterns [Hubley-Kozey et al, 2009a]. Increased burst duration of muscles have also been identified in gait for older adults at the knee and ankle, with these authors also believing that older adults utilize this pattern to increase joint stability [Hortobagyi et al, 2009].

Although older adults use more feedforward control they also attempt to make use of feedback control during movement task. The concept of increased reliance of cortical feedback is consistent with studies that find older adults have increased jerk and burst in muscle activation suggesting that older adults attempt to make multiple corrections to their movement pattern [Barry et al, 2005]. Increased reliance on feedback control is supported at the cortical level by the increase in somatosensory cortex activation in older adults when performing voluntary movements [Goble et al, 2011; Heuninckx et al, 2008; Van Impe et al, 2011b; Ward, 2006].

2.4 FACTORS THAT IMPAIR THE USE OF AGE-RELATED COMPENSATORY MECHANISMS

Older adults utilize greater cortical resources in order to plan for and execute a movement. These resources in turn may be linked with changes in movement performance. However, for some movements the use of these cortical strategies can be impaired. In particular unanticipated movements, choice reaction time experiments, and dual-task experiments have the potential to alter age-related movement performance by minimizing the use of compensatory cortical control.

2.4.1 Unanticipated Movement caused by External Loads

In unanticipated movements, movement is generated by the application unexpected external force [Santos et al, 2010] or a change to ones base of support [Allum et al, 2002]. These movements often occur in the form of slips, trips, or bumps [Horak, 2006; Tang and Woollacott, 1999] and are corrected for by compensatory postural adjustments where one must sense a joint perturbation and then react to it via reflexes and cortical control [Mohapatra et al, 2012; Santos et al, 2010]. As older adults have reduced reflex response [Granacher et al, 2006; Obata et al, 2010], deteriorated sensory information [Shaffer, 2007] and slower nerve conduction velocity [Rivner et al, 2001] they have poor performance to unanticipated movements compared to younger adults. Using unanticipated perturbations older adults have increased center of mass and trunk motion relative to younger adults [Allum et al, 2002; Bugnariu and Sveistrup, 2006; Grabiner et al, 2008a; Van Ooteghem et al, 2009]. These movement patterns are further supported by older adults having latter onset latency of postural muscles in response to externally induced motion [Allum et al, 2002; de Freitas et al, 2010; Hwang et al, 2008].

This would suggest that for involuntary movement the compensatory mechanisms used by older adults have a reduced efficacy to control undesired movement. However, perturbation studies also show that older adults attempt to use compensatory mechanisms to maintain balance. Older adults may rely on long latency reflexes (cortically mediated) in muscles to correct for unanticipated movements [Santos et al, 2010]. This feature is evident with older adults typically responding to unanticipated movements with a prolonged burst duration to restore balance [Bugnariu and Sveistrup, 2006; de Freitas et al, 2010; Tang and Woollacott, 1999] and a higher amplitude of EMG activity for these later responses [Allum et al, 2002; Madhavan et al, 2009]. A second mechanism utilized by older adults in an experimental setting is increased joint co-activation in anticipation to a laboratory induced perturbation. In one study by Hwang et al, compared anticipated (allowing vision) and unanticipated trunk perturbations, however for older adults, he found no difference in CoM motion. The amplitude response of EMG was not measured in this study it was identified that the shift from anticipated to unanticipated paradigms did change the onset latency of erector spinea muscles in both younger and older adults [Hwang et al, 2008]. It can be speculated that the little change in CoM motion from the anticipatory to no anticipatory condition was the consequence of increased trunk co-activation to minimize joint motion in response to an external perturbation [Brown et al, 2006; Granata et al, 2001; Stokes et al, 2000a].

2.4.2 Movement Preparation and Choice Reactions

Older adults have increased planning before initiation a movement evident by increased pre-motor [Inuggi et al, 2011; Seidler et al, 2010], and motor cortex activity before movement onset for simple reaction time experiments [Levin et al, 2011; Yordanova et al, 2004]. However, increased motor planning can be reduced by introducing a choice reaction paradigm, where a participant is uncertain what motor plan to use. Compared to simple reaction times, choices reaction time result in a slower movement and motor reaction time in both younger and older adults [Brauer and Burns, 2002; Cohen et al, 2011; Hodges, 2001; Luchies et al, 2002; Tomita et al, 2011]. In choice reaction time experiments older adults have a greater increase in choice relative to simple reaction times suggesting an increases the reliance on motor planning before starting a movement when there is certainty [Pijnappels et al, 2010a]. Reduced preparation not only alters reaction time, choice reaction time conditions have also alter movement time in older but not younger adults [Brauer and Burns, 2002; Luchies et al, 2002] suggesting potential changes in movement patterns. It remains unknown if motor patterns change in choice reaction time paradigms in older adults. Brauer et al., determined that both older and younger adults had changes in the onset times of postural muscles, including the trunk, relative to movement onset in a choice leg step in high and low preparation conditions. However, no groups by preparation interactions were identified suggesting that both groups were influenced equally by the loss of preparation [Brauer and Burns, 2002]. Despite this inconsistency other authors have suggested that choice reactions may be better at discriminating motor performance between younger and older adults [Yordanova et al, 2004], providing additional information to identify older adults with a fall risk [Pijnappels et al, 2010a].

2.4.3 Minimizing Cortical Resources Available for Motor Performance

Older adults not only utilize increased cortical activity to plan and initiate movement, they also have increased cortical activation to monitor movement performance [Goble et al, 2010; Seidler et al, 2010; Ward, 2006]. Increased cortical activity used by older adults represents feedforward control of movement and increased attention to movement performance [Goble et al, 2012; Van Impe et al, 2011b]. The role of attention in movement performance has been speculated by the recruitment of specific cortical areas in the frontal cortex [Van Impe et al, 2011b], and changes in motor task performance during in dual-task experiments. The theory behind dual-task experiments is that both the cognitive and motor task must compete for cortical resources. During a dual-task experiment performance of both the motor and cognitive task exhibit a U shape relationship where if one has to much focus on a single task it begins to deteriorate as it loses automaticity. At the other extreme if two difficult task are performed simultaneously competition for cortical resources can decline the performance of either or both tasks [Huxhold et al, 2006; Lacour et al, 2008]. This competition theory is supported with MRI studies finding an overlap in cortical areas involved in a movement task and a cognitive mathematical task [Van Impe et al, 2011b]. Expanding dual-task into older adults their increased utilization of cortical resources during movement performance combined with their decrease in cortical structure result in older adults, compared to younger adults, having a lower threshold where dual-task performance can deteriorate motor performance [Huxhold et al, 2006; Laessoe and Voigt, 2008; Rankin et al, 2000; Simoneau et al, 2008]. Consistent with the suggestion that these changes are associated with loss of cortical resources studies have found that older adults with greater brain atrophy [Doi et al, 2011a] and lower working memory [Goble et al, 2012] experience greater change in motor and cognitive task performance indual-task conditions.

Perhaps the most interesting finding from dual-task studies is the reversal of specific compensatory strategies utilized by older adults to increase joint stiffness (Section 2.3.2). In a dual-task paradigm compared to a single task older adults had altered neuromuscular function including: decreased agonist-antagonist co-activation of peripheral joints [Lacour et al, 2008; Rankin et al, 2000; Simoneau et al, 2008; Tsang et al, 2008], and reductions in anticipatory co-activation [Tsang et al, 2008]. These findings suggest that the motor patterns observed in older adults may be the result of cortical involvement in motor performance.

2.5 METHODS FOR QUANTIFY AGE-RELATED CHANGES IN MOTOR PERFORMANCE

The preceding sections have identified the age-related declines in physiology, but these declines may be compensated for by differences in neural control that are also associated with the aging process. In turn, these change affect the motor and movement patterns used by older adults. These previous sections have identified that comparing age groups must be interpreted with caution for many factors may skew group differences. At this point it should be acknowledged that there are multiple ways to quantify age-related changes in movement including, but not limited to: measuring joint movement represented by the levels of joint kinetics and kinematics, a neuromotor interpretation of EMG, or capturing changes at the cortical level using fMRI or EEG. This section will focus on how to quantify age-related changes in coordination for voluntary movements, the techniques that could be used including their potential strengths and weaknesses, and what the analysis techniques that can be used to quantify coordination.

2.5.1 Kinematic and Kinetic Analysis

Multiple studies have compared older and younger adults using joint kinematics, and kinetics (using the joint kinematic data in an inverse dynamic approach). This technique is valuable for it can compare how one utilizes different ranges of motion at multiple joints [Kuo et al, 2011; McGill et al, 1999; Van Emmerik et al, 2005; Vernazza-Martin et al, 2008], their velocity [Marras, 1994; Paquette et al, 2006], and the number of changes in direction one makes [Seidler et al, 2002] in order to produce a movement or maintain posture. Joint kinematics and kinetics can be used to compare how multiple joints move synchronously in order to achieve a movement goal [Burgess et al, 2009; Paquette et al, 2006; Van Emmerik et al, 2005]. However, to analyze coordination, joint kinematics and kinetics are not without their limitations. Analysis of a single joint permits motion or forces to be quantified in only 6 degrees of freedom (3 linear, and 3 rotational), thus limiting the redundancy of the multiple forces that exist around the joint to be summed as a net moment or force [Winter, 2009]. This approach is limited, as the net moment does not consider agonist-antagonist co-activation, as the moments produced by these opposing muscles would cancel one another out. Losing this information limits interpretation of forces generated by co-activation that correspond with the loading and stiffness of a joint [Brown and McGill, 2008b; Hansen et al, 2006; Simoneau et al, 2009; Vera Garcia et al, 2007]. A second limitation from the reduced degrees of freedom provided by joint kinematics or kinetics, is that a motion or joint moment can be produced by multiple combinations of muscle activities [Davis and Mirka, 2000; Rashedi et al, 2010]. Therefore, a similar net moment could be achieved using entirely different muscles, passive tissues, or momentum properties of linked segments. To quantify differences in coordination with respect to age related changes, the fewer degrees of freedoms one has to interpret may lead to false-negative results where one considers that movement patterns of older adults is no different than young adults, despite using different motor patterns [Hubley-Kozey et al, 2009a; Seidler et al, 2002].

2.5.2 Neuroimaging

Neuroimaging can include multiple techniques that can quantify chances in the electromagnetic properties or oxygen consumption of cortical areas with respect to time. Electroencephalography, fMRI, and transcranial magnetic stimulation (TMS) have been used to study age-related differences in the recruitment of various cortical structures [Van Impe et al, 2011b; Yordanova et al, 2004]. Using these technique one can identify how the cortex plans, initiates, and terminates movements [Inuggi et al, 2011; Ward, 2006]. Determining which areas of the central nervous system are used for these functions, and to what extent they are recruited during voluntary movements [Goble et al, 2010; Van Impe et al, 2011b]. All of these interpretations depend on the organization of the cortex corresponding to regions of interest [Kandel et al, 2000]. Within specific cortical structures such as the primary motor cortex the concept of a homunculus exist where through somatotopic organization each region represents a particular area of the body [Tsao et al, 2011a]. This representation can change with time from neuroplasticity [Tsao et al, 2010; Tsao et al, 2011b; Tsao et al, 2011c]. Thus, the knowledge of which cortical structures are recruited during movement do not necessarily coincide with which motor neurons are activated during movement, what amplitude they are recruited, or how they interact with agonist, antagonist, and synergist muscles at a joint level [Inuggi et al, 2011]. This problem is further compounded when one considers that the alpha motor neurons activated by the primary motor cortex are also innervated by sub cortical structures [Hortobagyi and Devita, 2006; Kandel et al, 2000; Kido et al, 2004a]. This modulation in activity could not be quantified by simply imaging the cortex of younger and older adults and therefore would be make it difficult to interpret if differences in cortical activation are representative of altered motor patterns.

2.5.3 Electromyography

EMG is the sum of the muscle action potentials generated across muscle fibers within the capture volume of an electrode referred to as an interference signal [Farina et al, 2010; Staudenmann et al, 2010]. The capture volume of surface electrodes is determined by the size of the conductive surface [Fuglevand et al, 1992; Staudenmann et al, 2010; U.S. Dept of Health and Human Services, 1992], interelectrode spacing [Fuglevand et al, 1992; Winter, 2009], and the amount and conductivity of tissue below the electrode site [Burden, 2010b; Hug, 2011; Solomonow et al, 1994]. This technique captures the sum of cortical and spinal excitement to motor units and provides detail of the motor patterns that are used to recruit muscles that contribute to the movement pattern observed in joint kinematics and kinetics [Staudenmann et al, 2010; U.S. Dept of Health and Human Services, 1992; Winter, 2009]. However, given that muscle force depends on multiple parameters the relationship between EMG and force is inconsistent. Signal amplitude are altered by changes in the length [Alkner et al, 2000; Hansen et al, 2006; Mohamed et al, 2002; Sheikhzadeh et al, 2008], velocity [Arjmand et al, 2010; Burden, 2010a; Davis and Mirka, 2000; Gagnon et al, 2011], and fatigue [Farina et al, 2010; Hug, 2011; Sanchez-Zuriaga et al, 2010; Yassierli et al, 2007] of muscle fibers. If these factors are controlled the analysis of motor patterns may allow for some interpretation on how maximum normalized EMG is associated with external loads [Brown and McGill, 2008b]. To effectively use electromyography one must be aware of methods necessary to maximize the interpretation of signals, and the limitations, and reliability of these signals.

2.5.3.1 Normalization of Electromyographic Signals

A limitation of electromyography is that it records the voltage change within the capture volume of the electrode, therefore voltage is highly dependent on the resistance provided by the tissue between the electrode site and the muscle fibers it is capturing [Fuglevand et al, 1992; Solomonow et al, 1994; U.S. Dept of Health and Human Services, 1992]. Given this relationship, there is a high variability in the amount of tissue between the skin surface and the muscle which makes interpretation of the signal amplitude and frequency content difficult between individuals and various electrode sites [Farina et al, 2010; Staudenmann et al, 2010]. To account for this feature one can perform a maximum voluntary contraction where a participant attempts to recruit all motor units by performing a standardized task [Burden, 2010b]. By normalizing values of EMG one can make comparisons individuals and muscle sites by EMG voltage as a percentage of maximum. The validity of this technique is questioned with many researchers believing that some populations may have a difficulty maximally recruiting all motor units. However, absolute EMG measures have limited interpretation due to their poor reliability. Thus, guidelines suggest that some normalization protocol be performed [Farina et al, 2010]. A recent literature review of normalization techniques has established that isometric MVC normalizations have good to excellent test-retest reliability [Burden, 2010b]. MVC are feasible for older adults [Hanada et al, 2008; Hanada et al, 2011; Hubley-Kozey et al, 2009a; McGill et al, 1999] and both healthy younger and older adults can maximally recruit a comparable number of motor units in maximum normalization task, with improved performance if a training session is provided [Klass et al, 2007].

2.5.3.2 Cross Talk as Potential Signal Contamination

Using surface electromyography one must be aware of signal crosstalk. Electrical activity within a muscle scatters as it is conducted through non excitable tissue. For this reason, greater amounts of tissue between the surface electrode and muscle border can cause increased signal cross talk, recording of EMG voltage changes from adjacent muscles [Hug, 2011; Solomonow et al, 1994]. Crosstalk can be minimized with proper application of surface EMG, electrode size, and interelectrode distance can minimize the pickup volume of a surface electrode providing some protection from signal contamination [Fuglevand et al, 1992]. Placement of surface electrodes such that it is at a maximal distance from the boundaries of adjacent muscles minimizes their contribution to EMG signal [U.S. Dept of Health and Human Services, 1992; Winter et al, 1994]. Once electrodes are placed, manual muscle testing can identify crosstalk if EMG activity is recorded from a muscle that would not produce a particular moment [Winter, 2009; Winter and Yack, 1987]. However, this technique has limited validity for synergistic muscle sites, which would co-activate to a manual muscle test because of their similar lines of action.

Controversy exist on how much signal contamination exist due to cross talk, Solomonow et al. cut the innervations to multiple muscles and to then stimulate motor units within in a cat, to determine that cross talk would only contributed to 5% of a surface EMG signal of adjacent muscles [Solomonow et al, 1994]. This is consistent with Winter's theoretical level of 3% cross talk if electrodes are positioned 3cm lateral to the border of an adjacent muscle [Winter et al, 1994]. With this model predicting that at 2 cm one would share 13% common signal, and experimentally this distance had 20% common signal assessed by cross-correlation techniques of raw EMG. This suggests that a small contribution of adjacent muscle activity would have an impact on the recorded muscle signal and may influence the EMG pattern of a desired muscle. However, cross-talk is most influential if the signal of the muscle of interest is silent, for example Davis et al., identified that surface electrodes of the ipsilateral external oblique muscles recorded activity from the underlying deep internal oblique muscles during a rotation exercise [Davis and Mirka, 2000]. If crosstalk exists within muscles, one may overestimate the degree of muscle co-activity in adjacent muscles. This is a potential problem in surface EMG studies particularly in muscles with low activation.

2.5.3.3 Reliability of Electromyography

Despite the limitations involved with surface EMG, amplitude measures have good to excellent within and between day reliability [Dankaerts et al, 2004; Lariviere et al, 2000b; Ng et al, 2003; Perez and Nussbaum, 2002] with intraclass correlation (ICC) ranging from .5-.99. Although some low ICC measures are reported, one must consider these studies use different loading intensities including maximal [Ng et al, 2003], and various levels of submaximal loading [Lariviere et al, 2000b] which can influence the reliability of sEMG measures. Secondly the method of resistance application can influence the reliability of sEMG measures as manual resistance [Dankaerts et al, 2004] has lower reliability than using restraints to stabilize the joint during contractions [Ng et al, 2003]. Finally, the direction of force application influences reliability as some muscles work as an agonist in some scenarios but an antagonist in others [Perez and Nussbaum, 2002]. This would reduce the reliability of EMG measurements for muscle activations are most reliable when the muscle acts as an agonist [Lariviere et al, 2000b]. Not only are the amplitude measures obtained from sEMG reliable but it is also determined that temporal measures such as onsets [Silfies et al, 2009a; Silfies et al, 2009b] and temporal waveforms [Lariviere et al, 2000a] also have good to excellent reliability.

2.6 DESIGNING AN EXPERIMENT TO CAPTURE AGE-RELATED CHANGES IN MOTOR PERFORMANCE WHILE MINIMIZING CONFOUNDERS

Ideally joint kinematics, EMG, and neural imaging could all be collected during human movement task to compare older versus younger populations. However, if one can ensure that the movement pattern performed by younger and older adults are similar EMG best quantifies alterations in coordination for captures the ability of the motor system to exploit muscle redundancies and in part measures the end product of cortical and spinal mechanisms. The additional information provided by neural imaging could determine if altered motor pattern observed in these populations was correlated with an altered cortical mechanism, which can only be speculated using EMG. However, experimental tasks can be designed to work within the limitations associated with EMG, and may have the potential to capture age-related changes in the use of cortical compensatory mechanisms.

If one is using sEMG to compare movement performance the task should be both isometric, to eliminate the confounding changes in EMG activity required to compensate for altered force-length [Brown and McGill, 2008b] and force-velocity [Klass et al, 2005] relationships. This isometric task should be centered on the neutral zone of a joint minimizing the involvement of passive tissue to maintaining joint stability [Cholewicki and VanVliet, 2002; Panjabi, 1992b]. One study suggests that the outcome measures of muscle model would be valid even if passive stiffness was not considered, as long as the spine was in a neutral position [Kavcic et al, 2004]. Therefore, this would minimize the interference of age-related changes of force production contributed by the passive system.

If one wished to determine what age-related changes are accounted for by changes in cortical compensatory a movement task should be designed to include two constraints. First, one should attempt to minimize the influence of subcortical reflex contribution to motor unit recruitment. This goal can be achieved by reducing the magnitude of perturbations experienced during voluntary movement, and reducing the velocity and external load of the movement [Brown and McGill, 2009; Goodworth and Peterka, 2009; Stokes et al, 2000a]. Secondly if one wished to identify how age-related changes in cortical compensation influence movement control one could design a task to interfere with the ability to utilize cortical resources. One common paradigm is a dual-task where one concurrently performs a mental operation while performing a movement task. It is possible that observing the changes in muscle activation patterns during a dual-task could elucidate how aging may change the use of cortical resources for movement task.

If age-related changes in motor control alter intra-joint coordination patterns, it would be important to analyze muscle activation patterns in a joint with numerous mechanical redundancies. This would add a spatial complexity to EMG to capture potential group differences across multiple muscle fibers. The trunk is composed of multiple muscles each with unique fiber orientations that can produce rotations around all 3 anatomical planes [Ebenbichler et al, 2001; Hansen et al, 2006; Hansen et al, 2006; Jorgensen, 2001; Marras, 2001; Rashedi et al, 2010]. These muscles are identified to have unique innervations [Tong et al, 2005] and cortical representation within the primary motor cortex [Tsao et al, 2010; Tsao et al, 2011a]. Furthermore, within the multiple muscle groups and at different spinal levels specific fiber bundled of the trunk muscles have unique lines of action, cross sectional areas, and effective moment arm lengths [Davis and Mirka, 2000; Gagnon et al, 2011; Gatton et al, 2011; Stokes et al, 2011; Urquhart et al, 2005]. Thus, the muscles of the trunk and their individual fibers have differential recruitment to external moments that best utilize their line of action and relative cross sectional areas [Butler et al, 2010; Hanada et al, 2011; Hubley-Kozey et al, 2009b; Hubley-Kozey et al, 2010; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008; Talebian et al, 2010; Urquhart and Hodges, 2005]. It is important to acknowledge that because of the multi rotational moment production capability of the trunk muscles few trunk muscles have a direct role of producing a moment in any pure plane [Hu et al, 2011] but rather their activations are summed to produce a moment [Rashedi et al, 2010]. Because, of this relationship many authors conclude that no one trunk muscle contributes to spinal stability [Cholewicki and VanVliet, 2002; Kavcic et al, 2004; Stokes et al, 2011] and therefore the trunk is misrepresented in studies that analyze too few trunk muscles assuming synergistic activity [Allum et al, 2002; Asaka and Wang, 2008; Hwang et al, 2008; Kuo et al, 2011; Vernazza-Martin et al, 2008].

Perhaps more interesting than understanding how trunk muscles respond to constant external moment [Kavcic et al, 2004; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008; Talebian et al, 2010], is how trunk muscle activation patterns change during a dynamic perturbation task. Some work exist that analyze the trunk muscle response during dynamic loading task or during a range of motion task [Butler et al, 2010; Hu et al, 2011; Lariviere et al, 2000a; Lariviere et al, 2000b; McGill et al, 1999]. Dynamic loading can be achieved by continually changing the external moments generated around the trunk. For example our lab changes the external moment experienced by the spine by changing the position of one’s arms [Hubley-Kozey et al, 2012] or legs [Hubley-Kozey et al, 2010] while participants are instructed to keep a neutral spine. These studies show how muscles increase and decrease their activation with respect to the changing moment demands generated by limb movement [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2010; Hubley-Kozey et al, 2012]. Furthermore, using this paradigm been shown to discriminate between older and younger adults in their capacity to modulate muscle activity in response to a dynamic leg loading task [Hubley-Kozey et al, 2009a]. Since these muscle activation patterns can be generated by feedforward and feedback information during the continually changing moment it may be a better task to discriminate how older adults utilize compensatory mechanisms to modulate activity of trunk muscles, which would not be characterized if a movement goal only required constant force generation in a single vector.

2.7 METHODS TO QUANTIFY SPATIAL TEMPORAL DIFFERENCE IN MOTOR PERFORMANCE

Although a task can be designed to capture age-related changes in trunk muscle activation patterns. One must determine what outcome measure should be analyzed to quantify the response of numerous trunk muscles. Electromyography measures can be parameterized to simple temporal measures of including: muscles onset, offset, and burst duration [Allum et al, 2002; Brauer and Burns, 2002; Hodges, 2001; Hu et al, 2011; Hwang et al, 2008; Kumar, 2010; Mehta et al, 2010; Silfies et al, 2009a; Silfies et al, 2009b; Stokes et al, 2000a; Tang and Woollacott, 1999; Woollacott and Manchester, 1993], and amplitude measures including: mean, and peak EMG amplitude over an entire movement task [Carvalho et al, 2010; Kang and Dingwell, 2009; Madhavan et al, 2009; Perez and Nussbaum, 2002]. Using these measures authors often quantify co-activation as a phase where muscle sites are simultaneously activated [Hortobagyi et al, 2009; Mehta et al, 2010] or the ratio of the amplitude of two muscles or a group of muscles during a task [Bautmans et al, 2011; D'hooge et al, 2012; Hoffren et al, 2007; Hoffren et al, 2011; Hortobagyi et al, 2009; Nagai et al, 2011a; Perez and Nussbaum, 2002; Song, 2004]. Such measures may be appropriate for experimental task that require constant muscle activation. However, if muscle activation were to change with respect to changing joint moment demands this parameterization could mask age-related changes in the temporal pattern of muscle activation patterns. These features can be characterized by breaking down a movement into multiple events or phases thereby providing a representation of the mean or peak amplitude within these phases [Butler et al, 2009a; Butler et al, 2009b; Butler et al, 2010; Hanada et al, 2011; Hoffren et al, 2007; Hoffren et al, 2011; Kuo et al, 2011]. However this method is still unable to capture if muscle activation are slowly increasing, sustained, or have multiple burst of corrective feedback control throughout the phase [Barry et al, 2005; Clark and Fielding, 2011; Hubley-Kozey et al, 2009a; McGill et al, 1999].

Previous studies using a discrete parameter to quantify muscle activation patterns, and the analysis of too few muscles would be a method of data reduction. However, there are a growing number of techniques that can be used to reduce the complexity of multidimensional data. These methods include neural networks [Chen et al, 2007; Schollhorn, 2004], Fourier transformations [Chen et al, 2007; Hu et al, 2011; Zhang et al, 1991], auto correlations [Asaka and Wang, 2008; Hu et al, 2011; Nelson-Wong et al, 2009], wavelet transformations [Chen et al, 2007], and principal component analysis [de Freitas et al, 2010; Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2010; Hubley-Kozey et al, 2012; Lariviere et al, 2000a; Vernazza-Martin et al, 2008]. All of these techniques are mathematical representations that are interpreted with a physiological meaning, resulting in various limitations. Neural networks represent an input pattern as a weighted average of nodes allowing classification and reconstruction of a signal. However, for this to work one requires a large database for the learning phase. Once a neural network is established the outcome measure whether it be parametric or non-parametric provided the limited information to where this value came from. Giving neural networks a "black box" property for those that do not what comparisons were made in the algorithm to generate the outcome measure [Schollhorn, 2004]. Wavelet and Fourier transformation are similar to one another as they decomposes a signal into a single equation represented by a series of sine waves frequencies with various coefficients [Chen et al, 2007; Zhang et al, 1991]. This method of data reduction is not readily compared with statistical analysis but instead one will remove a parameter such as mean or median frequency, or give the coefficients to a non-parametric technique such as cluster analysis [Zhang et al, 1991] or neural networks [Chen et al, 2007] to analyze these coefficients making it difficult to make between or within participant comparisons. A second problem with these techniques are they depend on a periodic signal to identify the frequencies to do this the linear envelope during a discrete movement must be extrapolated by repeating the waveform or adding multiple trials in series [Zhang et al, 1991]. Cross correlation is able to determine if a time series of points has a time shift and how well the waveforms are correlated. The benefit of this measure is one can compare the waveforms statistically using the correlation coefficient output in this method. However, this technique is problematic for interpreting multiple muscles would require numerous cross-correlation analysis, for example if one analyzed 24 muscles one would need to make about 300 comparisons to determine the relationship between all muscle sites. Additional constraints to this technique include the feature that one can only determine how related two waveforms are but not what divergent features exist between these waveforms. Secondly, the correlation coefficient is highly dependent on the time shift function in cross correlation therefore, one must identify if the time shift represent a physiologically acceptable phenomena [Nelson-Wong et al, 2009]. Finally, cross-correlation technique only recognizes if muscle activation patterns move in the same direction but ignores the amplitude feature of compared EMG waveforms.

2.7.1 Principal Component Analysis

Principal Component Analysis (PCA) represents a method of data reduction that can overcome some of the limitations mentioned in other techniques. This technique uses a data matrix formed from the original EMG waveform data, to create an orthogonal covariance matrix that can be used in eigenvector decomposition. From this procedure a transformation matrix is produced that contains the eigenvectors which define a new axis onto which the original EMG waveforms can be transformed. The new axis defined by the eigenvectors represent the principal component of the waveform that best captures the variance within the original waveforms [Jackson, 2003]. This stage would make PCA similar to cross-correlation, but PCA has the additional advantage of extracting additional waveform variance that would represent the residual of the subsequent waveform increasing the ability to interpret and reconstruct data. PCA removes the most explained variance within a series of waveforms, thus by only including PCs that explain up to 1% of the total variance PCA also functions as a filter removing variance contributed by noisy EMG data [Hubley-Kozey et al, 2006]. This function potentially increases the reliability of EMG waveforms when using PCA analysis [Lariviere et al, 2000a]. An additional advantage of PCA is that by transforming an individual waveform onto the principal component results in a single number for each EMG waveform PC known as a PC score. These scores identify how similar individual waveforms are and can be used in statistical techniques [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012]. Thus PCA is powerful in that it can remove the large number of data points that exist in a temporal

EMG waveform. Secondly, it is able to identify muscle synergies and group differences, since similar PC score would suggest that the EMG waveform shares a common feature [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012]. The limitation of this technique is that waveform data must be normalized to the same number of data points, which removes the ability to discriminate movement timing. However, this data can be captured and compared using discrete measures. PCA may represent an ideal way to quantify muscle synergies and has been used in the past to quantify synergies in the trunk for both younger [Butler et al, 2009b; Butler et al, 2010; Butler et al, 2010; Hu et al, 2011; Hubley-Kozey et al, 2012], and older adults [Asaka and Wang, 2008; de Freitas et al, 2010; Hubley-Kozey et al, 2009; Vernazza-Martin et al, 2008].

2.8 GENERAL SUMMARY

Aging is associated with changes in the structure and function of multiple tissues that influence the passive, active, and neural subsystems recognized to control motion at a joint. However, these physiological changes may be compensated for by altered neural control of the skeletal musculature in older populations. To date it remains unclear how age-related changes in human movement are a consequence of altered physiology, neural compensation, or explained by numerous confounded explained by differences in movement complexity. In particular, although research identifies trunk control as an important predictor of balance performance in the older adult, only limited research has investigated if neuromuscular performance, in particular muscle synergies of the trunk musculature are altered as a function of advanced age. This review illustrates that this gap in our knowledge may be an important factor on the age-related increase in fall and musculoskeletal injury risk. Throughout this review of the literature gaps in the current knowledge provided guidance toward an experimental design capture age-related changes in neuromuscular activation patterns, while minimizing the risk of known confounders. Furthermore, methods of data collection and data reduction were reviewed so that the spatial-temporal complexity of the trunk neuromuscular system could be captured, while using a method of data reduction to make these data manageable.

CHAPTER 3 GENERAL METHODOLOGY

3.1 GENERAL PARTICIPANT RECRUITMENT

A cross-sectional design was used, to address the primary objective of the study and sub objective 1. Two groups of participants were defined: i) younger adult (20-45) and ii) an older adult (65+) group. All of the young participants and 60% of the older adults were part of a larger related study. The remaining older participants were recruited for the experimental study that formed sub objective 2. In all cases participants were excluded if they reported incidence of low back pain in the past 12-months, previous abdominal or back surgery, spinal fractures, major musculoskeletal injury, cardio-respiratory or neurological conditions.

Participants were recruited by web advertisement, posters, and notices around Dalhousie University and local gyms. All participants expressing interest in the study underwent a telephone interview including a health screening questionnaire to determine his or her eligibility for the study. This questionnaire was designed to eliminate participants based on the exclusion criteria (experiencing falls or low back pain within the last 5 or 1 year respectively, surgical procedures leaving abdominal scaring, surgical procedures to the spine, and the presence of any neurological, cognitive, musculoskeletal, or cardiorespriatory conditions) that could affect study results or put participants at risk. Additional exclusion criteria older adults underwent a postural, neurological, or cognitive evaluation conducted by a registered physiotherapist (initials JM).

3.2 GENERAL DATA COLLECTION PROCEDURE

3.2.1 Anthropometrics, Demographics, and Health Screens

Data collection involved two lab sessions. During the first session approval for this study was obtained from the Health Sciences Research Ethics Board, Dalhousie University. All participants were informed of the testing procedures and the risk and benefits of the study, and were notified that they could withdraw from the study at any time. Participants were introduced to the lab and provided informed consent. The participants demographic (age (years), sex, physical activity/ core exercise habits (activity/ week)) [Gilleard and Brown, 1994] along with anthropometric parameters (mass (kg), height (m), waist girth (cm), standing elbow height (cm)) were recorded. For older adults, cognitive abilities were screened using a mini mental status examination were participants were required to have a score over 27 [Folstein et al, 1975]. All older participants underwent a postural and neurological assessment which included spinal reflex, myotome, and dermatome function. A registered physiotherapist conducted this screen. Postural assessment included measures for abnormal pelvic tilt, and spinal curvature (lodrosis, kyphosis, or scoliosis). Neurological assessment evaluated any abnormal myotome and dermatome function, as well as their spinal reflexes. Once demographic/ anthropometric data were collected participants were ask to practice, the various exercise tasks that would be performed on their data collection day. This included practice of maximum voluntary contraction exercises as it is known to increases the chance of achieving a higher maximum voluntary contraction [Klass et al, 2007], and the various lifting tasks to ensure participants could complete all tasks and minimize potential learning effects. During the second lab session participants performed these task with data measured from the following sources.

3.2.2 EMG Measurements

Using a standardized protocol 24 pairs of Ag/AgCl surface electrodes (10mm diameter, Red Dot, 3M, London, Ontario, Canada) were positioned bilaterally along the fiber orientation of 6 abdominal and 6 back muscle sites [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012] with an interelectrode distance of 3cm. These sites (Figure 3.1) include: lower rectus abdominis (LRA), half way between the umbilicus and the pubis; upper rectus abdominis (URA), half way between the umbilicus and sternum [Gilleard and Brown, 1994]; anterior external oblique (EO1), orientated 30° angle over the 8th rib adjacent to costal cartilage [Ng et al, 1998]; lateral external oblique (EO2), orientated 45° 15cm lateral to the umbilicus [McGill, 1991]; posterior external oblique (EO3), orientated 80° between the lowest part of the rib cage and iliac crest[Nouwen et al, 1987]; internal oblique (IO), in the triangle formed by the inguinal ligament the lateral border of the rectus abdominis sheath and the line between the anterior superior iliac spine [Ng et al, 1998]; first and third level of the lumbar longissimus (L13/L33) and iliocostalis (L16/L36) erector spinea, 3 and 6 cm lateral to the spinous process respectively [Vink et al, 1987]; quadrates lumborum (L48), 8 cm lateral to the 4th lumbar vertebrae; superficial fibres of the multifidus (L52), 1-2cm lateral to the 5th lumbar vertebrae [Kavcic et al, 2004]. If needed, to account for anthropometric variability minor adjustments to electrode sites were made based on palpation and validation activations. Once electrode placements were established standard skin preparation minimized skin/surface electrode impedance to input impedance ratio less than 1% [Winter, 2009]. Three reference electrodes (one per amplifier) were placed on the right iliac crest.

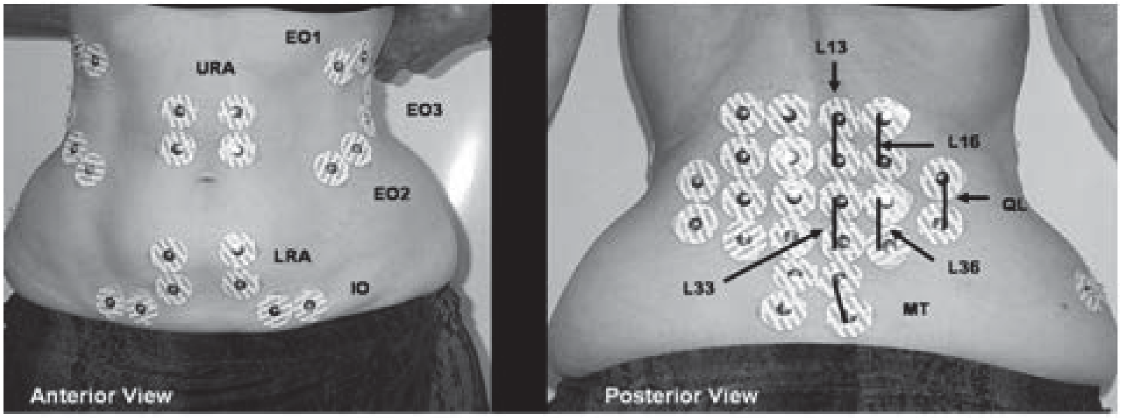

Figure 3.1: Bilateral surface electrode placement on the abdominals (left) and back (right) muscle sites on an older adult participant. Surface electrode positions denoted by their abbreviated form.

EMG signals were pre-amplified (500x), and amplified using 3x AMT-8 EMG systems (Bortec^{TM}, Canada, Bandpass 10-1000Hz, CMRR=115db, input impedance 10GΩ) to best capture maximum EMG activity within ±5 V without signal clipping but to maximize the resolution of a 16 bit A/D card (NI PCI-6259, 16 Bit, National Instruments, Austin, TX). Signals were sampled at 2000 Hz using a custom Labview^{TM} program. This program allowed real time display of EMG signals to validate electrode placement using specific exercises to isolate muscles [Winter et al, 1994] and to ensure high signal quality (low visual signal to noise ratio). All data were stored on an internal hard drive for post processing.

3.2.3 Kinematic Measurements

An electromagnetic Flock of Birds motion tracker system (Ascension Technologies Corp., Burlington, VT) recorded angular motion of the spine and pelvis during the dynamic tasks in 3D space with respect to a global coordinate system (GCS) (Figure 3.2). However, the participants were positioned such that they were parallel with the GCS axis allowing for anatomical interpretation of the global coordinate system angular changes [Hubley-Kozey et al, 2012; Silfies et al, 2005]. Sensors were placed on the left iliac crest, and the 8th thoracic spinous process. These motion data were collected to confirm that participants could successfully minimize motion during the task ensuring that task performance, and external moments from segment masses applied to the body were similar.

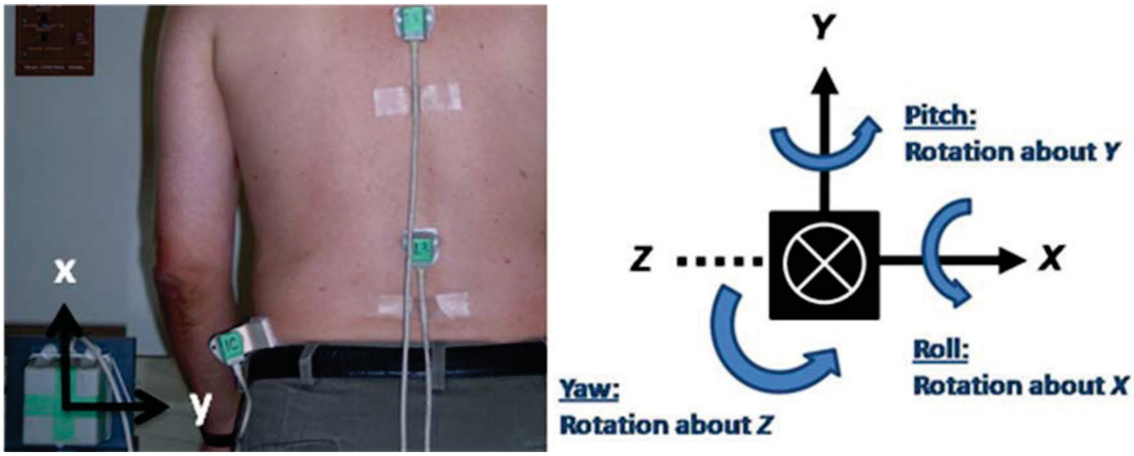

Figure 3.2: Experimental setup for motion analysis showing Flock of Bird Sensors on a participant (left), and how their positions and rotations are expressed with respect to the global coordinate system (right).

3.3 EXPERIMENTAL TASK

To satisfy the objectives of this research electromyographic data were collected from a series of normalization task and three lift and replace tasks were examined. Each lift and replace task involving modifications designed to change the physical intensity for sub objective 1 or cognitive demands of the task related to sub objective 2.

3.3.1 Normalization Task

Prior to completing the experimental tasks, the participants performed a series of exercises that require brief (3 sec) maximum voluntary isometric contractions (MVIC)

[Butler et al, 2009a]. During these normalization exercises, participants were provided with standard verbal encouragement to increase the likelihood of an improved MVIC [McNair et al, 1996]. To ensure participants' safety non-elastic straps ensured that minimal trunk movement would occur during these contractions. Eight different MVIC exercises were performed including: A supine sit-up was used to recruit the rectus abdominis sites. Seated axial rotation (right and left) and side-lying lateral flexion (right and left coupled with ipsilateral hip hike) were used to maximally activate the oblique muscle sites. Back extension and back extension coupled with axial rotation (right and left) were performed in a prone position to maximally recruit the back musculature. These were repeated twice with a 2-minute rest between trials resulting in a total of 16 trials. As the maximum activation of trunk musculature was not found to consistently be elicited with any particular task [Vera-Garcia et al, 2010] the highest root mean square amplitude over a .5s period, regardless of the task used to evoke this activity, was used to normalize activation during the experimental trials [Vezina and Hubley-Kozey, 2000]. Muscle activation amplitudes can then be expressed as a percent of maximal voluntary isometric contraction (%MVIC). This normalization protocol allowed for the comparison between participants and muscle site, by minimizing the variability in surface EMG voltage related to the low pass filtering properties of skin and subcutaneous adipose tissue providing reports of EMG relative to a physiological maximum voluntary activation [Burden, 2010a; Hemingway et al, 1995].

3.3.2 Sub Objective 1: Horizontal Transfer Task with Changing Reach Distance

This task involved lifting a 2.9 kg mass approximately 5cm above standing elbow height off a table surface then transferring the mass from the right side of their body at 60° from their body midline to the left side of their body, 60° from the midline of their body (Figure 3.3a-c). To prevent spinal motion participants were instructed to maintain a natural standing posture while performing the task. To assist the participants' tactile feedback was provided using a vertical jig (Figure 3.3f). [Butler et al, 2010; Hubley-Kozey et al, 2012]. To control the speed and position of arm movement during the lift participants were instructed to transition the load smoothly to an external 5 second count.

Each count corresponded with a particular event. Participants were provided a verbal cue "Ready-Set-Go" and began the lift for each count participants were asked to 1) grasp and begin lifting the mass towards their midline with their right hand; 2) have the mass transferred approximately 30°, or halfway between starting position and the midline of their body; 3) having the mass at the midline of their body transferring it from the right hand to the left hand; 4) moving the mass 30° left to the midline of their body; 5) lowering the mass with their left hand onto a line that was orientated 60° to the midline of their body. Events 1 and 5 were identified using a pressure sensor positioned on the bottom of the mass, and event 3 was captured by a photoelectric relay that participants would break as they transferred the load. Both motion and timing were monitored by examining the traces from the two motion sensors and the timing differences between events. In the event of a greater than 10% trial to trial variation, the trial was repeated as it did not conform to the task specifications.

This task was performed in two conditions performed in random order, which modified the physical intensity of the lift. The intensity of the lift was increased by changing the reach of the participant from either a normal reach (with the elbow bent at 90 degrees) or a maximum reach with the elbow fully extended (Figure 3.3ef) [Butler et al, 2009a; Butler et al, 2010]. By changing the position of the arm, one increased the moment around the spine by i) increasing the moment arm length between the external mass, and center of mass of the forearm and upper arm relative to the spine.

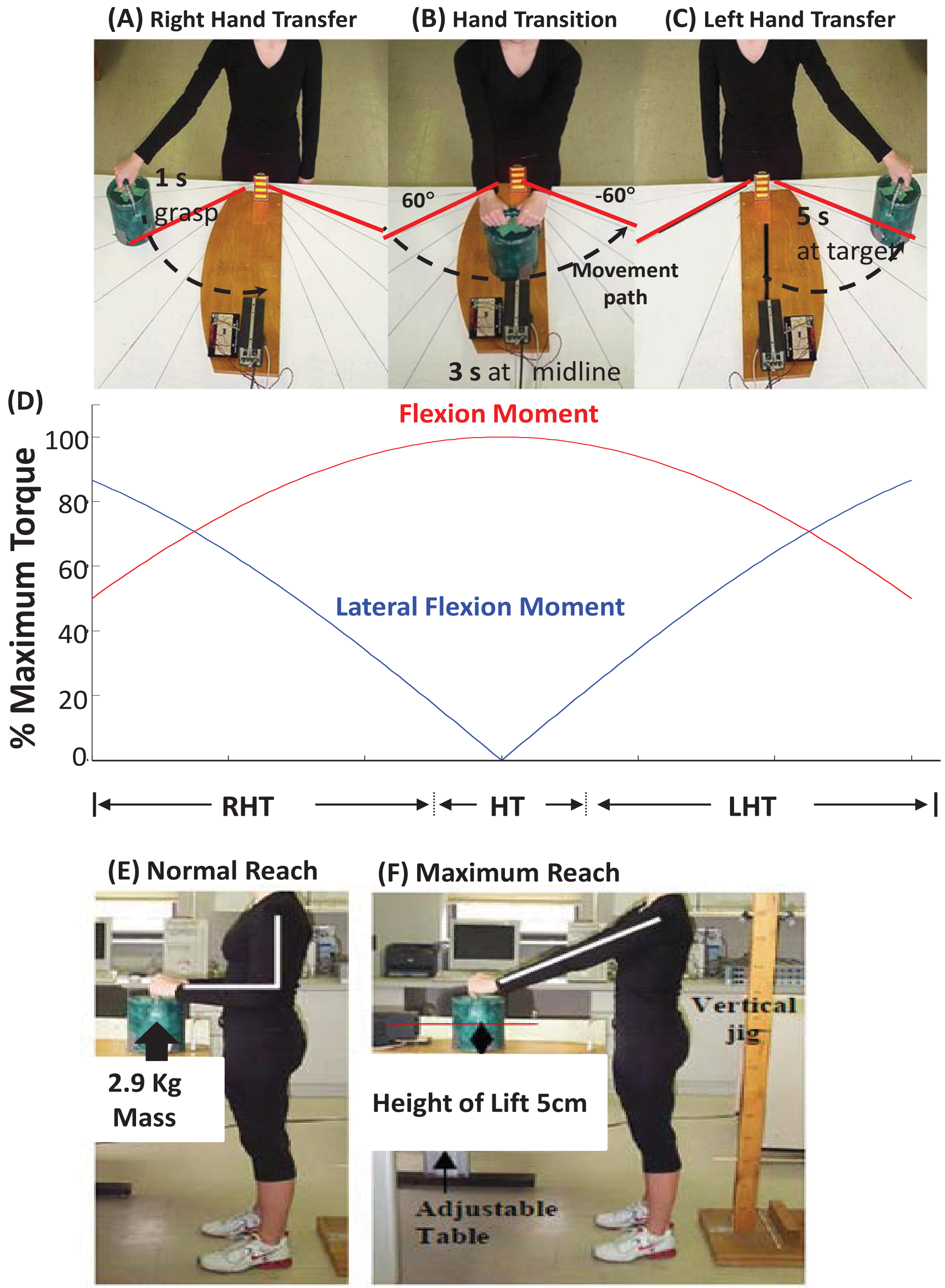

Figure 3.3: Experimental setup for the controlled dynamic transfer task, adapted from Hubley-Kozey et al. 2012 and Butler et al. 2009. As the participants transfer the mass from right to left (a-c) to a controlled count, the external moment generated around the spine changes from right lateral flexion moment (RHT), to a maximum flexion moment (HT), towards a left lateral flexion moment (LHT) (d). To change the external moment of this task participants would lift the mass with either normal reach (e), or maximum reach (f), with additional tactile feedback being provided by a vertical jig to prevent spinal motion (f).

3.3.3 Sub Objective 2: Horizontal Transfer Task in Maximum Reach with Cognitive Dual-task

This task was similar to the lift and replace task done for sub objective 1 performed in maximum reach with the older adults only. However, to modify the cognitive challenge of this task it was either presented as a single or dual-task paradigm. In the single task condition participants performed the horizontal transfer task in maximum reach presented in the previous section (Figure 3.3). For the dual-task a secondary cognitive task involved participants recognizing errors from an auditory subtraction of 3 from a previously stated random number. For this condition, participants were provided with a modified 5 second count where they simultaneously performed mental subtraction by three and were required to remember the number of errors that occurred in the count. For example, participants were provided with a count "Ready Set Go 33 - 31- 28 - 24 - 21". In this example, two errors were made by the external counter. For the participants to have an appropriate dual-task, they would have to recognize these 2 errors. To ensure consistent timing between numbers these counts were prerecorded to a metronome. Multiple trials were recorded starting with a random number with a range of 0-4 errors in their counting sequence. If the participant gave an inappropriate answer to the dual-task count, had excessive movement, or incorrect timing the trial was repeated. This setup was modified from previous work that identified mental subtraction of threes changed neuromuscular response of ankle muscles in older adults [Rankin et al, 2000]. However, this method modified the task to include error recognition to maintain the tempo that exists in the single task condition, and recognize if participants were successfully performing their dual-task. This represented a total of 2 conditions presented in a random blocked order, each requiring a total of 3 trials for ensemble averaging of EMG waveforms.

3.4 DATA PROCESSING

The 3D motion data, and the respective events trigger, were sampled at 50Hz using a custom Labview^{TM} program (National Instruments, Austin, TX, Version 7) and stored on the internal hard drive for post processing using a custom Matlab^{TM} program (The Mathworks Inc, version R2007a, Natick, MA). Motion data were synchronized with EMG data using the first recorded event trigger. These data was low-pass filtered at 2Hz using a second-order recursive Butterworth filter, and then windowed for right hand transfer, hand transfer, and left hand transfer indicated by the event triggers. The maximum angular displacements were determined for all orthogonal angles (yaw, pitch, and roll) during the entire movement and expressed in their approximate anatomical rotations (lateral flexion, flexion extension, and axial rotation) determined by the participants alignment to the global coordinate system [Hubley-Kozey et al, 2012; Silfies et al, 2005].

EMG signals were high passed filtered at 30Hz, to remove electrocardiogram noise [Butler et al, 2009], inverse FFT filtered to remove single frequency power contributions from electromagnetic noise, and corrected for bias and gains. All EMG data were full wave rectified and low-pass filtered at 6Hz using a second-order Butterworth recursive filter. For normalization trials the highest voltage over a 500ms moving average for each muscle was determined regardless of what exercise evoked the MVIC [Vezina and Hubley-Kozey, 2000]. For lifting trials the linear envelope signal was windowed for the entire movement phase (can lift pressure switch trigger- can replace pressure switch trigger) amplitude normalized to MVIC, and time normalized to 101 points using a quadratic interpolation. Through this processing ensemble average profiles were created for each muscle in each participant using the average of all trials within a condition [Shiavi et al, 1998].

3.5 DATA ANALYSIS

The EMG ensemble-average waveforms were used in a pattern recognition technique, principal component analysis model [Hubley-Kozey and Vezina, 2002]. This was achieved using a matrix transformation technique known as eigenvector decomposition of an orthogonal co-variance matrix constructed using the waveforms from all of the participants and muscles within a task. This orthogonal matrix was used to create a transformation matrix (T) the rows of which contained eigenvectors that define a new coordinate system. This coordinate system was a waveform (principal component PC_n) that best captured the variance of the original data. A subsequent principal component (PC_{n+1}) capture the variance of the residuals after characteristic of the previous principal component was removed (Figure 3.4 (PC2)). The process was repeated until an orthogonal transformation matrix of the same dimensions of the covariance matrix was constructed. For each PC a weighting coefficient or PC score was calculated by transforming the original data on the multivariable coordinate system defined by the PC thus determining how much a PC contributes to a measured waveform. Positive PC scores identify that the PC represents the waveform, whereas negative PC scores indicated a negative correlation with the PC waveform. Through the PC scores one can compare between participants', conditions, and muscle sites for similar coefficients would share a common feature in EMG waveforms. PC analysis also allows for the interpretation of muscle synergies and co-activations defined by muscle sites sharing a time varying pattern. Consistent with previous work [Butler et al, 2010; Hubley-Kozey and Vezina, 2002; Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012; Quirk and Hubley-Kozey, 2012] the number of principal components analyzed were determined using two criteria: i) the total explained variance of the waveforms must exceed 90% to accurately reconstruct the original waveform, and ii) the explained variance of an individual PC must be greater than 1%. For each objective, one model was constructed using both the abdominal and back muscle EMG waveforms. For the model each participant ensemble-average waveform for a particular condition for their 24 trunk sites were used to create one [X x101] data matrix. This data matrix was used to construct the orthogonal co-variance matrix necessary for the principal component analysis model described in Figure 3.4. The scores generated from this technique were used for follow up statistical analysis.

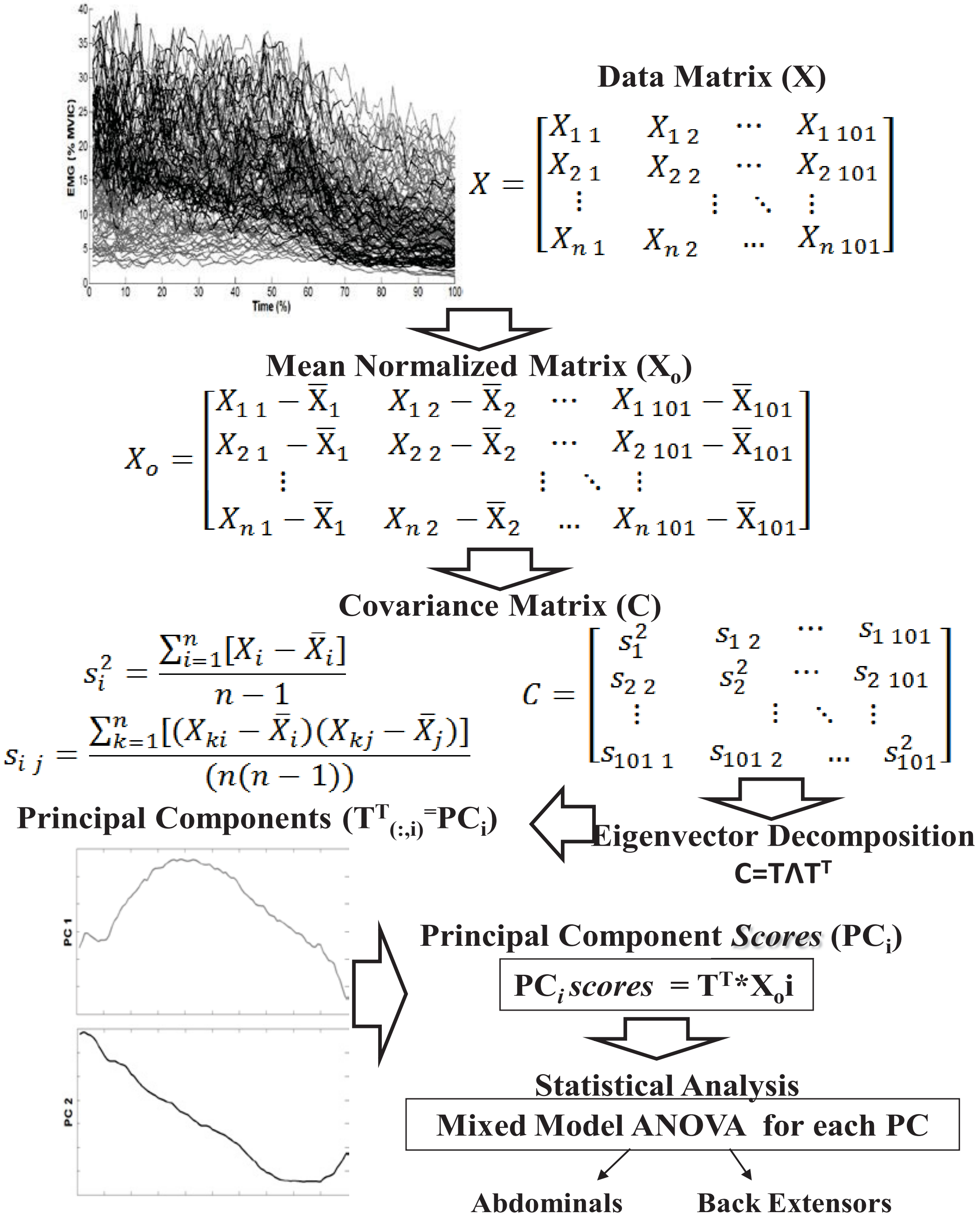

Figure 3.4: Visual display of PCA analysis. A covariance matrix (C) was generated from a data matrix (X) including all ensemble average waveforms using the equations for variance and covariance outlined on the left. The covariance matrix (C) was used in eigenvector decomposition generating an orthogonal transformation matrix (T). The rows of this matrix describe a vector known as the principal component (PCi). For each PC all individual waveforms are transformed on the new coordinate system providing a PC score based on their similarity to the common waveform. These scores were used for subsequent statistical models.

3.6 STATISTICAL ANALYSIS

Independent Student *t* test were performed to compare participant's demographic, anthropometric, and time to complete task between groups. Proportion data, such as sex and normal abdominal function, were tested using a Fisher's exact test. PC scores generated for each task were separated into abdominal and back scores and compared using a mixed model analysis of variance tested for significant differences in the PC scores ($p<0.05$) with the exact model for each objective as follows. For sub objective 1 a three-way (Group*Muscle*Reach) mixed model ANOVA compared the effects of external moment on neuromuscular patterns. For sub objective 2 a three-way (Group*Muscle*Task) mixed model ANOVA tested for the effects of dual-task on trunk muscle activation patterns. For each sub objective, pair-wise comparisons were made using Tukey simultaneous tests for any significant interactions. Normality was confirmed using a Kolmogorov-Smirnov test, with non-normal data transformed using a Johnson Transformation. Statistical operations were performed using Minitab (Minitab Inc, State Collage, PA, version 16).

CHAPTER 4
ARE AGE-RELATED CHANGES IN TRUNK NEUROMUSCULAR ACTIVATION PATTERNS DURING A CONTROLLED FUNCTIONAL TRANSFER TASK EXPLAINED BY INCREASING PHYSICAL TASK INTENSITY

4.1 INTRODUCTION

North America has an aging demographic, by 2030 it is expected that at least one in four adults will exceed the age of 65 years old [Scott et al, 2005; Vincent and Velkoff, May 2010] . While a majority of older adults live and complete activities of daily living independently [Scott et al, 2005], they have an increased risk of experiencing both falls [Pijnappels et al, 2010a; Scott et al, 2005] and low back injuries [Gourmelen et al, 2007; Plouvier et al, 2011] while performing such activities. Both of these injuries are partially related to a common mechanism, the inability to control excessive spinal and trunk motion [Grabiner et al, 2008a; McGill et al, 2003a; Panjabi, 1992b]. Analyzing aging within the theoretical model proposed by Panjabi, physiological changes arising from impaired function and loss of cells in the passive (ligaments, bones, disc, joint capsules, etc.) [Hangai et al, 2008; Iida et al, 2002a], active (skeletal muscles) [D'Antona et al, 2003; Kubo, 1994], and neural sub system (central and peripheral nervous system) [Goble et al, 2010; Kido et al, 2004a; Shaffer, 2007; Taube et al, 2008], can contribute to an increased challenge to maintain spinal stability in older adults. The combined effect of these changes in older adults relative to younger adults results in less passive stiffness [Gallagher et al, 2007] [Allum et al, 2002; D'Antona et al, 2003; Hwang et al, 2008; Kido et al, 2004a; Taube et al, 2008] in older adults. The delay in muscle tension being a multisystem problem partially explained by changes in the neural and active subsystems (the neuromuscular system) of older adults. While current literature indicates an association between trunk function and balance [Davidson et al, 2009; Goldberg et al, 2005; Grabiner et al, 2008a; Kell and Bhambhani, 2006], musculoskeletal injuries [Hicks et al, 2005b], and physical function [Hicks et al, 2005a], few studies have compared trunk muscle activation between younger and older adults while performing functional tasks.

Most studies analyzing movement tasks have found that both trunk kinematics and kinetics differ between younger and older adults [Burgess et al, 2009; Grabiner et al, 2008a; McGill et al, 1999; Van Emmerik et al, 2005]. Trunk motion patterns are partially controlled by the time varying tension generated by multiple trunk muscles (coordination) [Rashedi et al, 2010], thus, trunk muscle activation patterns may differ between these populations. The limited research on the trunk muscle response between older and younger adults [Allum et al, 2002; Asaka and Wang, 2008; de Freitas et al, 2010; Hwang et al, 2008; Kuo et al, 2011; McGill et al, 1999] supports this speculation. Typical findings include: i) increased overall activation of both agonist [Asaka and Wang, 2008; Kuo et al, 2011] and antagonist muscles [Asaka and Wang, 2008; McGill et al, 1999], and ii) delayed onset time [Allum et al, 2002; de Freitas et al, 2010; Hwang et al, 2008] from a small subset of trunk muscles during static balance, or voluntary trunk motion tasks.

Despite the current literature, which compares age-related changes in trunk neuromuscular function, a majority of these studies have methodological limitations that do not capture the complexity of the trunk neuromuscular system. Most studies have only analyzed 2-4 trunk muscle sites [Allum et al, 2002; Asaka and Wang, 2008; de Freitas et al, 2010; Kuo et al, 2011]. However, the trunk is a complex system having multiple muscles, many of these muscles with multiple fiber orientations with unique mechanical advantages (ability to produce moments in a particular plane) [Granata and Marras, 2000; Kavcic et al, 2004; Rashedi et al, 2010; Stokes et al, 2011], and innervations [Urquhart et al, 2005]. Previous studies in young healthy participants illustrate the specificity of both abdominal and back extensor muscle activation during different experimental tasks [Butler et al, 2012; Davidson and Hubley-Kozey, 2005; Hubley-Kozey et al, Submitted; Mirka et al, 1997; Perez and Nussbaum, 2002] thus, under sampling of the muscles could confound age-related differences. A comprehensive study of multiple trunk muscle sites identified that muscle activation characteristics differed between older and younger adults for specific abdominal sites during a leg-loading exercise task [Hubley-Kozey et al, 2009a]. A second potential methodological limitation relates to the outcome measure of neuromuscular activation, especially when examining dynamic tasks. Discrete parameters such as, onsets/offsets [Allum et al, 2002;

Brown et al, 1994; Hwang et al, 2008], or peak/average muscle activation amplitudes throughout the movement [Kuo et al, 2011; McGill et al, 1999], or within particular phases are typically examined. Studies of young healthy adults have shown that temporal relationships are specific to muscle sites during various experimental tasks [Hubley-Kozey and Vezina, 2002; Hubley-Kozey et al, 2012]. Parameterization techniques would not capture these time varying changes in muscle activation. Thus, parameterization may miss age-related differences in neuromuscular control as measured by the time varying muscle amplitude pattern. The activation pattern of multiple abdominal muscles during a dynamic leg-loading exercise task showed that, in specific muscle sites overall amplitude did not significantly differ between younger and older adults. However, comparing temporal activation for specific sites revealed that older adults used a more continuous activation of abdominal sites, whereas younger adults used more variable temporal activation that corresponded with the changing external moment [Hubley-Kozey et al, 2009a; Quirk and Hubley-Kozey, 2012]

To date no comprehensive analysis of trunk neuromuscular activation patterns has been conducted comparing older and younger adults performing a functional lifting task. Thus we have minimal knowledge on whether the trunk muscles have similar synergistic and coordinated patterns in older and younger adults to maintain spinal stability. To isolate changes to muscle activation patterns, a highly controlled dynamic lift, transfer, and replace task has been developed that utilizes voluntary arm movements to constantly change the position of external load thus changing the external moment generated around the spine in two primary directions (flexion, lateral flexion). To minimize trunk motion, the abdominal and back muscle sites have unique activation patterns corresponding to their relative mechanical advantage as reported in healthy young adults [Butler et al, 2010; Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012]. Deviations from these patterns have discriminated individuals based on both sex [Hubley-Kozey et al, 2012], and subacute low back injury [Hubley-Kozey et al, Submitted]. These results identified group differences in the temporal pattern and overall activation of specific sites. Furthermore, these studies show group differences between the synergistic relationships among multiple muscle sites.

The dynamic lift and replace task has been designed to minimize trunk motion that could confound the interpretation of EMG amplitude through altered force length [Kumar, 2010], force velocity [Hodges and Richardson, 1999], and external loads induced by trunk center of mass motion [Hubley-Kozey et al, 2012]. However, by using a standardized external load the demands of this task do not relate to the absolute moment generating ability (muscle strength) of the participants. This is problematic since the various trunk muscles do not scale proportionally with increasing task intensity but rather increase their activation relative to their mechanical advantage [Ng et al, 2003; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008; Song, 2004]. This is true for the dynamic lift and replace task, whereby changing the external moment demands changed the relationship and recruitment of specific trunk muscle sites in younger adults [Butler et al, 2010]. Thus, this non-proportional scaling of trunk muscle activation may confound the interpretation of neuromuscular differences as group differences can be explained by the changes in relative task intensity.

As healthy aging is associated with a decrease in absolute trunk muscle strength [Hasue et al, 1980; Kubo, 1994; Sinaki et al, 2001], higher relative activity would be expected in older adults performing the same task compared to younger adults. This was illustrated by reports that older adults have higher normalized EMG amplitudes of abdominal and back extensor sites while performing experimental tasks typically using body weight as an external load [Kuo et al, 2011; McGill et al, 1999]. How older adults scale their trunk muscle activation patterns to increasing task intensity is unknown, but around the ankle older adults modify their activation amplitudes of different muscle sites in a non-proportional fashion [Hoffren et al, 2007; Hortobagyi et al, 2009]. Whether differences in trunk muscle recruitment utilized by older adults is primarily explained by age or a confounding factor such as increasing physical task intensity is a secondary question of this research.

The purpose of this study was to test if older adults have different trunk muscle activation patterns compared to younger adults to complete a controlled dynamic functional lift and replace task. Due to decreased overall strength, reduced passive stiffness, and delayed response to unexpected loading we hypothesized that compared to younger adults older adults would have: i) higher overall activation of all muscle sites, ii) higher antagonist co-activation, and iii) and muscle activation patterns that have less response to changing external moments. A secondary purpose was to compare both groups at different task intensities, to test if age-related differences in neuromuscular patterns were modified by increasing relative task intensity. We hypothesized that both older and younger adults would modify their spatial and temporal response of muscle sites in a non-proportional fashion to increasing task demands. Principal Component Analysis (PCA) was used to capture both the amplitude and temporal characteristics of the EMG muscle activation waveforms allowing for the examination of similarities or differences between muscle sites, groups, and task intensities [Hubley-Kozey and Vezina, 2002; Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012].

4.2 METHODS

4.2.1 Participants

Participants were recruited via advertisements and electronic notices posted to various sources within the Halifax community. Prior to testing all participants signed an informed consent approved by the Health Science Research Ethics Board at Dalhousie University. Twenty older adults (65+ years old) were matched with 26 younger adults (20-45 years old) selected from a larger group of 60 participants based on sex, mass, and height. Exclusion criteria for both groups included any known cardiovascular, neurological, cognitive, or musculoskeletal conditions, in addition participants were excluded if they experienced a fall or low back injury within the last year that required medical attention, or limited daily function. Three older adult participants were excluded from the study because of an inability to complete the experimental task in the maximum reach. These three participants did not have different demographics from the other older adults that volunteered for this study (Table 4.1).

4.2.2 Test Procedure

A health screening was conducted over the telephone, then confirmed during testing. Participants attended an initial session, which took place within 2 weeks of the testing session to familiarize them with the protocol and test experiment task. Anthropometric data, number of weekly aerobic activity lasting over 30 minutes

[Gilleard and Brown, 1994] and abdominal training sessions they undertook, as well as abdominal function ability [Kendall and McCreary, 1983] were recorded. The older adult group underwent postural, neurological assessment performed by a physiotherapist. To screen for any unknown cognitive impairments older adults required a Mini Mental Status Exam score exceeding 27 [Folstein et al, 1975].

All participants performed a controlled right-to-left transfer task, described in previous studies [Butler et al, 2010; Hubley-Kozey et al, 2012] (Figure 4.1a-c). Briefly, participants stood with their body midline aligned with the centre of a table adjusted vertically to their measured standing elbow height. They performed three trials of the lift, transfer and replace task (2.9 kg) from a standard lift position (60^o right from centre), height lifted (5 cm), and replace position (60^o left from centre) within a standardized 4 second count: lift on 1, midline on 3, replace on 5. Pressure sensors on the bottom of the mass indicated time of lift off and replace; an optoelectric light sensor indicated when the load passed across the midline. These events defined 3 phases: right hand transfer (RHT), hand transition (HT) and left hand transfer (LHT). Time to complete each phase and the total time were recorded. To minimize trunk motion, participants were aligned to a vertical bar and requested to maintain contact with a tactile feedback sensor placed at the mid thoracic spine (Figure 4.1f) [Butler et al, 2010]. If the timing was not adhered to or motion was detected (either visually by the tester or from the recorded event and motion traces), the trial was repeated. Quantitative motion was later determined from electromagnetic motion sensor data (described below) to confirm observations. These constraints resulted in a dynamic task that produced a continuously changing flexion and lateral flexion moments on the trunk, created primarily by the external load (Figure 4.1d). To modify the external moment of these tasks, participants performed the task in two conditions; normal reach and maximum reach where participants maintained an elbow position of 90^o flexion or full extension respectively (Figure 4.1ef).

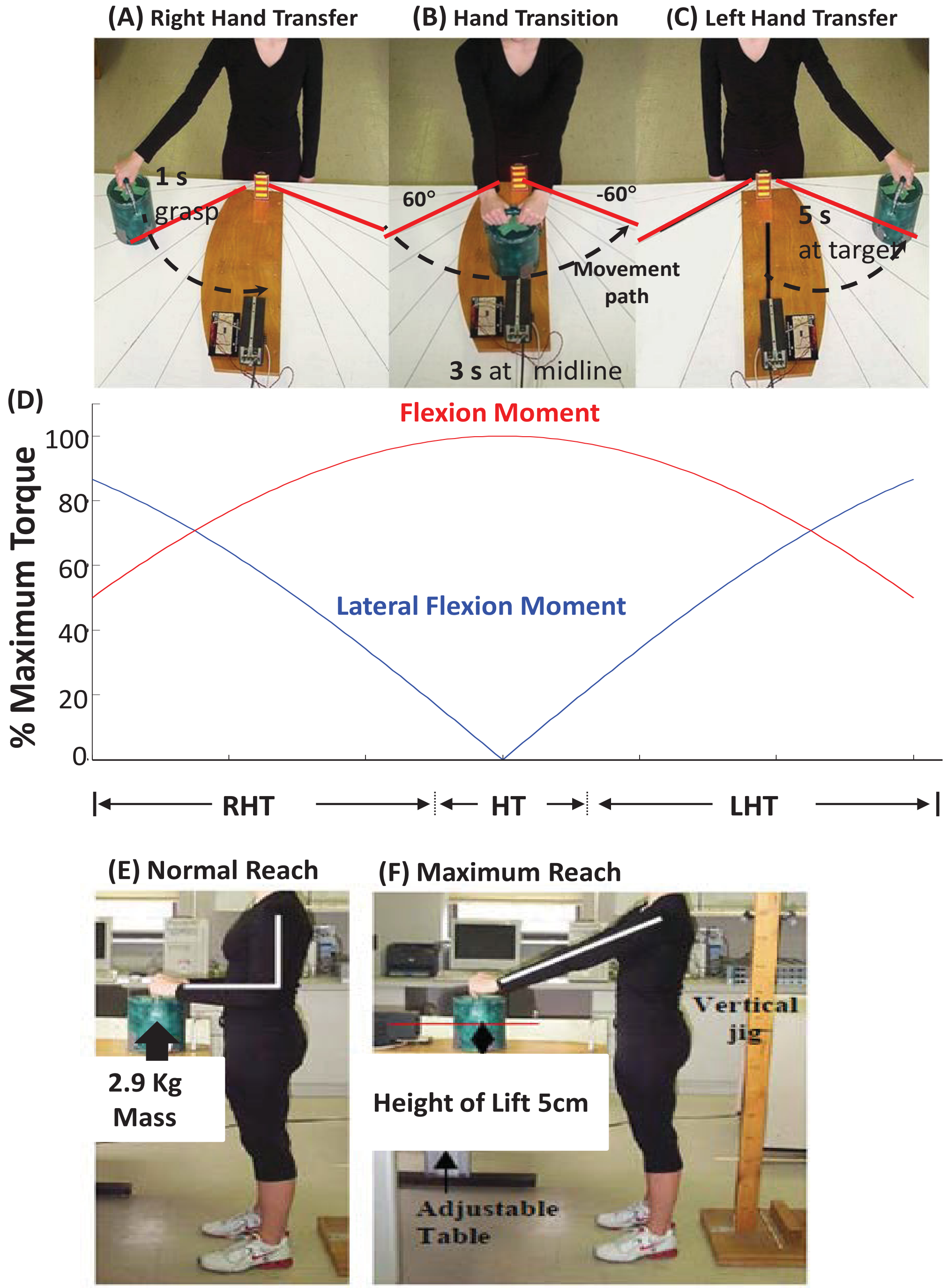

Figure 4.1: Experimental setup for the controlled dynamic transfer task, adapted from Butler et al. 2012 and 2010. As the participants transfer the mass from right to left (a-c) to a controlled count, the external moment generated around the spine changes from right lateral flexion moment (RHT), to a maximum flexion moment (HT), towards a left lateral flexion moment (LHT) (d). To change the external moment of these task participants would lift the mass with either normal reach (e), or maximum reach (f), with additional tactile feedback being provided by a vertical jig to prevent spinal motion (f).

4.2.3 Normalization Procedure

EMG data were amplitude normalized as a percentage of maximum voluntary effort. Each participant performed two trials of eight maximum voluntary isometric exercises. For each exercise, participants were requested to maintain a constant force for 3 seconds with a 2-minute rest between trails. These exercises have been found feasible for older adults [Hanada et al, 2008] and included, restrained sit-up, resisted lateral bend (left/right), resisted trunk extension, resisted trunk extension with left/right rotation, and resisted seated rotation (left/right) [Butler et al, 2010]. A series of normalization tasks as opposed to a single maximum voluntary contraction exercise has been found to better elicit maximum response in trunk muscles [Vera-Garcia et al, 2010].

4.2.4 EMG Data Collection and Processing

Following standard skin preparation, surface electrodes (Ag/AgCl, 10mm circular electrodes; Meditrace, Graphics Control Canada Ltd) were positioned in a bipolar configuration (inter electrode distance of 30mm) along the fiber orientation of 12 bilateral muscle sites. Abdominal muscle sites included placement over upper rectus abdominis (URA), lower rectus abdominis (LRA), internal oblique (IO) and three sites over external oblique (EO1-3), representing the anterior, lateral and posterior fibers of this muscle, respectively. Posterior sites included erector spinae (ES) at the lumbar level 1 (L1) and 3 (L3) , positioned 3cm and 6cm lateral to the midline to represent the longissimus (L13, L33) and iliocostalis (L16, L36) sites; as well as over quadratus lumborum (L48) and multifidus (L52). Specific anatomical landmarks used for these electrode sites and supporting literature have been previously described [Butler et al, 2010]. Electrode placements were validated using a series of manual muscle tests [Kendall and McCreary, 1983; Vezina and Hubley-Kozey, 2000] with slight changes in placement to accommodate individual anthropometry when necessary.

EMG signals were pre-amplified (500X) and further amplified using three AMT-8 EMG systems (band pass 10 – 1000 Hz; CMRR = 115db, input impedance 10GΩ; Bortec Inc., Calgary Alberta). Raw EMG signals and event markers were digitized at 2000 samples per second, using a 16 bit resolution National Instruments analog-to-digital conversion board (National Instruments, CA-1000) and Labview^{TM} software (version 7) and stored for processing.

Processing of EMG signals included correcting the signals for subject bias, calculating the amplitude at the skin level using the calibration constant. All signals were then high pass filtered at 30Hz to remove electrocardiogram artifact [Butler et al, 2009] and an inverse fast-Fourier filtered to remove power generated by electromagnetic sensors. The raw corrected signals were full wave rectified and low passed filtered at 6Hz using a second order recursive Butterworth filter to produce a linear envelope. Signals were amplitude normalized to the maximum voltage regardless of the exercise recorded from a 500ms moving average amplitude recorded from each muscle site during the normalization exercises [Vezina and Hubley-Kozey, 2000], and time base was normalized from lift off (0%) to replace (100%) using a quadratic interpolation algorithm.

EMG ensemble average waveforms for each participant, muscle and condition (2064 X 101) were entered into a PCA model (Figure 4.2) as described previously [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012; Jackson, 2003]. Briefly, an eigenvector decomposition was performed on the covariance matrix of the original waveform matrix, resulting in a set of principal components (PCs) that explained the patterns of variation within the measured EMG waveforms. For each waveform, a PC score was calculated by transforming the original waveform data on the PC, thus calculating its similarity to the PC. This allowed for the comparison of EMG waveforms. Those similar in shape and amplitude would share similar PC scores [Ivanenko et al, 2004]. Thus, quantitative statistical analysis can be performed on the temporal and amplitude characteristics of EMG patterns by comparing the PC scores. PCs were included for analysis if they explained greater than 1% of the total variance.

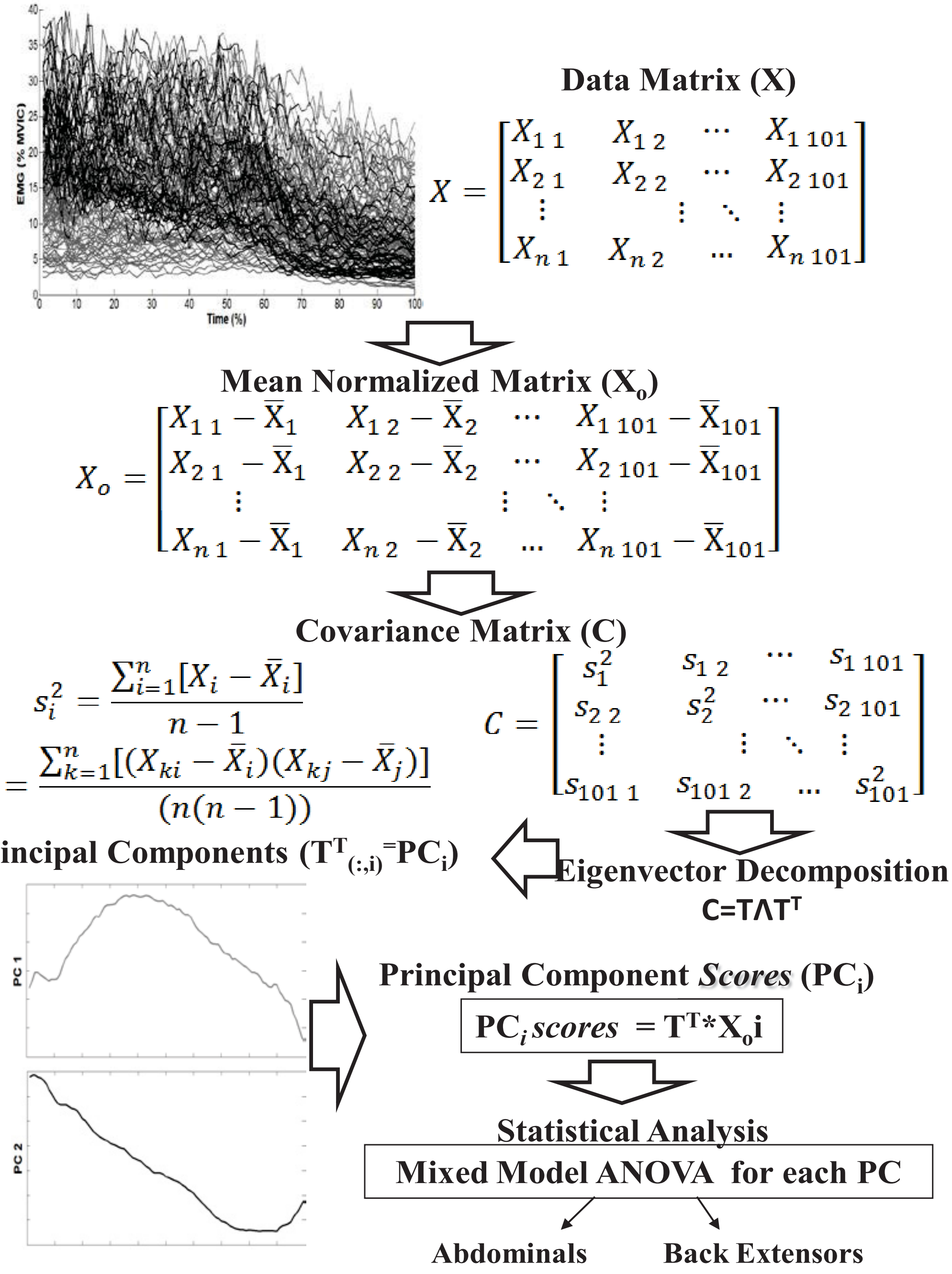

Figure 4.2: Visual description of PCA analysis. A covariance matrix (C) was generated from a data matrix (X) including all ensemble average waveforms using the equations for variance and covariance outlined on the left. The covariance matrix (C) was used in eigenvector decomposition generating an orthogonal transformation matrix (T). The rows of this matrix provide the waveform generated in the principal component (PCi). For each PC all individual waveforms are transformed on the new coordinate system providing a PC score based on their similarity to the common waveform. These scores were used for subsequent statistical models.

4.2.5 Motion Capture Data Collection and Processing

An electromagnetic Flock of Birds^{TM} (FOB) Motion Capture system (Ascension Technology Inc., Burlington, Vermont) recorded the angular motion of the trunk and pelvis throughout the task in 3D with respect to a global coordinate system [Silfies et al, 2005]. One sensor was placed superior to the left anterior superior iliac crest, the second over the T8 spinous process. Participants were positioned such that sensor motion corresponded with anatomical planes of motion (x= Axial Rotation (AR), y= Flexion/Extension (FE), z= Lateral Bend (LB)). Angular motion data were windowed for each phase of the movement. From these the maximum angular displacements for the 3 planes were calculated for each sensor.

4.2.6 Statistical Analysis

When appropriate student t-test or Fishers exacts test were used to test parametric and non-parametric demographic and anthropometric variables. Angular displacement data were compared using a mixed model analysis of variance (ANOVA) (Group*Reach*Phase). PC scores for the abdominal and back muscle sites were used in separate analyses, compared using a Mixed Model ANOVA (Group*Reach*Muscle). Post-hoc comparisons were performed on significant effects using a Tukey simultaneous test. Normality was confirmed using a Kolmogorov-Smirnov test, with non-normal data being transformed using a Johnson Transformation. Statistical analyses were performed in Minitab (Minitab Inc, State Collage, PA, version 16), with α =0.05.

4.3 RESULTS

4.3.1 Participant Demographics and Performance: Timing and Kinematic Variables

Both groups were similar for mass, height, BMI, abdominal function, and self-reported weekly aerobic and abdominal exercise. Older adults had a significantly greater waist circumference, and age compared to younger adults (Table 4.1). The overall total mean time to complete the task was 3.9 ± 0.3s with no significant differences between groups or conditions ($p>0.05$). In addition to the total movement time there were no significant differences between groups or conditions in the time to complete each phase of the lift and replace task ($p>0.05$).

Table 4.1: Descriptive statistics for participants in this study.

Comparison	Older Adults	Younger Adults	Older Adults Excluded
Participants (number)	17	26	3
Age (years)	67.8 (2.5) *	29.7 (7.3)	71.3 (4.7)
Sex (% Male)	76	69	33
Mass (Kg)	82.6 (15.0)	79.4 (13.0)	77.7 (17.0)
Height (cm)	171.7 (7.6)	173.4 (7.9)	169.5 (12.7)
BMI (Kg/m^2)	27.9 (4.1)	26.2 (2.7)	26.8 (1.8)
Waist Girth (cm)	94.9 (12.8) *	91.2 (13.2)	89.7 (12.0)
Aerobic Training (sessions/week)	3.5 (3.1)	4.4 (3.8)	6.0 (1.7)
Abdominal Training (sessions/week)	1.4 (2.0)	1.7 (2.0)	2.7 (2.5)
Normal Abdominal Function (% complete)	82	88	66

Mean (SD)
*=Significant difference ($p<0.05$) between younger and older adults

The overall trunk and pelvis motion was 3.2° or less (Table 4.2) for both groups and all conditions, with the largest motion at both sensors in axial rotation. There were significant ($p<0.05$) main effect and interactions for the maximum angular displacements as indicated in Table 4.2. Significant group main effects were determined for pelvis lateral flexion, and flexion/extension where older adults had greater motion than younger adults; however differences were less than 1 degree. Reach main effects were significant for trunk flexion/extension and axial rotation where maximum reach had more motion than normal reach. Phase main effects were reported for trunk flexion/extension where phase 1(RHT) and 3(LHT) had more motion than phase 2(HT) (Table 4.2). In addition to these main effects Reach*Phase interactions were reported in trunk lateral flexion, and pelvis lateral flexion, and flexion/extension. Group*Phase interactions were identified for both pelvis and trunk axial rotation with post hoc findings illustrated in Table 4.2. There was minimal variability for each measure, which confirms that participants in both groups performed the task as per the instructions to minimize motion.

4.3.2 Qualitative EMG Waveform Analysis

Example ensemble average waveforms for the abdominal (Figure 4.3) and back extensor (Figure 4.4) muscles show qualitative differences between muscle sites, groups, and reach distances for the entire movement. Qualitatively most abdominals and back extensor muscles had higher overall EMG activation levels in the older adults, but this difference was not consistent throughout the task (Figure 4.3efh and 4.4ef). In all back (Figure 4.4) and specific abdominal (Figure 4.3cdef) sites increasing reach distance increased the overall EMG activation in both younger and older adults. However, for specific sites these differences were dependent on the phase of the lift (Figure 4.4f versus 4.4g). Finally, qualitative analysis showed differences between muscle sites. For example, the more lateral sites in both the abdominals (Figure 4.3ef) and back (Figure 4.4abef) responded with high to low activity or a reverse pattern if they were on the left or right respectively. Three PCs captured over 97% of the total variance in the EMG waveforms.

Table 4.2: Motion data for older and younger adults in normal (N) and maximum (M) reach distances, and for the phases right hand transfer (RHT), hand transition (HT), and left hand transfer (LHT)

	Pelvis (^{o})			**Trunk (^{o})**		
	Lat. Flex	**Flex.-Ext.**	**Axial Rot.**	**Lat. Flex.**	**Flex.-Ext.**	**Axial Rot.**
Old RHT N	0.7 (0.3)	0.8 (0.4)	**2.4 (1.3)**	**1.9 (1.6)**	**1.5(1.7)**	**3.1 (2.1)**
Old RHT M	1.2 (0.7)	1.3 (0.9)	**3.2 (2.5)**	**2.4 (2.0)**	**1.7 (1.7)**	**3.2 (1.9)**
Young RHT N	0.8(0.6)	0.7(0.5)	**1.5 (1.1)**	**1.0 (0.8)**	**0.6 (0.5)**	**1.2 (0.7)**
Young RHT M	1.0 (0.7)	1.5 (1.1)	**2.3 (1.5)**	**1.3 (0.9)**	**0.8 (0.8)**	**1.6 (1.0)**
Old HT N	0.6 (0.6)^{ab}	0.5 (0.3)^{ab}	1.3 (0.9)^{ab}	**0.6 (0.5)**^{a}	**0.6 (0.3)**^{ab}	0.9(0.3)^{ab}
Old HT M	0.7 (0.8)^{ab}	0.6 (0.5)^{ab}	1.3 (1.1)^{ab}	**0.5 (0.3)**^{ab}	**0.6 (0.3)**^{ab}	1.0 (0.4)^{ab}
Young HT N	0.5 (0.2)^{ab}	0.5 (0.3)^{ab}	1.1 (0.6)^{ab}	**0.4 (0.2)**^{a}	**0.3 (0.2)**^{ab}	0.6 (0.4)^{ab}
Young HT M	0.6 (0.4)^{ab}	0.8 (0.6)^{ab}	1.4 (0.9)^{ab}	**0.4 (0.3)**^{ab}	**0.3 (0.2)**^{ab}	0.6 (0.4)^{ab}
Old LHT N	0.9 (0.4)	0.6 (0.4)	1.9 (1.1)	**0.6 (0.3)**^{c}	**0.7 (0.4)**^{c}	1.5 (0.9)^{c}
Old LHT M	1.3 (0.9)	1.2 (1.1)	2.5 (1.4)	**1.0 (0.5)**^{c}	**1.0 (0.6)**^{c}	1.7 (0.9)^{c}
Young LHT N	1.0 (0.5)	0.9 (0.6)	1.9 (0.8)	**0.6 (0.5)**^{c}	**0.4 (0.3)**^{c}	1.4 (0.8)
Young LHT M	1.2 (0.7)	1.5 (1.0)	2.9 (2.6)	**0.7 (0.4)**^{c}	**0.5 (0.3)**^{c}	1.2 (0.5)

Mean (SD)

Significant differences in Tukey HST Post hoc: within the same phase and group grey shading= Maximum>normal, within the same phase and reach bold= Old> Young, and within the same group and reach a=RHT>HT, b= LHT>HT, c=RHT>LHT

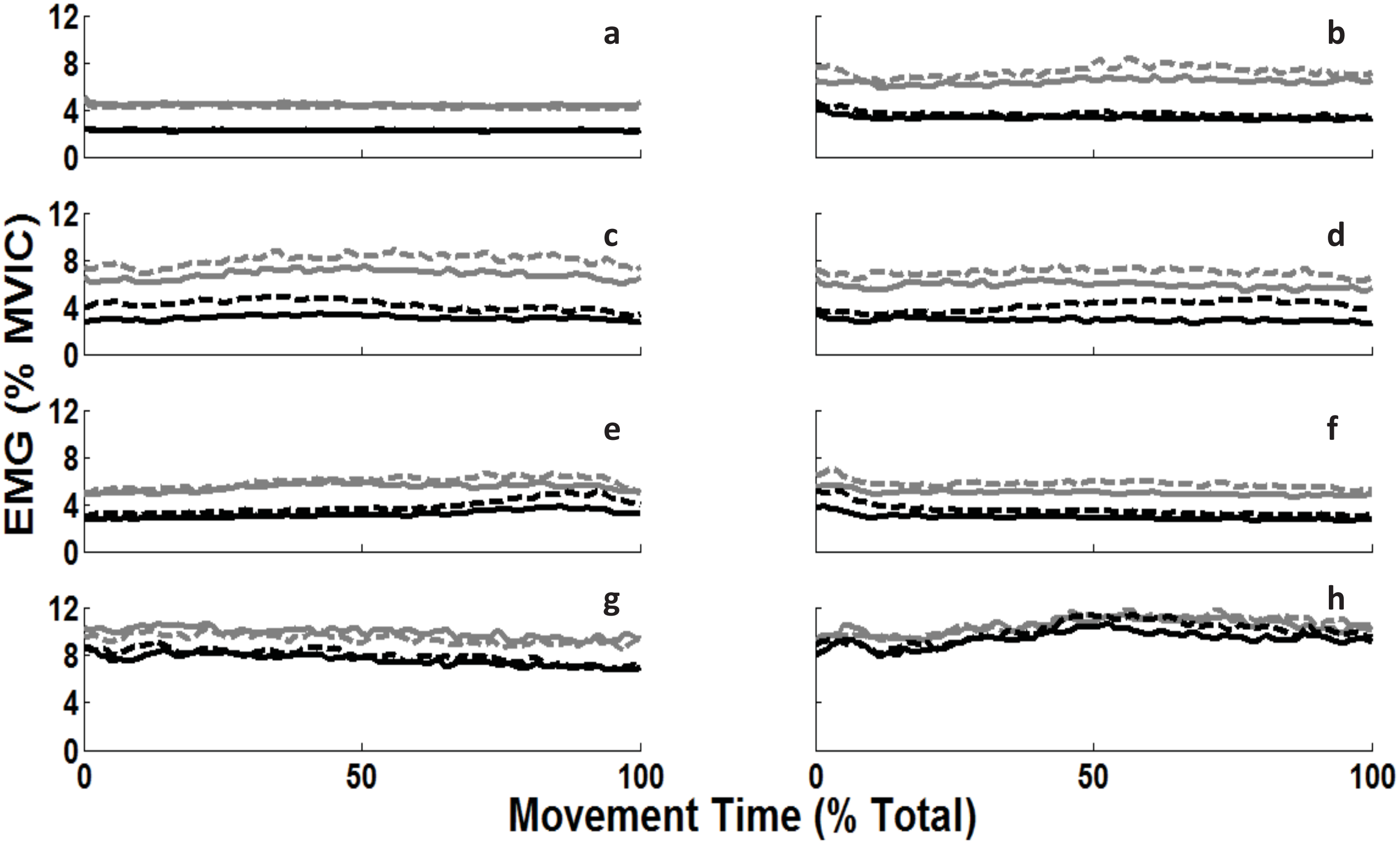

Figure 4.3: Examples of ensemble average waveforms for specific abdominal sites. Black lines= younger adult group, grey lines= older adult group, solid lines= normal reach, dashed lines= maximum reach. Specific abdominal sites include left (L) and right (R), a) LURA, b) REO2, c & d) L & REO1, e & f) L & REO3, and g & h) L & RIO.

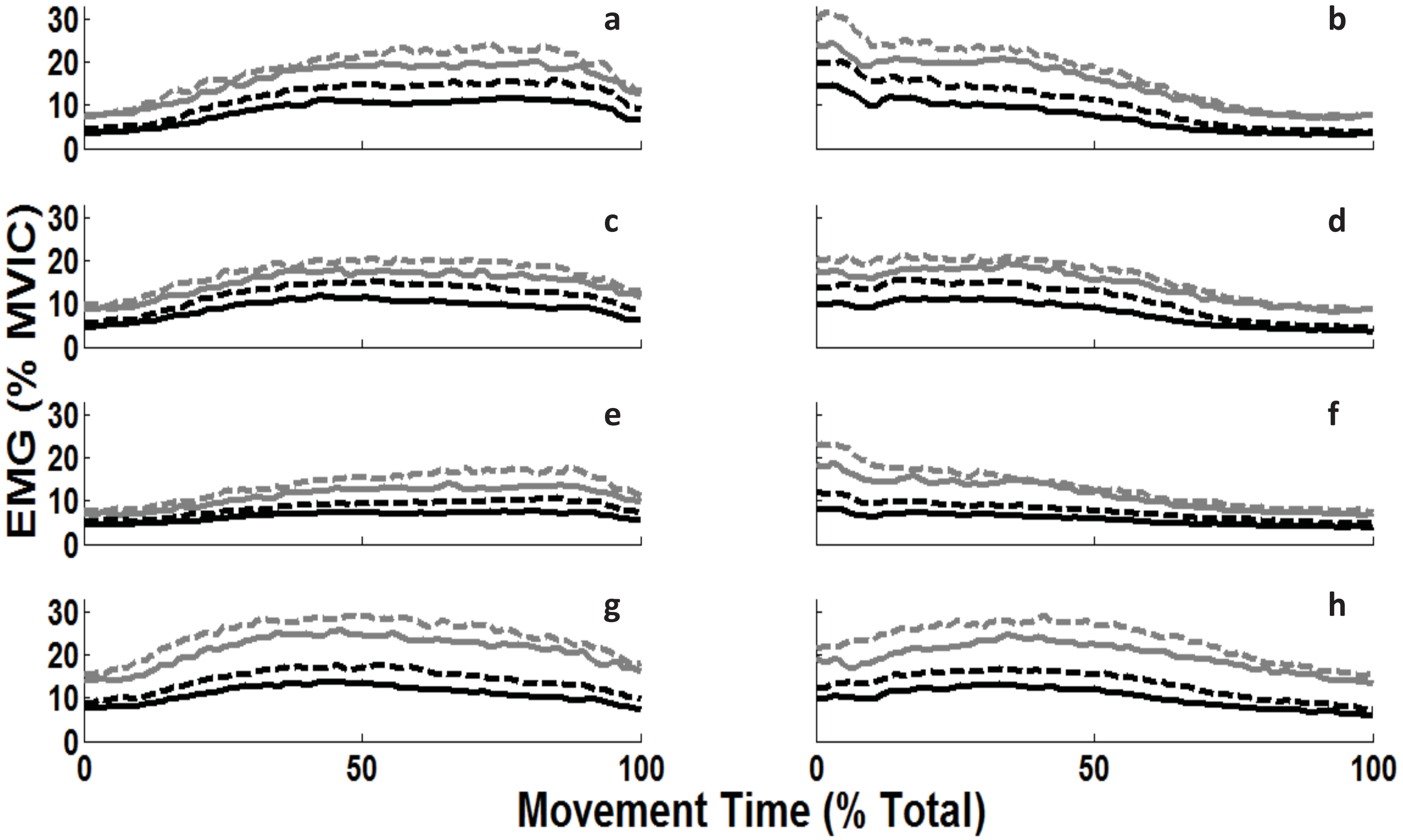

Figure 4.4: Examples of ensemble average waveforms for specific back sites. Black lines= younger adult group, grey lines= older adult group, solid lines= normal reach, dashed lines= maximum reach. Specific back sites include left (L) and right (R), a & b) L & RL16, c & d) L & RL33, e & f) L & RL48, and g & h) L & RL52.

4.3.3 Principal Component 1

PC1 explained over 85% of the total variance capturing a gradual increase in muscle activity corresponding with the increasing flexor moment at hand transition (HT), followed by a gradual decrease (Figure 4.5a). High-low score curves reveal that PC1 featured overall activation for muscle sites with high scores associated with higher EMG activity (Figure 4.5b).

There was a significant Group*Reach*Muscle interaction $p<0.001$ for the abdominal sites PC1. For all muscles and both reach distances, older adults had higher overall activation (PC1) than younger adults (Figure 4.6a). Progressing from normal to maximum reach PC1 scores increased for EO1 in both groups and in EO3 for only the younger adults (Figure 4.6a). Comparing PC1 scores among abdominal sites demonstrated that overall activation of IO was greater than all other abdominal sites within both groups and reach distances (Figure 4.6a & Table 4.3). However, these groups had altered relationships between EO and RA sites (Table 4.3). Analyzing contralateral muscle asymmetries, both groups demonstrated L>R PC1 scores in the IO for both reach distances (Table 4.3). However, older adults had additional asymmetries with up to 4/6 abdominal sites having asymmetric PC1 scores in maximum reach (Figure 4.6a).

For the back sites PC1 scores resulted in a Group*Reach ($p<0.001$) interaction, where older adults had higher overall activation than younger adults for both reach distances (Figure 4.7b & Table 4.4). Both the Group*Reach and the Muscle*Reach ($p=0.005$) post hoc analysis confirmed in both groups all back sites increased overall activation (PC1) with increasing reach distance (Figure 4.7a). Comparing individual muscles sites, L52 had the highest overall activation (PC1), followed by a progression where rostral medial sites had higher PC1 scores then lateral caudal sites (Table 4.4). Progression to maximum reach resulted in more asymmetries between back muscle sites with up to 4/6 back sites having an R>L difference (Figure 4.7a) for both groups.

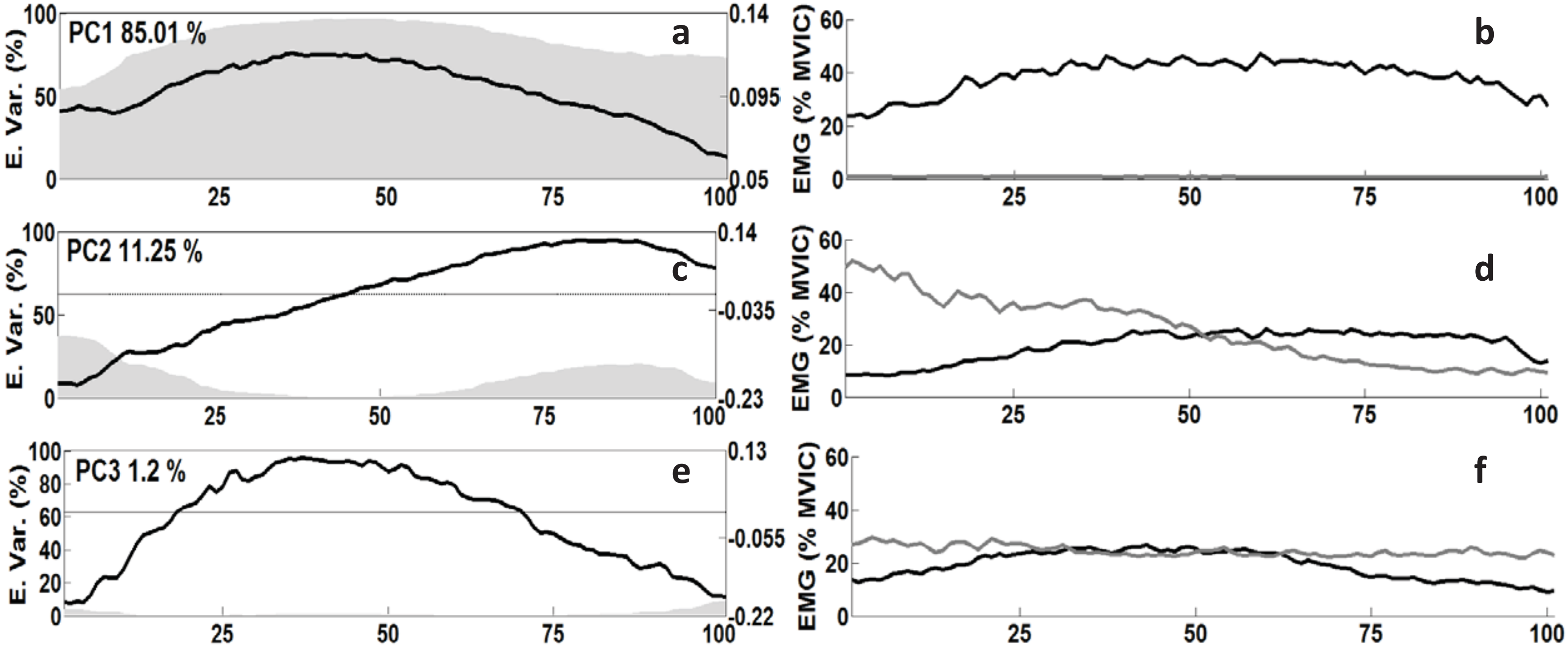

Figure 4.5: For principal components (1-3) plots a, c, and e show the principal component (black line), and explained variance relative to total movement time depicted by grey shading. Total explained variance for each principal component is shown in the top left corner of these plots. To assist with interpretation, for each principal component plots b, d, and f show ensemble average waveforms of EMG activation patterns from the 5 highest (black line) and the 5 lowest (grey line) PC scores.

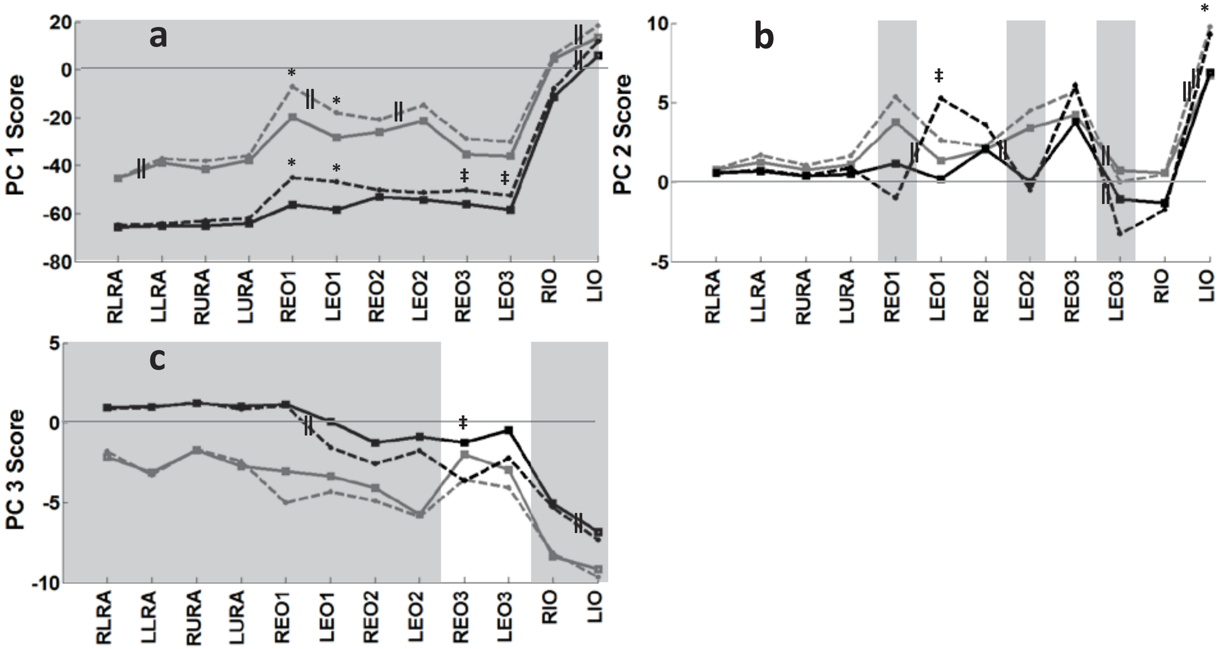

Figure 4.6: Interaction plots for principal components scores 1-3 (a,b,c) corresponding to PC1-3 in Figure 4.5 a,c,e respectively for the abdominal muscle sites of older (grey) and younger adults (black) in normal (solid line) and maximum (dashed line) reach. Grey shading indicates significant group differences in maximum reach. Reach interactions are shown by *= significant reach different in both groups, and ‡= significant reach difference in younger adults only. Significant R/L differences in muscle sites are indicated by ‖. To assist with interpretation of PC scores see corresponding High/Low score plots (Figure 4.6 b,d,f).

Table 4.3: Principal component scores (1-3) of abdominal sites for older and younger adults in normal and maximum reach

Muscle	Reach	PC1 Old	PC1 Young	PC2 Old	PC2 Young	PC3 Old	PC3 Young
RLRA	Normal	**-45.1 (21)**^{cdef}	**-65.6 (11)**^{cdef}	0.8 (1)^{e}	0.6 (1)^{e}	**-2.2 (3)**	**1.0 (1)**
RLRA	Max	**-45.3 (22)**$^{\S bcdef}$	**-65.0 (11)**^{cdef}	0.8 (1)^{ce}	0.6 (1)^{de}	**-1.8 (3)**	**0.9 (2)**
LLRA	Normal	**-38.9 (23)**^{cdf}	**-65.2 (12)**^{cdef}	1.3 (1)	0.7 (1)	**-3.1 (4)**	**1.0 (2)**
LLRA	Max	**-37.2 (23)**$^{\S bcdef}$	**-64.5 (12)**^{cdef}	1.7 (1)	0.8 (1)	**-3.3 (4)**	**1.0 (2)**
RURA	Normal	**-41.6 (29)**^{cdf}	**-65.2 (12)**^{cdef}	0.8 (1)^{e}	0.4 (1)^{e}	**-1.7 (4)**	**1.3 (2)**
RURA	Max	**-38.2 (32)**^{cdef}	**-63.1 (14)**^{cdef}	1.1 (1)^{ce}	0.4 (1)^{de}	**-1.7 (4)**	**1.3 (1)**
LURA	Normal	**-37.8 (42)**^{cdf}	**-64.1 (12)**^{cdef}	1.1 (1)	0.5 (1)	**-2.7 (7)**	**1.0 (2)**
LURA	Max	**-35.9 (42)**^{cdf}	**-62.0 (14)**^{cdef}	1.7 (2)	0.9 (1)^{e}	**-2.4 (6)**	**0.9 (2)**
REO1	Normal	**-19.8 (51)**^{*f}	**-56.5 (14)**^{f}	3.7 (6)	1.2 (5)^{e}	**-3.0 (6)**	**1.1 (3)**
REO1	Max	**-7.3 (58)**^{*f}	**-45.2 (18)**^{f}	**5.3 (11)**	**-1.0 (8)**$^{\S de}$	**-5.0 (7)**^{ab}	**1.1 (4)***
LEO1	Normal	**-28.4 (42)**^{*f}	**-58.4 (16)** ^{f}	1.4 (4)	0.2 (4)	**-3.4 (6)**	**0.1 (3)**
LEO1	Max	**-18.0 (49)**^{*f}	**-46.9 (18)**^{f}	2.6 (7)	5.3 (6)$^{\S abde}$	**-4.3 (6)**	**-1.5 (4)**^{*ab}
REO2	Normal	**-26.1 (30)**^{f}	**-53.1 (19)**^{f}	2.1 (6)	2.1 (3)	**-4.1 (5)**^{b}	**-1.2 (3)**^{abc}
REO2	Max	**-20.9 (33)**$^{\S cf}$	**-50.2 (19)**^{f}	2.3 (7)	3.6 (4)*	**-4.9 (6)**^{ab}	**-2.6 (4)**^{abc}
LEO2	Normal	**-19.7 (44)**^{f}	**-54.4 (18)**^{f}	**3.4 (5)**	**0.0 (2)**	**-5.5 (6)**^{be}	**-0.9 (3)**^{ab}
LEO2	Max	**-13.3 (56)**$^{\S f}$	**-51.3 (19)**^{f}	**4.3 (6)**^{e}	**-0.5 (4)***	**-5.6 (6)**^{ab}	**-1.8 (4)**^{ab}
REO3	Normal	**-35.3 (22)**^{cdf}	**-56.3 (13)**^{f}	4.2 (3)*	3.8 (3)*	-2.0 (4)	-1.3 (2)^{abc}
REO3	Max	**-28.8 (27)**^{cdf}	**-50.2 (15)**^{f}	5.7 (4)*	6.1 (5)*	-3.6 (4)	-3.6 (4)^{abc}
LEO3	Normal	**-36.1 (26)**^{cdf}	**-58.5 (16)**^{f}	0.7 (2)*	-1.1 (3)*	**-2.9 (3)**	**-0.5 (2)**
LEO3	Max	**-30.0 (28)**^{cdf}	**-52.6 (19)**^{cf}	**0.0 (2)***	**-3.2 (6)***	-4.1 (5)	-2.2 (3)^{ab}
RIO	Normal	**4.6 (48)**§	**-11.4 (33)**§	0.6 (3)$^{\S ce}$	-1.3 (4)$^{\S cde}$	**-8.4 (11)**^{abcde}	**-5.0 (4)**^{abcde}
RIO	Max	**6.1 (50)**§	**-7.9 (39)**§	0.5 (6)$^{\S ce}$	-1.7 (5)$^{\S de}$	**-8.1 (11)**^{abcde}	**-5.3 (6)**^{*abcd}
LIO	Normal	**13.3 (45)**§	**1.9 (46)**§	6.7 (5)$^{\S abcde}$	5.3 (7)$^{\S abcde}$	**-9.1 (8)**^{abcde}	**-6.0 (6)**^{abcde}
LIO	Max	**18.3 (48)**§	**6.3 (51)**§	9.8 (9)$^{\S abcde}$	7.9 (8)$^{\S abcde}$	**-9.6 (9)**^{abcde}	**-7.3 (8)**^{*abcde}
Mean	Normal	-25.9 (40)	-50.6 (29)	2.2 (4)	1.0 (4)	-4.0 (6)	-0.8 (4)
Mean	Max	-20.9 (44)	-46.0 (32)	3.0 (6)	1.6 (6)	-4.5 (7)	-1.6 (5)

Post hoc analysis indicating difference between, groups= bold lettering, reach distance= grey shading, side differences between paired muscle sites *=R>L, §=L>R, and within a particular group and reach distance when a particular site is < (or in the circumstance of PC2 because of reversed directions left sites are presented > for ease of interpretation) a=LRA, b=URA, c=EO1, d=EO2, e=EO3, f=IO

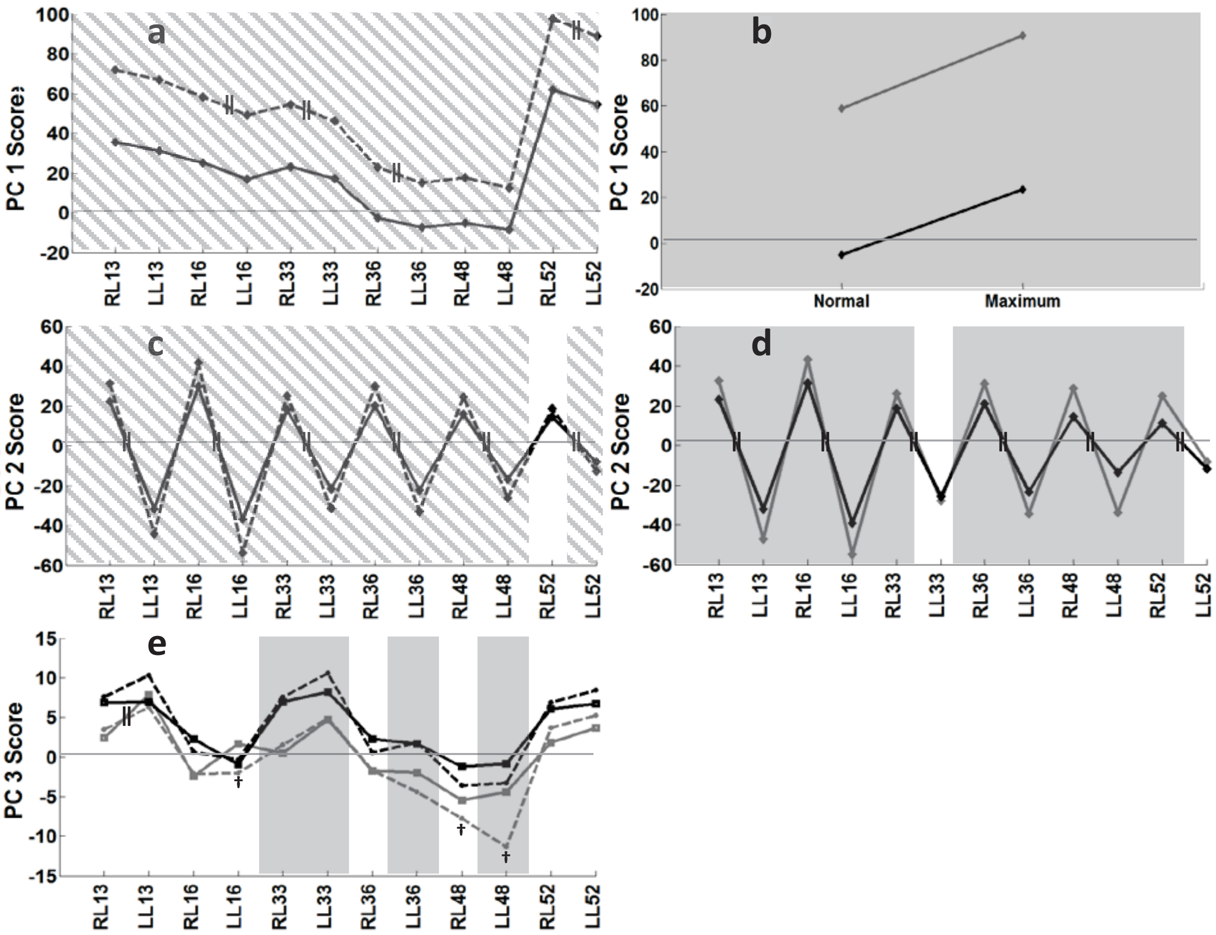

Figure 4.7: Interaction plots for principal component scores 1-3 (as indicated in the y-axis label) for the back muscle sites corresponding to PC1-3 in Figure 4.5 a,c,e repectively. Common to all group interactions (b,d,e) line colour corresponds to older (grey) and younger (black) adults. Common to reach interactions (a,c,e) line style corresponds to normal (solid line) and maximum (dashed line) reach. Grey shading indicates group differences in group by reach interactions (d), and group differences in maximum reach for group*reach*muscle interactions (e). Significant between reach differences are indicated by diagonal line shading. For

muscle*reach interactions (a,b), † indicates significant reach differences in older adults only (b). Common to all muscle interactions significant R/L differences in muscle sites indicated by ‖ (subplots a,c,d,e). To assist with interpretation of PC scores see corresponding High/Low score plots (Figure 4.6 b,d,f).

Table 4.4: Principal component scores (1-3) of back sites for older and younger adults in normal and maximum reach

Muscle	Reach	PC1 Old	PC1 Young	PC2 Old	PC2 Young	PC3 Old	PC3 Young
RL13	Normal	**81.0 (90)**^{f}	**6.0 (34)**^{f}	**28.3 (14)**^{*b}	**18.5 (9)**^{*b}	**2.4 (12)** §	**6.9 (6)**
RL13	Max	**121.8 (113)**^{f}	**39.6 (40)**^{f}	**37.0 (18)***	**27.8 (12)***	3.5 (12)	7.6 (7)
LL13	Normal	**82.0 (75)**^{f}	**-2.4 (31)**^{f}	**-39.2 (32)**^{*b}	**-26.6 (19)**^{*b}	7.8 (11) §	6.9 (6)
LL13	Max	**119.8 (94)**^{f}	**32.3 (37)**^{f}	**-54.8 (40)***	**-37.2 (24)***	6.3 (10)	10.3 (7)
RL16	Normal	**60.9 (88)**^{af}	**1.9 (46)**^{af}	**36.7 (25)***	**25.7 (21)***	**-2.4 (9)**	**2.3 (6)**^{ac}
RL16	Max	**96.6 (113)**^{*af}	**33.3 (56)**^{*af}	**49.7 (33)***	**36.8 (27)***	-2.1 (11)^{af}	0.6 (8)^{acf}
LL16	Normal	**59.2 (70)**^{f}	**-11.0 (40)**^{f}	**-44.7 (30)***	**-31.6 (32)***	1.6 (7)^{a}	-1.0 (12)^{acf}
LL16	Max	**92.1 (84)**^{*af}	**21.0 (48)**^{*af}	**-65.0 (41)***	**-46.3 (41)***	-2.0 (9)^{acf}	-0.4 (15)^{acf}
RL33	Normal	**51.8 (65)**^{af}	**4.5 (62)**^{af}	**23.6 (12)**^{*ab}	**14.4 (6)**^{*b}	**0.5 (8)**	**6.9 (9)**
RL33	Max	**85.0 (77)**^{*af}	**34.7 (73)**^{*af}	**28.8 (14)***	**22.9 (9)***	**1.5 (12)**	**7.6 (9)**
LL33	Normal	**53.1 (85)**^{af}	**-6.4 (46)**^{af}	-22.7 (14)^{*abde}	-21.0 (18)^{*ab}	4.7 (13)	8.2 (6)
LL33	Max	**78.8 (92)**^{*af}	**24.8 (59)**^{*af}	-32.9 (19)*	-30.3 (25)*	**4.9 (11)**	**10.6 (7)**
RL36	Normal	**21.7 (79)**^{abcf}	**-18.5 (46)**^{abcf}	**26.7 (28)**^{*b}	**16.0 (8)**^{*b}	-1.8 (7)	2.2 (5)^{ac}
RL36	Max	**49.3 (93)**^{*abcf}	**5.6 (56)**^{*abcf}	**35.8 (33)***	**25.9 (12)***	-1.7 (14)^{af}	0.6 (6)^{acf}
LL36	Normal	**17.0 (69)**^{abcf}	**-23.1 (43)**^{abcf}	**-28.3 (27)**^{*ab}	**-18.3 (14)**^{*ab}	-2.0 (13)^{acf}	1.7 (6)^{acf}
LL36	Max	**38.0 (79)**^{*abcf}	**0.3 (49)**^{*abcf}	**-40.2 (38)***	**-28.4 (20)***	**-4.4 (16)**^{acf}	**1.8 (8)**^{acf}
RL48	Normal	**20.8 (60)**^{abcf}	**-22.3 (37)**^{abcf}	**23.6 (25)**^{*b}	**11.0 (5)**^{*abcd}	-5.4 (6)^{acf}	-1.2 (4)^{acf}
RL48	Max	**49.3 (90)**^{abcf}	**-2.8 (41)**^{abcf}	**34.2 (39)***	**18.3 (9)***	-7.7 (12)^{abcdf}	-3.6 (5)^{abcdf}
LL48	Normal	**23.8 (62)**^{abcf}	**-29.8 (34)**^{abcf}	**-27.4 (33)**^{*ab}	**-10.2 (8)**^{*abcd}	-4.4 (9)^{acf}	-0.9 (4)^{acf}
LL48	Max	**46.6 (73)**^{abcf}	**-9.9 (42)**^{abcf}	**-40.0 (48)***	**-16.9 (12)***	**-11.3 (16)**^{abcdf}	**-3.3 (7)**^{acdf}
RL52	Normal	**119.3 (68)**	**24.3 (59)**	**24.3 (23)**^{*abd}	**8.2 (8)**^{*abcde}	1.8 (10)	6.0 (7)
RL52	Max	**160.6 (78)***	**56.7 (73)***	**25.5 (21)***	**14.3 (12)***	3.7 (14)	6.9 (9)
LL52	Normal	**114.3 (76)**	**15.7 (47)**	-4.5 (12)^{*abcde}	-10.4 (10)^{*abcd}	3.7 (19)	6.7 (6)
LL52	Max	**151.8 (83)***	**48.0 (55)***	-12.0 (18)*	-13.1 (14)*	5.2 (21)	8.4 (8)
Mean	Normal	**58.7 (80)**	**-5.1(47)**	-2.0 (24)	-0.3 (37)	3.7 (7)	0.5 (11)
Mean	Max	**90.8(56)**	**23.6(56)**	-2.2 (35)	-2.8 (51)	3.9 (9)	-0.3 (14)

Post hoc analysis indicating difference between, groups= bold lettering, reach distance= grey shading, side differences between paired muscle sites *=R>L, §=L>R, and within a particular group and reach distance when a particular site is < (or in the circumstance of PC2 because of reversed directions left sites are presented > for ease of interpretation) a=L13, b=L16, c=L33, d=L36, e=L48, f=L52

4.3.4 Principal Component 2

PC2 explained over 11% of the total variance capturing, a gradual increase in activity corresponding with the lateral flexion moment as it moved right to left (positive scores) (Figure 4.5c). With, high magnitude negative scores associated with an opposite pattern (Figure 4.5d).

There was a Group*Reach*Muscle interaction ($p<0.001$) in the abdominal sites. Older adults had less response to the right lateral flexion moment (lower absolute PC2 score magnitudes) than younger adults for LEO2 in both reach distances (Table 4.3), and for REO1, and LEO3 in maximum reach (Figure 4.6b). Progression to maximum reach resulted in a greater left lateral flexion moment response (low PC2) for the LIO in both groups, and the LEO1 in younger adults (Figure 4.6b). This feature differentiated the muscle asymmetries between left and right (bilateral) sites in older and younger adults. While both groups had PC2 score differences between sides for IO and EO3, only younger adults had bilateral asymmetries between EO1 and EO2 in maximum reach (Figure 4.6b). While older adults had no asymmetry in PC2 scores for EO1, post hoc analysis between muscle sites suggested a reversal in the right-left EO1 relationship compared to younger adults during maximum reach. This was implied from the between group difference in LEO1 scores, and no difference between ipsilateral EO1 and EO3 sites in older adults, whereas these sites differed in younger adults (Table 4.3).

In the back sites there was a Group*Muscle ($p<0.001$) interaction where older adults had a greater response to the lateral flexor moment (higher absolute PC2) for all back sites except LL33, and LL52 (Figure 4.7b). A Reach*Muscle ($p<0.001$) interaction, captured the response to the lateral flexion moment for all back sites except RL52 with increasing reach distance (Figure 4.7c). Comparing muscle sites for all groups and reach distances, L16 had the highest response to the lateral moment, and L52 was the least responsive (Table 4.4). Reach*Muscle post hoc showed that increasing reach L13 increased and L48 decreased its response relative to other back sites (Table 4.4). This differed from the Group*Muscle interaction, suggesting older adults have an increase in the lateral flexion response of L48 response relative to other back sites (Table 4.4).

4.3.5 Principal Component 3

PC3 captured just over 1% of the total variance. This pattern was similar in shape to PC1. However, it captured a differential between muscle activation during the initial and final 10% with activity during hand transition corresponding with the increased flexion moment (Figure 4.5e). Positive scores were associated with an increased response to flexion moment; whereas negative scores were associated with more constant activation (Figure 4.5f).

In the abdominal sites, there was a Group*Reach*Muscle interaction, where older adults had less response to the flexion moment (negative PC3 scores) compared to younger adults for all sites except EO3 (Table 4.4). For both groups and reach distances IO had activation that differed from (lower PC3 scores) the other abdominal sites (Figure 4.6c). The relationship between RA and EO sites varied between groups and reach distances (Table 4.4). In general, RA sites had higher scores than most EO sites with exceptions for group differences illustrated in Table 4.4. Comparing paired sites, only younger adults had R>L asymmetries in EO1 and IO (Figure 4.6).

For the back sites there was a Group*Reach*Muscle interaction. Older adults had lower scores than younger adults for 7/24 back sites comparisons, with four in maximum reach (Figure 4.7e). Increasing the reach distance decreased PC3 scores for 3/12 back sites in older adults only (Figure 4.7e). While numerous site differences existed between back sites (Table 4.5), in general medial sites had more response to the flexion moment (higher PC3 scores) than lateral sites.

4.4 DISCUSSION

The aim of this study was to determine whether healthy older adults have altered trunk muscle activation patterns during a controlled functional dynamic lift and replace task compared to a younger group matched for sex, height, and body mass. A principal component analysis captured features from EMG waveforms at multiple trunk muscle sites during the controlled dynamic transfer task. Three principal components captured the amplitude and temporal characteristics of the EMG waveforms from twenty-four muscle sites explaining over 97% of the waveform variability. Statistical analysis of these features allowed for the comparison of group, and reach effects within and between muscle sites [Ivanenko et al, 2004]. Since trunk motion was similar between older and younger adults, the results of this study identified that healthy aging altered trunk muscle activation characteristics supporting age-related changes in the ability to control joint motion and spinal stability. The results of this study suggest that older adults can independently recruit their trunk muscle fibers. However, older adults differed from younger adults in neuromuscular response including an increased overall amplitude, decreased temporal variation, and altered synergies for specific trunk muscle sites. This study showed that specific group differences were not explained by increase physical task intensity in older adults, related to their decreased absolute strength [Hasue et al, 1980; Kubo, 1994; Sinaki et al, 2001].

4.4.1 Older Adults Have Similar Trunk Motion to Younger Adults

For the lifting task conditions, there was minimal trunk and pelvis motion reflecting that both groups were able to minimize motion as per the instructions. Older adults had the greater difference from younger adults, for axial rotation during the lifting phase for both trunk and pelvis. However, differences were no greater than 1.6 degrees. Thus, it would have minimal effect on the main moments of force (i.e. sagittal and frontal planes). Furthermore, the increased motion in all directions with increased task intensity was less than 1 degree. While, the significant differences in trunk and pelvis motion indicates age-related differences in the ability to control spinal motion, the low variability associated with the control placed on the task contributed to these statistical differences. It is likely that less than 1 degree rotation has little clinical/practical meaning. This is demonstrated using participants' anthropometric data to express thorax and abdomen mass and length as a percentage of total body mass [Dempster et al, 1959], and % stature [Plagenhoef, 1983]. Using this model the additional moment from 1 degree of lateral flexion or flexion would produce, at most, an additional external moment of .3Nm. This would represent a 1% increase to the flexion and lateral flexion moment previously reported for the maximum reach dynamic lift and replace task [Hubley-Kozey et al, 2012]. Therefore, additional trunk motion in older adults would have minimal effect on EMG interpretation through changing external moments or EMG-force relationship differences [Brown and McGill, 2008b].

4.4.2 Ability to Selectively Recruit Trunk Muscles to Changing Task Demands

The significant muscle interactions found in all PC scores for both abdominal and back muscle sites suggests that the various trunk muscle sites have unique response to the changing external moment generated by the lift and replace task as previously shown for younger adults [Butler et al, 2010; Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012]. Notable differences include IO sites differing from most abdominal sites in PC1, 2, and 3. Similarly, in the back, the superficial fibers of the multifidus (L52) differed from most other back sites for these PC scores. The unique activation patterns of these sites are consistent with their theoretical role as local stabilizers [Arjmand et al, 2008b; Bogduk et al, 1992; Ward et al, 2009]. In other studies, medial and lateral trunk muscle sites had differential recruitment in younger [Butler et al, 2010; Hu et al, 2011; Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012; Kavcic et al, 2004; Song, 2004; Talebian et al, 2010; Vera-Garcia et al, 2010] and older adults [Hubley-Kozey et al, 2009b; Quirk and Hubley-Kozey, 2012]. Overall, these findings are consistent with models that suggest trunk muscles are recruited relative to their mechanical role [Bogduk et al, 1992; Brown and Potvin, 2007; Dumas et al, 1991; Mirka et al, 1997] to generate a net external moment while attempting to i) maintain a necessary level of stability [Brown and Potvin, 2005; Rashedi et al, 2010], and ii) minimize joint forces and muscular fatigue [Arjmand et al, 2008a; Brown and Potvin, 2005; Rashedi et al, 2010].

Consistent with these models, an increasing external moment would scale trunk muscle activity such that the relationship between trunk muscle sites would differ with agonist having the greater increase in activation. This was shown in the present study for all PC scores had a reach by muscle interactions, identifying that both older and younger adults adjust the recruitment of trunk muscles sites dependent on their mechanical advantage. This is consistent with studies using increasing static and dynamic spinal loads in younger adults [Butler et al, 2009a; Butler et al, 2009a; Butler et al, 2010; Perez and Nussbaum, 2002; Sheikhzadeh et al, 2008; Song, 2004; Talebian et al, 2010]. While others showed that older adults increase muscle activation around the ankle in a non-proportional fashion to increasing task intensity [Hoffren et al, 2007; Hortobagyi and DeVita, 2000], this is the first study to show that older adults scale the activation of the numerous trunk muscle sites dependent on their mechanical advantage to increasing external moments. These findings provide evidence that older adults have a preserved ability to selectively recruit and scale trunk muscle activation patterns. This is an important mechanism for optimal joint loading [McGill et al, 2003a]. However, group interactions suggest that older adults compared to younger adults have different amplitude, and temporal recruitment for specific trunk muscle sites, with some of these group differences altering the synergies between these trunk muscle sites.

4.4.3 Older Adults have Increased Agonist Activation and Antagonist Co-activation for Trunk Muscles

At both reach distances older adults had higher overall neural drive than younger participants did (PC1 scores) in all muscle sites. Higher activation for the back extensors sites (PC1) (Table 4.4, Figure 4.7a&b) is consistent with other studies finding that older adults have increased agonist activation during movement tasks [Asaka and Wang, 2008; Kuo et al, 2011]. This can be explained by reports of 45% decreases in back extensor strength when comparing 70 to 30 year olds [Hasue et al, 1980; Kubo, 1994; Sinaki et al, 2001]. As the primary agonist of this task, all back extensor sites increased their overall activation (PC1) with increasing task intensity (Figure 4.7a) [Butler et al, 2009a; Butler et al, 2010; Perez and Nussbaum, 2002; Song, 2004]. Therefore, group differences in overall neural drive of back extensor sites can be explained by the increased relative task demands resulting from age-related strength loss.

Unlike the clear agonist back extensors, the abdominal sites have less defined patterns. Specific muscle fibers within the external oblique (EO2 and 3) can act as an agonist/synergist to the lateral flexion moment [Brown and Potvin, 2007; Dumas et al, 1991] applied to the spine at the beginning and end of the experimental task (Figure 4.1d). Furthermore, the horizontal fibers of the oblique muscle (EO1 IO) can act as stabilizers to minimize trunk axial rotation produced by the EO2 and EO3 [Arjmand et al, 2008a; Mirka et al, 1997]. The ability of the oblique sites to produce moments in these planes supports their increasing neural drive with increasing task intensity (Table 4.3 & Figure 4.6a). In general, all abdominal muscles can produce a flexor moment serving an antagonistic role to the lifting task studied [Arjmand et al, 2008b; Dumas et al, 1991;

Vera-Garcia et al, 2006; Vera-Garcia et al, 2006]. The higher overall abdominal activation (PC1 scores) for older adults in all sites supports more antagonist co-activation (Table 4.3, Figure 4.6a). This finding is unlikely influenced by increasing task intensity as neither group had a systematic activation increase in all abdominal sites with increasing reach distance (Table 4.3). Overall this finding is consistent with other studies around both the trunk [Allman and Rice, 2002; Asaka and Wang, 2008; de Freitas et al, 2010; Kuo et al, 2011; McGill et al, 1999; Quirk and Hubley-Kozey, 2012], and other joints [Hoffren et al, 2007; Hortobagyi and DeVita, 2000; Hortobgyi et al, 2011] suggesting that older adults have greater antagonistic co-activation.

Reasons for increased antagonist co-activation could include the 35% decline in abdominal strength when comparing 70 and 30 year olds [Hasue et al, 1980; Kubo, 1994; Sinaki et al, 2001]. However; since increased antagonist activation is known to increase spinal stiffness in younger adults [Arjmand et al, 2008b; Brown and Potvin, 2005; Granata and Marras, 2000; Vera-Garcia et al, 2006], increased antagonist activation in older adults supports computer optimization models that suggest neural drive to the spine is partially explained by maintaining a stiffness requirement [Brown and Potvin, 2005; Rashedi et al, 2010]. Beyond meeting a hypothetical minimum stiffness requirement, older adults relative to younger adults have been shown to have exaggerated stiffness around the knee and ankle [Hoffren et al, 2007; Hortobagyi and DeVita, 2000]. Thus, older adults may increase antagonist co-activity to increase spinal stiffness beyond that used in younger participants.

4.4.4 Older Adults have Altered Temporal Activation Patterns of Trunk Muscles

Principal component 1 reflects the overall neural drive and temporal activation of EMG patterns. Subsequent PC scores explain temporal features of EMG waveforms. Consistent with previous studies PC2 captured the response of trunk muscles to the changing lateral flexion moment [Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012]. For a majority of back sites older adults had greater response to lateral flexion moments (PC2 scores) compared to younger adults (Table 4.4 Figure 4.7d) and consistent with previous work, maximum reach distance increased the response of back sites (Table

4.4 Figure 4.7c) [Butler et al, 2010]. Thus, older adults' higher PC2 scores in the back likely represent decreased lateral flexor strength requiring increased activity of back extensors to respond to the lateral flexion moment generated around the spine at the beginning and end of the transfer task.

Lateral external oblique muscle sites (EO2 and EO3) had less responsiveness to the lateral flexion moment in older adults. This finding was supported by older adults having lower magnitudes for PC2 scores compared to younger adults for LEO2 and LEO3 (Table 4.3 & Figure 4.6a). Furthermore, older adults had no difference between REO2 and RA a site with minimal response (near zero PC2 scores) to the lateral flexion moment (Table 4.3, Figure 4.3a, & Figure 4.6b). This group difference is not a feature of increased task intensity as progression to maximum reach increased the magnitude of PC2 scores in EO2 and 3 causing them to differ from RA sites in younger adults. The lateral and posterior fibers of the external oblique (EO2 and EO3) participate in generating lateral flexor moments [Dumas et al, 1991]. In older adults, lower magnitude PC2 scores would either suggest a reduced burst of activation during the beginning of RHT or end of LHT or less differential between these two phases. Interpretation of the ensemble average EMG waveforms (Figure 4.3b,e,f) suggest that for the abdominal sites older adults utilized a more continuous activation of these lateral flexor sites suggesting more antagonist co-activation between these left and right sites which produce opposite lateral flexor moments.

Consistent with previous studies activation of trunk muscles to the flexion moment generated during HT was captured by PC3 [Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012]. In most abdominal sites (Table 4.3 & Figure 4.6c) and specific back sites (Table 4.4 & Figure 4.7e), older adults had lower (negative) scores than younger adults suggesting more constant activation that did not modulate with the increasing flexion moment. Depending on the muscle site analyzed, this had a different context. For the abdominal sites, the high PC3 scores would suggest younger adults temporally co-activated their antagonist muscle sites to increase spinal stiffness [Brown et al, 2006; Stokes et al, 2000a] during a period of greatest back extensor activation and flexion moment. This strategy differed from older adults who had constant activation of abdominal sites (lower PC3 scores), perhaps suggesting a shift towards a continuous abdominal stiffness response. In the back muscles, lower PC3 scores for older adults indicated a reduced ability to modulate the activity of medial back extensors (L13, and L33) to the flexion moment generated by the task. These group differences are not likely explained by increasing physical task demands, for increasing reach distance had no difference on PC3 scores in younger adults (Table 4.4 & Figure 4.7e).

Continuous temporal activation of abdominal (PC2 and 3) and back (PC3) muscle sites in older adults is consistent with previous studies for older adults [Hubley-Kozey et al, 2009a; Quirk and Hubley-Kozey, 2012] and interestingly for individuals with chronic low back during performance of a therapeutic exercise [Hubley-Kozey and Vezina, 2002]. Such changes are consistent with individuals with a sub-acute low back injury performing the same lifting task [Hubley-Kozey et al, Submitted]. The similarities in the recruitment patterns between these two populations' could be the result of common mechanisms. Both older adults and individuals with low back pain have decreased proprioception [Goldberg et al, 2005; Lee et al, 2010] and delayed onsets of trunk muscles to unanticipated perturbations [Allman and Rice, 2002; de Freitas et al, 2010; Hodges, 2001; Hwang et al, 2008; Silfies et al, 2009a]. This alteration in the proprioceptive system combined with delayed spinal reflex responses suggest both populations have a reduced capacity to fine tune trunk muscle activity in response to dynamic moment demands. To compensate for this change these populations rely on recruitment patterns that globally increase spinal stability and postural control [Stokes et al, 2000a; Vera-Garcia et al, 2006] utilizing more continuous activation of agonist, and antagonist muscles [D'hooge et al, 2012; Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2009a; Jacobs et al, 2011; Quirk and Hubley-Kozey, 2012; Silfies et al, 2005]. This strategy may compensate for both reduced feedback information, and reduced passive spinal stiffness in older adults that would be the result of joint space narrowing [Hangai et al, 2008; Hicks et al, 2009] increasing joint laxity in the neutral zone [Gallagher et al, 2007].

The implication of the combined effect of continuous muscle activation, and increased agonist and antagonist activation [Hubley-Kozey et al, 2009a; Quirk and Hubley-Kozey, 2012] can be increased spinal stability (resistance to motion) [Brown et al, 2006; Stokes et al, 2000a]. This mechanism however has a potential cost of greater cumulative loading [Granata and Marras, 2000; Vera-Garcia et al, 2006] and increased risk of trunk muscle fatigue [Yassierli et al, 2007]. Both are risk factors for future low back injuries [Davidson et al, 2009; Davidson et al, 2011; Norman et al, 1998]. However, age-related changes result in a non-uniform decline in strength [Hasue et al, 1980; Kubo, 1994; Sinaki et al, 2001], cross sectional area [Anderson et al, 2012a; Hicks et al, 2005b; Ikezoe et al, 2012; Ota et al, 2012] and muscle quality [Anderson et al, 2012b; D'Antona et al, 2003; Hicks et al, 2005a] in various fibers of the abdominal and back extensor muscles. Thus, conclusions that age-related changes in neuromuscular activation patterns would alter spinal loading or spinal stiffness require detailed 3-dimensional modeling of the spine, which are beyond the scope of this study.

4.4.5 Older Adults have Altered Trunk Muscle Synergies

There were other notable differences between older and younger adults when comparing muscle site recruitment. Specifically there was an unexpected change in the role of EO1 sites between younger and older adults in maximum reach. Consistent with previous work when analyzing the lateral flexor response (PC2), younger adults used a recruitment pattern where EO1 shared a temporal synergy with the horizontal fibers of the IO [Hubley-Kozey et al, 2012]. Unique to this study was the finding that older adults EO1 fibers did not share this temporal synergy but rather shared a synergy with the posterior more vertically oriented EO3 fibers. The explanation for this temporal EO1/IO synergy in younger adults may be explained by their moment production abilities. For example during the first 10% of the task REO1 activity produces a left axial rotation moment, stabilizing the right rotation produced by the left lateral (LEO2) and posterior fibers (LEO3) of the external oblique [Arjmand et al, 2008a; Dumas et al, 1991], and the RIO [Arjmand et al, 2008a]. The contribution of back extensors would be minimal as these fibers have minimal contribution to axial rotation [Macintosh et al, 1993]. Considering which planes the abdominal muscles can produce moments, the older adults lower LEO2/3 PC2 scores (Table 4.3 & Figure 4.6b) could produce less rotation to the right. Thus, there would be less necessity to produce a corrective REO1 rotational moment. However, in this highly controlled task, older adults had slightly more pelvis and trunk axial rotation during RHT (Table 4.2). In studies with less instruction to control trunk motion it has been found that older adults produced trunk movement in undesired planes (particularly axial rotation) [McGill et al, 1999; Van Emmerik et al, 2005]. Recent work suggests decreased control of trunk axial rotation during gait is a predictor of fall risk in older adults [Doi et al, 2013]. The results of the present study suggest that this undesired motion may in part be the result of an inability to fine tune specific temporal synergies of muscle fibers. Future studies should investigate if changes in ipsilateral LEO1 fibers synergy participates in uncontrolled trunk rotation, either by an reduced ability to minimize rotational moments produced by other fibers (ipsilateral LEO2 and 3) or the ipsilateral LEO1 acting synergistically with these sites producing an undesired right rotational moment.

Another age-related change in muscle synergies was the increased utilization of the superficial fibers of the posterior quadratus lumborum (L48 site) in response to the lateral flexion moment (PC2). Interactions for PC2 scores suggest that older adults utilized this muscle with a similar response as the inferior iliocostalis sites (L36). This differed from younger adults where L48 was less responsive to the lateral moment compared to the inferior longissimus (L33) and illiocostalis (L36) (Table 4.4). Increased L48 involvement could be an attempt to distribute lateral flexor moment across agonist sites [Park et al, 2012]. However, analyzing the effects of increasing task intensity revealed that L48 PC2 scores were equivalent to the erector spinae (L33 and L36) in normal reach, however, progressing to maximum reach the L48 had lower PC2 scores relative to these sites (Figure 4.7c). These findings suggest that older adults had less ability to separate responses between lateral back extensor sites. Increased recruitment of the L48 would not be an advantageous strategy as this site has relatively small contribution to lateral flexor moment production [Phillips et al, 2008], yet the fibers are orientated in such a way that they may produce more lateral shear to the spine, in older adults.

Changes in fine-tuning of muscle sites such as the EO and lateral back extensors could be the result of diminished spinal reflexive activation in older adults as identified at other joints [Granacher et al, 2006; Kido et al, 2004a; Obata et al, 2010]. Trunk muscles exhibit directionally specific reflexive activity to restore balance [Masani et al, 2009] and this mechanism could explain why EO1 activation restores the balance of rotational moment minimizing spine motion in younger adults. However, as the reflex response of trunk muscles were not measured in this study this theory is not conclusive. Hence other explanations for the changes in recruitment in EO and L48 sites should be examined. For example, the more diffuse motor cortex activation noted in older adults exhibited during voluntary movements [Van Impe et al, 2011b] could cause activation of motor units in proximity to their neighboring fibers.

4.4.6 Limitations

A potential limitation exists in the use of expressing neural drive in units of maximum normalized EMG. A concern is whether older adults produce a non-maximal normalized EMG value that would bias their results compared to a younger population [McGill et al, 1999]. Previous studies have indicated that older adults did not report discomfort while performing trunk maximum voluntary contractions [Hanada et al, 2008]. At other joints, older adults have the same ability to maximally activate their muscles as younger adults [Klass et al, 2005], and the efficacy of an individual to produce a maximum voluntary contraction can be maximized with practice trials prior to the testing day [Klass et al, 2007]. As the older adults in this study practiced all experimental tasks including maximum voluntary contraction, it was assumed that they obtained a similar level of voluntary recruitment as younger participants. However, the overall amplitude measure does not tell the entire story as the qualitative analysis of EMG waveforms show that the increased activation was not uniform throughout the task for all muscle sites as illustrated in Figure 4.3 and 4.4, suggesting that there was not a systematic bias. In addition, these non-uniform differences in part were captured by the statistical findings for PC2 and PC3 scores. Finally, this issue would only effect the interpretation of PC1 scores that capture differences in overall neural drive. Subsequent PCs explaining changes in temporal patterns would not be influenced by a systematic bias in neural drive.

A second limitation of this study is the low level of variance (1.2%) in the EMG waveforms explained by PC3. This low level of variance is in part due to the predominance of explained variance captured by PC1 including large amplitude differences between agonist and antagonist, and the strength differences between older and younger adults resulting in high variation of amplitudes. If analysis were performed using normalized to average EMG values [Davidson and Hubley-Kozey, 2005; Hubley-Kozey and Vezina, 2002] this feature would explain more variance in the EMG waveforms. However, as one of the research questions were to determine if older adults had increased overall activation and increased co-activation, it was necessary to use EMG data normalized to a physiological maximum [Burden, 2010a].

A final limitation in differentiating among muscle sites is the potential for crosstalk. Despite no between group differences in mass, older adults had higher waist circumferences than younger adults, a measurement known to be correlated with abdominal subcutaneous fat [Kanehisa et al, 2004]. As crosstalk increases with increasing subcutaneous body fat [Solomonow et al, 1994; Winter et al, 1994] it is possible that some cross talk was present. However, numerous precautions minimized the potential risk of cross talk. Electrode placement guidelines were fused to position electrodes to maximize the distance of surface electrodes from the borders of adjacent muscle fibers. In the abdominal sites these guidelines typically resulted in surface electrodes being >5cm from adjacent sites, a distance outside of the signal detection area of surface electrodes [Fuglevand et al, 1992]. For the back sites electrode placement were closer, about 2cm, from one another representing a distance that would only account for 5% of the capture volume power. However, differential EMG waveforms, and PC scores of medial and lateral back extensor sites suggest these sites did contain a majority of unique signal from the muscle fibers of interest. Finally, validation exercises were performed to ensure electromyographic activity was recorded from muscle site of interest [Winter et al, 1994]. In light of these precautions, the only age-related difference potentially confounded by cross talk would be an over estimation of the role of the quadratus lumborum (L48) through signal contamination from L36. However, use of a hip hike validation to confirm L48 electrode placement does suggest this site did record activity from the underlying fibers of interest.

4.5 SUMMARY

The purpose of this study was to investigate if there are age-related changes in trunk muscle recruitment while performing a functional lift and replace task. This study confirmed that healthy older adults have altered trunk neuromuscular activation patterns compared to younger adults and specific differences were not all accounted for by changes in task intensity only. Consistent with the hypothesis, older adults had higher agonist and antagonistic activation. They also demonstrated continuous activation levels despite changing flexion and lateral flexion moments throughout the task in specific muscle sites. Differences in the temporal response of specific muscle sites confirm the third hypothesis that healthy aging altered trunk muscle synergies.

CHAPTER 5

DOES A DUAL-TASK PARADIGM DURING A CONTROLLED FUNCTIONAL TRANSFER TASK PERFORMED BY HEALTHY OLDER ADULTS ALTER TRUNK NEUROMUSCULAR ACTIVATION PATTERNS?

5.1 INTRODUCTION

Older adults have altered trunk muscle recruitment patterns compared to younger adults to complete dynamic movement tasks [Hubley-Kozey et al, 2009a; Kuo et al, 2011; McGill et al, 2003a; Quirk and Hubley-Kozey, 2012]. These differences can be partially explained as a compensation for age-related changes in the tissues representing the passive [Hangai et al, 2008; Iida et al, 2002a], active [D'Antona et al, 2003; Kubo, 1994], and neural sub system [Goble et al, 2010; Kido et al, 2004a; Shaffer, 2007; Taube et al, 2008] essential for maintaining joint stability. In particular, older adults have increased antagonist co-activation [Asaka and Wang, 2008; McGill et al, 1999; Quirk and Hubley-Kozey, 2012], and more continuous activation amplitudes for both agonist [Hubley-Kozey et al, 2009a; Quirk and Hubley-Kozey, 2012] and antagonist sites [Quirk and Hubley-Kozey, 2012] relative to changing external moments. Both mechanisms are known to increase spinal stiffness in younger adults [Brown et al, 2006; Granata and Marras, 2000; Stokes et al, 2000a; Vera-Garcia et al, 2006], as well as increase knee and ankle joint stiffness in older adults to an extent that it exceeds the stiffness used in younger adults performing the same task [Hoffren et al, 2007; Hortobagyi and DeVita, 2000]. Older adults' reliance on increased active stiffness could mitigate joint motion caused by unexpected perturbations [Brown et al, 2006; Stokes et al, 2000a; Vera-Garcia et al, 2006]. These strategies however have a cost of increased joint compression [Brown et al, 2006; Granata and Marras, 2000; Vera Garcia et al, 2007; Vera-Garcia et al, 2006], and muscular fatigue [Yassierli et al, 2007]. Therefore, older adults exhibit a paradox where a mechanism known to minimize spinal motion reducing immediate risk of low back problems, results in a long term consequence of increasing risk of low back problems by increased joint loading and muscular fatigue [Davidson et al, 2011; Norman et al, 1998; Sanchez-Zuriaga et al, 2010]. This paradoxical situation may explain why older adults have increasing incidence of low back injuries [Lacroix et al, 2006].

Age-related differences in neuromuscular activation patterns can be partially explained by functional changes in the neural sub system. With increasing age reduced speed [Kido et al, 2004a; Rivner et al, 2001] and quality [Goble, 2009; Shaffer, 2007] of feedback information results in decreased joint position sense [Goble et al, 2011; Goldberg et al, 2005; Verschueren, 2002]. The combination of delayed sensory afferents, and interneuron changes in sub-cortical structures result in older adults having minimized spinal reflex contribution for joint control [Granacher et al, 2006; Kido et al, 2004a; Obata et al, 2010]. In an effort to attenuate the detrimental effect of these changes, neuroimaging studies show that older adults have altered cortical activity, which may serve a compensatory function to these peripheral changes. In particular older adults have, increased (greater BOLD or electrical response), and diffuse (areas recruited) activation of cortical structures in the frontal, somatosensory [Goble et al, 2012; Heuninckx et al, 2008; Van Impe et al, 2011b; Yordanova et al, 2004], premotor cortex , and posterior parietal cortex [Inuggi et al, 2011; Levin et al, 2011; Van Impe et al, 2011b; Yordanova et al, 2004]. This suggests that older adults compared to younger adults utilize more cortical resources for feed forward activation of the motor cortex [Kandel et al, 2000]. The motor cortex of older adults also have increased, and diffuse cortical activation amplitudes compared to younger adults [Hortobagyi et al, 2009; Inuggi et al, 2011; Van Impe et al, 2011b].

Given the connection between motor cortex and trunk motor neurons[Tsao et al, 2011a; Tsao et al, 2011b] age-related changes in motor cortex activation may be linked to age-related changes in neuromuscular activation patterns observed in older adults. For example diffuse activation in the motor cortex could be associated with increased antagonist co-activity [Asaka and Wang, 2008; Hortobagyi and DeVita, 2000; Kuo et al, 2011; McGill et al, 1999; Quirk and Hubley-Kozey, 2012] and alterations in synergistic muscle activation (Chapter 4). Furthermore, continuous activation of the MC may be associated with the continuous activation patterns of older adults' trunk muscles sites during dynamic movement task [Hubley-Kozey et al, 2009a; Quirk and Hubley-Kozey, 2012]. Thus, cortical changes may reflect feed forward compensatory mechanism to alter trunk neuromuscular activation patterns in older adults to increase spinal stability. However, given age-related decreases in the quantity [Gow et al, 2012; Kido et al, 2010] and quality [Gow et al, 2012; Gunning-Dixon et al, 2009; Van Impe et al, 2011a] of cortical resources, age-related changes in cortical activation may not be a voluntary process but rather explained by a confounder that has an impact on trunk neuromuscular activation patterns. In particular diffuse cortical activation would also reflect increased relative cortical recruitment to compensate for loss of cortical tissue, and/or an impairment in neural inhibition [Bernard and Seidler, 2012; Hortobagyi et al, 2006; McGregor et al, 2011]. Such cortical changes may still be associated with the aforementioned neuromuscular activation pattern changes noted in older adults; however, these changes would not represent a voluntary effort to increase joint stability.

It is unknown to what extent age-related changes in cortical function participate in trunk neuromuscular control. Dual-task, involves an experimental paradigm where one completes a primary motor task while engaged in a secondary cognitive task. In these experiments, it is believed that both the cognitive and motor task compete for cortical resources, impairing the ability to perform either or both tasks [Huxhold et al, 2006; Lacour et al, 2008]. For a successful dual-task paradigm, both tasks must use sufficient cortical resources, to necessitate a competition for any shared cortical resources. This relationship results in dual-task paradigms exhibiting a U shape, where too much attention dedicated to a motor task can result in poorer motor task performance compared to performing the same task while simultaneously performing an easy cognitive dual-task [Huxhold et al, 2006; Li et al, 2012]. At the other end of the spectrum, as the challenge of the cognitive task increases, performance of the motor and/or cognitive tasks will decrease [Huxhold et al, 2006; Li et al, 2012]. Since older adults' have lower overall cortical resources, they have an earlier inflection point where the challenge of the motor or cognitive task can hinder dual-task performance [Huxhold et al, 2006]. This relationship provides a plausible explanation why older adults have decreased task performance, while younger adults may have no change or an improvement in task performance while participating in the same dual-task paradigm [Huxhold et al, 2006; Laessoe and Voigt, 2008; Li et al, 2012; Melzer et al, 2001]. Previous studies have shown that trunk motion during walking increases in older adults when they were engaged in a cognitive dual-task [Doi et al, 2011a; Doi et al, 2011b]. However, no study has determined if trunk neuromuscular activation patterns are altered during a dual-task paradigm. Around the ankle and knee joints dual-task modified neuromuscular control, with studies showing that older adults have reduced antagonist co-activation [Fraser et al, 2007; Li et al, 2012; Rankin et al, 2000], and anticipatory recruitment [Tsang et al, 2008] of muscle while engaged in a cognitive dual-task.

The purpose of this preliminary experiment was to determine whether a cortical challenge could change the trunk muscle activation patterns used by older adults while performing a highly controlled dynamic lift and replace transfer task. It was hypothesized that when older adults participated in a cognitive dual-task that they would have altered trunk muscle activation patterns including: i) decreased antagonist co-activation, and ii) less continuous trunk muscle activation.

5.2 METHODS

5.2.1 Participants

Participants were recruited via advertisements and electronic notices posted to various sources within the Halifax community. Prior to testing all participants signed an informed consent approved by the Health Science Research Ethics Board at Dalhousie University. Eight older adults volunteered to participate. Exclusion criteria included any known cardiovascular, neurological, cognitive, or musculoskeletal conditions, in addition participants were excluded if they experienced a fall or low back injury within the last year that required medical attention, or limited daily function.

5.2.2 Test Procedure

The test procedure for this study was consistent with that described in Chapter 4 including anthropometrics, health screening questionnaires, self-reported data and postural, neurological, and cognitive assessments [Folstein et al, 1975]. In brief, participants were screened over the telephone before participating in two experimental sessions. Two weeks after a practice session, all participants performed a highly controlled right-to-left transfer task, described in detail in both previous work [Butler et al, 2010; Hubley-Kozey et al, 2012] (Figure 4.1), and Chapter 4. Participants stood with their body midline aligned with the centre of the table that was height adjusted to their measured standing elbow height. The performed a lift, transfer and replace task using a mass of 2.9 kg with the arm in maximum reach [Butler et al, 2009a]. Time to complete the task was to a 4-second count and a tactile feedback sensor placed at the mid thoracic spine [Butler et al, 2010] provided feedback to minimize trunk motion. If the timing count was not adhered to or motion was detected (either visually by the tester or from the recorded event and motion traces), the trial was repeated. Time and quantitative motion was later determined from the event markers and electromagnetic motion sensor data (described below) to confirm observations.

To modify the cognitive demands of this task, participants performed the controlled dynamic right-to-left transfer task in either the simple condition outlined above, or while engaged in a cognitive dual-task. For the cognitive dual-task, the standard 5-second count "Ready-Set-Go-1-2-3-4-5" was replaced with audio presentation of numbers with a 1 second meter. These numbers represented a subtraction of 3 from the previously stated number. To ensure that participants were engaged in the subtraction from 3, participants were asked to recognize incorrect subtractions in the audio feed. For example in this dual-task "Ready-Set-Go-34-30-27-25-22" there were 2 error terms (30, and 25). For a trial to be correct participants had to adhere to motion and timing criteria outlined above, and provide a correct answer of the number of incorrect subtraction of 3's presented. Participants were required to complete three correct trials in both the simple and dual-task condition.

5.2.3 Normalization Procedure

Consistent with previous studies and Chapter 4 EMG amplitudes were normalized as a percentage of maximum voluntary effort contractions from eight isometric exercises using a standard protocol. These included, a restrained sit-up, resisted lateral bend (left/right), resisted trunk extension, resisted trunk extension with left/right rotation, and resisted seated rotation (left/right) [Butler et al, 2010]. For each exercise, participants attempted to maintain a constant effort for 3 seconds with a 2-minute rest between trails to minimize the risk of fatigue.

5.2.4 Surface EMG Data Collection and Processing

Following standard procedures, surface electrodes (Ag/AgCl, 10mm circular electrodes; Meditrace, Graphics Control Canada Ltd) were positioned in a bipolar configuration along the fiber orientation of 12 bilateral muscle sites outlined in both previous work [Butler et al, 2010] and in general methods of this thesis. Electrode placements were consistent with published guidelines, with slight changes in placement to accommodate individual anthropometry when necessary. All electrode placements were validated using a series of manual muscle tests [Kendall and McCreary, 1983; Vezina and Hubley-Kozey, 2000]. EMG signals were pre-amplified (500X) and further amplified using three AMT-8 EMG systems (band pass 10 – 1000 Hz; CMRR = 115db, input impedance 10GΩ; Bortec Inc., Calgary Alberta). Raw EMG signals and event markers were digitized at 2000 samples per second, using a 16 bit resolution National Instruments analog-to-digital conversion board (National Instruments, CA-1000) and Labview^{TM} software (version 7), and saved on the hard disk of a computer for post processing.

EMG signals were processed using a standardized protocol (Chapter 4) designed to condition signals (remove electrocardiogram trace, and electromagnetic noise) before producing a time and amplitude normalized linear envelope waveform for each trial. For each participant, the three trials within a condition were averaged to producing ensemble average waveform for each muscle site.

EMG ensemble average waveforms for each participant, muscle and condition were entered into a PCA model (Figure 4.2) as described previously in Chapter 4 and other publications [Hubley-Kozey et al, 2009a; Hubley-Kozey et al, 2012; Jackson, 2003]. Briefly, principal components captured key features within the EMG waveforms. These features could include amplitude patterns, amplitude differentials, or phase shifts. For each EMG waveform, a PC score was calculated indicating its similarity to the PC. Thus EMG waveforms with similar in shape and amplitude would share similar PC scores [Ivanenko et al, 2004]. Thus, quantitative statistical analysis was performed on the temporal and amplitude characteristics of EMG patterns by comparing the PC scores. To increase the stability of the model the waveforms of the eight participants were combined with nine additional older adults performing a single task maximum reach transfer resulting in PC's being created from a [600x101] data matrix.

5.2.5 Motion Capture

Electromagnetic (Flock of Birds^{TM} (FOB) sensors were placed superior to the left anterior superior iliac crest and over the T8 spinous process to record the angular motion of the trunk and pelvis throughout the task in 3D with respect to a global coordinate system [Silfies et al, 2005]. Participants were positioned such that sensor motion corresponded with anatomical planes of motion (x= Axial Rotation (AR), y= Flexion/Extension (FE), z= Lateral Bend (LB)). Maximum angular displacements were calculated for each sensor for each phase of movement.

5.2.6 Statistical Analysis

Descriptive statistics were calculated for the group and when appropriate for the conditions. Angular displacement data were compared using a two factor Repeated Measures Analysis of Variance (ANOVA) (Condition*Phase) model. PC scores for the abdominal and back muscle sites were used in separate two-factor Repeated Measures ANOVA (Condition*Muscle) models. Pair wise comparisons were made on significant effects using a Tukey simultaneous test. Normality was confirmed using a Kolmogorov-Smirnov test, with non-normal data being transformed using a Johnson Transformation. Statistical analyses were performed in Minitab (Minitab Inc, State Collage, PA, version 16), with α =0.05.

5.3 RESULTS

5.3.1 Participants Performance: Timing and Kinematics Variables

Eight older adults volunteered for this study with their demographics displayed in Table 5.1. The total time to complete the task was 4.1 ± 0.6s with no significant differences ($p<0.05$) between conditions. In addition to the total movement time, there were no significant differences between conditions in the time to complete each phase of the lift and replace task.

Overall trunk and pelvis motion were less than 3^{o} in all axis of rotation, suggesting that participants were successful at minimizing trunk motion. For both the trunk and pelvis in all axis of rotation there were no condition main effects or condition*phase interaction (Table 5.2), suggesting the cognitive dual-task did not affect movement performance. For all axis of rotation, there was a significant phase effect where right and left hand transfer had greater motion than hand transfer (Table 5.2).

5.3.2 Qualitative and Principal Component EMG Analysis

Qualitative analysis of the EMG ensemble average waveforms for abdominals and back sites (Figure 5.1) showed no trend of the cognitive dual-task modifying neuromuscular response. Three Principal Components captured over 98% of the total variance within the EMG waveforms.

Principal components generated using these data were consistent with previous work analyzing the spatial and temporal trunk neuromuscular recruitment patterns while performing the dynamic functional right-to-left lift, transfer and replace task [Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012]. PC1 captured the overall muscle activation pattern but in particular discriminated between sites with high or low overall EMG amplitudes (Figure 5.2a & b). PC2 captured the increasing trunk muscle activation at the beginning or end of the lift and replace responding to the changing lateral flexion moment during these phases (Figure 5.2c & d). PC3 captured additional trunk muscle response to the increasing flexion moment during hand transition (Figure 5.2e & f). All abdominal and back site PC scores had significant ($p<0.05$) muscle main effects (Table 5.3). There were no significant ($p>0.05$) condition main effects, or condition* muscle interaction (Table 5.3).

Table 5.1: Descriptive statistics for older adult participants in this study and older adults without a dual-task trial.

Comparison	Older Adult	Older Adult No Dual-task
Participants (number)	8	12
Age (years)	67.8 (2.5)	68.8 (3.5)
Sex (% Male)	87%	58%
Mass (Kg)	86.1 (15.2)	79.1 (14.8)
Height (cm)	173.4 (6.0)	170.0 (9.3)
BMI (Kg/m^2)	28.6 (4.4)	27.2 (3.5)
Waist Girth (cm)	98.5 (10.9)	91.2 (13.2)
Aerobic Training (sessions/week)	4.8 (3.0)	3.3 (3.0)
Abdominal Training (sessions/week)	1.8 (2.2)	1.4 (3.0)
Normal Abdominal Function (% complete)	100	58
Mini Mental State Score	29.4 (0.7)	29.4 (1.0)

Table 5.2: Motion data for older in single and dual-task for the phases right hand transfer (RHT), hand transition (HT), and left hand transfer (LHT)

	Trunk (^{o})			**Pelvis (^{o})**		
	Lat. Flex	**Flex.-Ext.**	**Axial Rot.**	**Lat. Flex.**	**Flex.-Ext.**	**Axial Rot.**
Single RHT	1.1 (0.6)	1.3 (1.2)	2.6 (3.0)	2.0 (0.9)	1.1 (0.5)	2.7 (0.9)
Dual RHT	1.0 (0.6)	1.1 (0.5)	2.5 (1.7)	1.7 (0.9)	1.2 (0.9)	3.1 (2.5)
Single HT	0.4 (0.4)^{ab}	0.5 (0.6)^{ab}	0.7 (0.4)^{ab}	0.6 (0.3)^{ab}	0.6 (0.3)^{ab}	1.1 (0.6)^{ab}
Dual HT	0.4 (0.2)^{ab}	0.4 (0.2)^{ab}	0.9 (0.6)^{ab}	1.0 (1.2)^{ab}	0.4 (0.2)^{ab}	0.9 (0.4)^{ab}
Single LHT	1.1 (0.8)	1.4 (1.7)	2.5 (1.7)	1.2 (0.6)	1.0 (0.4)	2.1 (1.1)
Dual LHT	1.2 (0.9)	1.4 (1.5)	2.5 (1.5)	1.3 (0.6)	1.3 (0.2)	2.8 (1.7)

Mean (SD)
Significant differences in Tukey HST Post hoc within the same task a=RHT>HT, b= LHT>HT

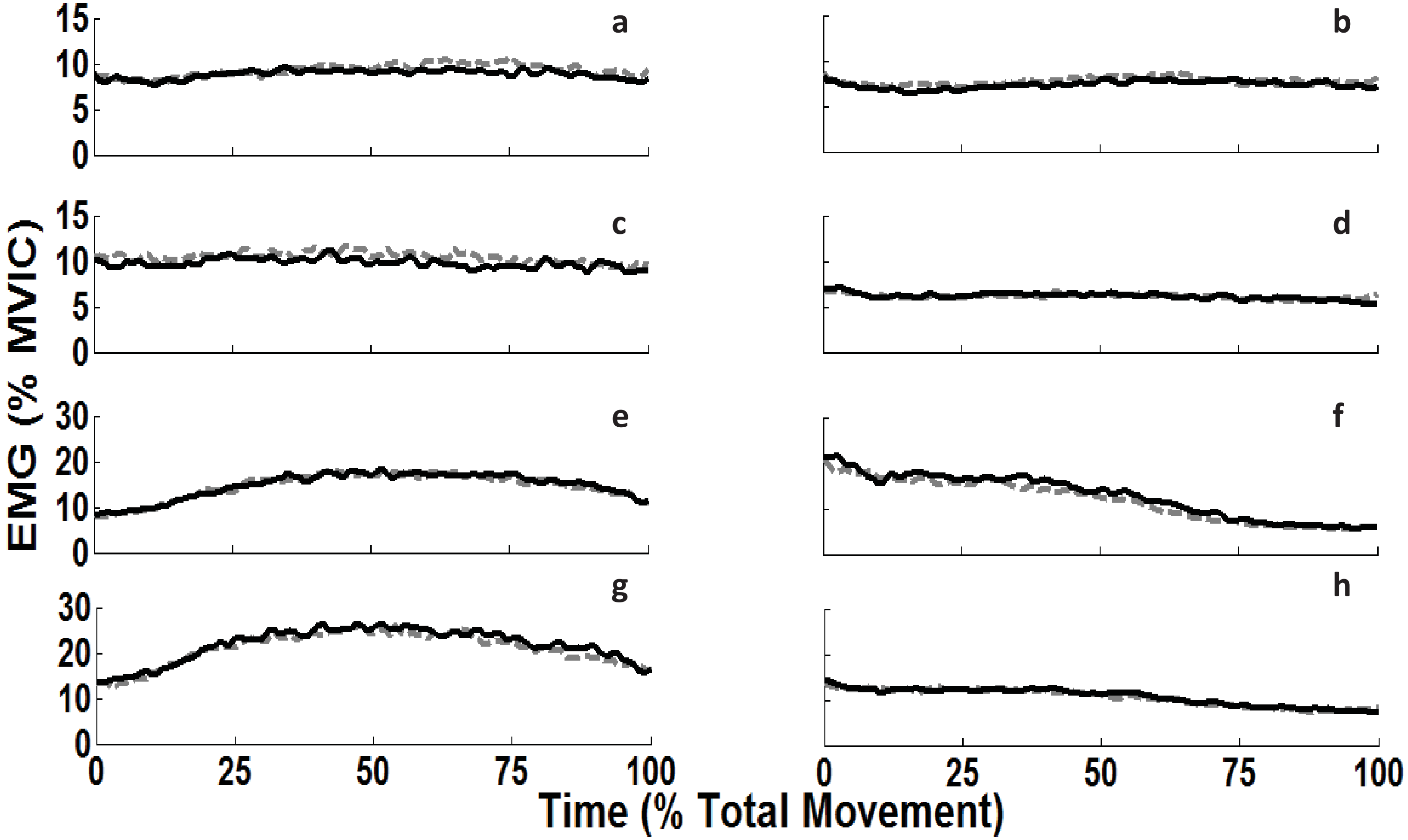

Figure 5.1: Examples of ensemble average waveforms for specific abdominal and back sites. Black lines= single task, grey dashed lines= dual-task. Specific trunk sites include left (L) and right (R), a) LEO2, b)REO1, c)LIO, d)REO3, e)RL33, f)LL16, g)RL52, h)LL48.

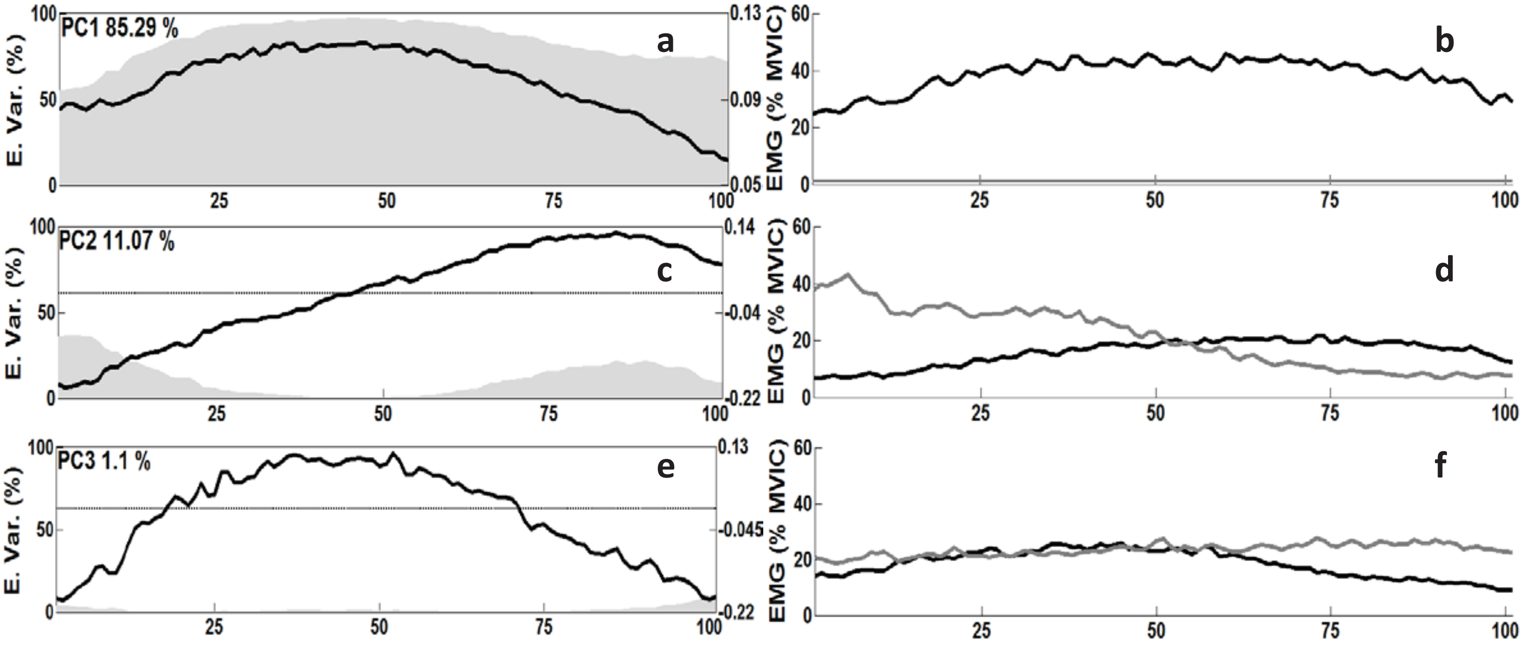

Figure 5.2: For principal components (1-3) plots a, c, and e show the principal component (black line), and explained variance relative to total movement time depicted by grey shading, Total explained variance for each principal component is shown in the top left corner of these plots. To assist with interpretation, for each principal component plots b, d, and f show ensemble average waveforms of EMG activation patterns representing the 5 highest (black line) and 5 lowest (grey line) PC scores.

Table 5.3: P-values from Repeated Measure ANOVA on Abdominals and Back Principal Component Scores

	Abdominals			**Back**		
	PC1	**PC2**	**PC3**	**PC1**	**PC2**	**PC3**
Cond	0.831	0.871	0.693	0.600	0.552	0.644
Muscle	<0.001	<0.001	0.029	<0.001	<0.001	0.000
Cond*Msl	1.000	0.997	1.000	1.000	1.000	0.996

5.4 DISCUSSION

The results of this study did not confirm the hypothesis that dual-task modified the neuromuscular activation patterns used by older adults while performing the dynamic right-to-left lift, transfer, and replace task. No change in neuromuscular response is consistent with findings from other joints comparing older adults engaged in dual-task conditions [Fraser et al, 2007; Hegeman et al, 2012; Li et al, 2012]. However, the results of the present study are inconsistent with others that did find neuromuscular patterns of older adults changed while engaged in a cognitive dual-task [Fraser et al, 2007; Li et al, 2012; Rankin et al, 2000; Simoneau et al, 2008]. Inconsistencies are likely accounted for by differences in study methodology since an infinite number of cognitive tasks and motor tasks can be combined for a dual-task.

The present study differed from other dual-task studies in two ways. First, the present study is the first to quantify trunk muscle activation patterns of older adults engaged in a cognitive dual-task. Second, other studies used continuous (example: gait, postural sway) [Doi et al, 2011a; Doi et al, 2011b; Fraser et al, 2007; Huxhold et al, 2006; Lacour et al, 2008; Li et al, 2012; Melzer et al, 2001; Simoneau et al, 2008; Van Impe et al, 2011b], or discrete motor tasks (example: perturbations or single movements) [Hegeman et al, 2012; Laessoe and Voigt, 2008; Rankin et al, 2000; Tsang et al, 2008] for a dual-task design, whereas the present study used a controlled serial transfer task. This motor task was chosen as it had known neuromuscular activation patterns that allowed us to elucidate if known age-related changes in these patterns interacted with cognitive resources. Despite methodological differences, three additional factors could explain why a cognitive dual-task had no effect on trunk neuromuscular activation patterns. These were: 1) the influence of dual-task is dependent on the complexity of both the motor and cognitive task performed, 2) the influence of a cognitive dual-task is dependent on the cognitive resources of participants, or 3) neuromuscular patterns are not engaged in the dual-task originating from a sub-cortical mechanism.

Dual-task paradigms disrupt motor and/or cognitive task performance through the competition of cortical resources [Huxhold et al, 2006; Lacour et al, 2008; Li et al, 2012].

However, this relationship depends on both tasks representing sufficient complexity to require overlapping cortical resources [Huxhold et al, 2006]. If either the cognitive or the motor task does not require competition for resources the performance of either task may not change, or improve in a dual-task paradigm. The subtraction of 3, dual-task paradigm used in the present study, has been established to alter neuromuscular function around the ankle in older adults [Rankin et al, 2000]. However, a subtle difference exists in the implementation of this dual-task paradigm. Rankin et al. used a continuous verbalized subtraction of 3, whereas the present study only required a single correct answer to ensure participants were engaged in the cognitive dual-task. Participants in the present study could have arrived at a correct answer without continuous processing. For example participants could have retained all 5 numbers and perform the calculation to arrive at the correct responses at the end of the trial. Despite this difference, older adults using a digit retention dual-task did have altered neuromuscular patterns [Simoneau et al, 2008]. This suggests that regardless of how participants arrived at the final answer either continuously or post trial this dual-task has the potential to alter neuromuscular performance around other joints and motor task. The data for incorrect trials were not retained, but it was observed that older adults had difficulties performing the dual-task paradigm. In general, older adults required more repeat trials in the dual-task condition compared to the single task condition. This would suggest that older adults did experience difficulties performing the dual-task.

When the challenge of a dual-task experiment is sufficient, task performance can decrease for either the cognitive task [Doi et al, 2011b; Hegeman et al, 2012; Melzer et al, 2001], the motor task [Fraser et al, 2007; Huxhold et al, 2006; Li et al, 2012; Simoneau et al, 2008], or both [Huxhold et al, 2006] tasks. The ability of dual-task paradigms to alter the performance of only one task represents task prioritization [Fraser et al, 2007; Li et al, 2012; Melzer et al, 2001]. In older adults, motor task prioritization in dual-task conditions is common [Fraser et al, 2007; Melzer et al, 2001]. In the present study, participants were instructed to perform both the motor and cognitive task correctly. However, these instructions may have caused older adults to prioritize the motor task. For example, the initial repeat trials in older adults were the result of motor errors. In particular, participants would have errors: transferring the mass between hands during the right time, stalling during the experiment, or having inappropriate timing with the position of the mass with respect to the external count. These trials were not retained because motor errors would change the external moment generated during this task, making comparisons of the neuromuscular activation patterns un-interpretable. An interesting observation was in later trials, dual-task trials were repeated because participants made errors in the cognitive task. These observations suggest that in later and correct trials older adults prioritized the motor task. However, as dual-task cost of cognitive performance was not measured, this conclusion cannot be made with the current data. To quantify this dual-task cost the latency of the correct answer should have been measured. This would allow for a measure of cognitive dual-task cost by comparing the latencies from the dual-task to those measured in the cognitive task alone. Future studies should retain these incorrect trials to capture motor errors, and consider a method to measure dual-task performance of the cognitive task to account for task prioritization.

A final limitation of this study was the small and biased sample of older adults. It is noted in dual-task paradigms that younger adults have no change or an improvement in task performance while engaged in a cognitive dual-task. This change is believed to occur from younger adults being less likely to experience a competition of resources to perform both tasks [Huxhold et al, 2006; Lacour et al, 2008]. In older adults those with less cortical volume experience more dual-task cost during walking compared to older adults with less cortical atrophy [Doi et al, 2011a]. Aging is not a homogenous process, and the rate of tissue change is determined by multiple factors, one of these factors includes physical activity. In general, physically active older adults differ from sedentary older adults with respect to brain atrophy [Gow et al, 2012; Gunning-Dixon et al, 2009] muscle co-activation [Carvalho et al, 2010] and proprioceptive function [Ribeiro, 2007; Ribeiro and Oliveira, 2010]. Functionally this translates to active older adults requiring less proportion of cortical resources to complete a motor task compared to sedentary older adults [McGregor et al, 2011]. Therefore, active older adults may have less dual-task cost compared to sedentary older adults. In the present study, older adults were active providing a self-report measure of engaging in over 30 minutes of continuous aerobic activity [Gilleard and Brown, 1994] on average 4.8 days a week. The participants of the present study had levels of physical activity, and proportion with normal abdominal function that were greater than the larger older adult group from Chapter 4 (Table 5.1). Furthermore, the physical activity levels of older adults in the present study are above the mean of older adults self-report of meeting Canadian activity guidelines (150 minutes of moderate activity/ week) [Canadian Fitness and Lifestyle Research Institute, 2009]. Although, a limitation of subjective measures of physical activity is potentially biased with adults performing less measurable physical activity than their self-reports, a weak correlation does exist between self-reports and direct measures [Tucker et al, 2011]. A second measure of the high function of the older adults in the present study, was that nearly all participants had a ceiling level of cognitive function assessed by the mini mental state exam (MMSE scores /30) [Folstein et al, 1975] (Table 5.1). These scores exceeded those reported in large studies of healthy older adults i.e. an average of 27±3 [Doi et al, 2011a]. These measures suggest that the interpretation of this study is limited by the high physical and cognitive function of its participants. Conversely, it supports that physically active older adults did not have neuromuscular pattern changes. Further studies should determine whether sedentary older adults or those with mild cognitive impairments would have altered neuromuscular function of trunk muscles while engaged in this cognitive dual-task. This information could further our understanding of aging, physical activity and cognitive function on neuromuscular activation patterns.

5.5 SUMMARY

The purpose of this study was to determine if a challenge to cognitive resources, while performing a controlled dynamic transfer task could alter trunk neuromuscular activation patterns in older adults. These data for the dual-task show that it had no effect on trunk muscle activation patterns in active older adults. However, this can be explained by two factors. First, the constraints of the lifting task may have caused the older adults to prioritize the movement task. Secondly, the physical activity level of the older adult participants was higher than average older adults, potentially minimizing the cortical resource competition between the cognitive and movement task. The results of this preliminary study provide evidence that older adults were challenged in the dual-task paradigm. Furthermore this study provides guidance for future studies aimed to explore cortical and physical resource competition in older adults. Specifically, future studies should i) include a measurable cognitive task to account for dual-task cost to determine task prioritization, ii) present different levels of cognitive task performance, iii) include less constraint to the motor task to ensure that participants have the option to modify their neuromuscular patterns, and iv) recruit a spectrum of older adults (particularly sedentary or those with mild cognitive impairments) to account for the heterogeneity of older adults.

CHAPTER 6 CONCLUSION

6.1 CONCLUSION AND SUMMARY

The overall aim of this thesis was to improve our understanding on how healthy aging modifies trunk neuromuscular activation patterns. Age-related changes to tissues involved in maintaining joint stability could result in older adults having reduced control of trunk motion (Figure 1.1). The ability of the neural system to control trunk muscle tension (neuromuscular system) to achieve a movement goal creates an association between trunk neuromuscular activation patterns and trunk motion. Despite this link, limited work has examined whether age-related changes in trunk neuromuscular patterns exist. In the few studies that have examined the trunk neuromuscular response in older adults, methodological issues existed that do not capture the complexity of the trunk neuromuscular system. Furthermore, these studies did not control for confounding factors, such as joint motion and relative physical task intensity, factors known to alter trunk neuromuscular activation patterns. Thus, the primary aim of this thesis was to establish whether older adults had altered trunk activation patterns during the performance of a dynamic movement task compared to younger adults. This thesis analyzed the time varying activation of multiple trunk muscle sites, thus capturing the spatial temporal complexity of the trunk, while using a controlled experimental designed to minimize the influence of known confounders. If age-related differences were identified in trunk neuromuscular activation patterns, predominant theories suggest that such differences in patterns are generated in the motor cortex to compensate for changes in passive and active tissues that contribute to spinal stability (Figure 1.2). Therefore, a secondary aim of this thesis was to determine if the neuromuscular activation patterns used by older adults were altered using a dual-task paradigm. To address these aims, two separate but related experiments were conducted. The key findings are described below:

6.1.1 Primary Objective: The Effect of Healthy Aging on Trunk Neuromuscular Activation Patterns

The primary objective was to determine if a group of older adults had altered trunk neuromuscular activation patterns compared to younger adults during a controlled right to left lift, transfer and replace task. To reduce confounding factors that would influence trunk neuromuscular function, joint range of motion, lifting velocity, and participant anthropometrics (sex, height, and mass) were controlled between the two groups. In this study, older adults had altered trunk neuromuscular activation patterns compared to younger adults with key differences summarized in Table 6.1. In particular, older adults had: i) increased overall activation for both agonist and antagonist, ii) less variable activation of trunk muscles to changing external moments, and iii) alterations in trunk muscle synergies. These results confirmed the hypothesis that older adults compared to younger adults would require altered trunk neuromuscular activation patterns to complete a dynamic functional transfer task with time and motion constraints. Older adults required greater overall activation thus they performed this task at higher relative physical task intensity compared to the younger group.

6.1.2 Sub-Objective One: Does Increasing Physical Task Intensity Explain Age-Related Differences in Trunk Neuromuscular Activation Patterns

Task intensity influences trunk neuromuscular activation patterns in younger adults. All participants were compared using two lifting conditions to change the external moment of the task. By comparing how older and younger adults altered their trunk neuromuscular activation patterns in response to increasing load, these data established if age-related differences in the primary objective could be accounted for by the higher relative physical task intensity. Secondly, these data established that the trunk neuromuscular activation pattern of these two groups responded differently to increasing physical task intensity. The key findings are summarized in Table 6.1. In particular, by comparing group differences in trunk neuromuscular activation patterns to differences explained by increasing reach distance, it was confirmed that older adults had altered trunk neuromuscular activation patterns including: i) increased antagonist muscle activation, ii) variable activation of trunk muscle to changing external moments, and iii) altered trunk muscle synergies. These findings confirmed hypothesis two that group differences in relative task intensity would not account for all age-related differences in trunk neuromuscular activation patterns.

6.1.3 Sub-Objective Two: Does Dual-Task Alter Trunk Neuromuscular Activation Patterns in Healthy Older Adults

Sub-objective two examined whether challenging cognitive resources altered the trunk neuromuscular patterns used by older adults. Older adults completed a controlled dynamic transfer task performed as a single task or simultaneously with a cognitive dual-task. It was hypothesized that altered neuromuscular activation patterns would be generated by a cortical mechanism requiring increasing cognitive resources. Thus, older adults would have altered neuromuscular activation patterns if these resources were challenged. To test this hypothesis a dual-task experiment examined whether competition of cortical resources would alter trunk neuromuscular patterns in older adults. The key results of this study did not support the hypothesis as the cognitive dual-task did not alter trunk neuromuscular activation patterns compared to a single task paradigm. While, the study sample was small, potentially explaining the lack of significant difference, the sample was also active compared to the average population. Previous research has found that physically active older adults have require less diffuse activation of cortical structures to complete motor task [McGregor et al, 2011] providing and explanation for the findings. However, the constraints involved in this highly controlled movement task may have caused older adults to prioritize the motor task explaining no change in neuromuscular activation patterns.

Table 6.1: Summary of key findings

Age-related Difference in Trunk Neuromuscular Activation Pattern	Neuromuscular Activation Pattern Explained by Increased Task Intensity?	Age-Related Difference in Neuromuscular Activation Pattern Interaction with Task Intensity
Increased Activation of Agonist Muscle Sites	Yes Increasing Task Intensity Consistently Increased Agonist Activation (PC1) for all Back Sites	No Both Groups Increase Agonist Activation
Increased Activation of Antagonist Muscle Sites	No Increasing Task Intensity did not Consistently Increase Antagonist Activation (PC1) of Abdominal Sites	Yes Only EO3 (Both Groups) and EO1(Older Adults) had Increased PC1 scores with Task Intensity
Continuous Activation/ Less Response of Abdominals to Lateral Flexion Moment	No Increasing Task Intensity Increased Abdominals Response to Lateral Flexion Moment (PC2)	Yes Abdominal Activity to Lateral Flexion Moment Increases in Younger Adults Only
Continuous Activation/ Less Response of Abdominals to Flexion Moment	No Increasing Task Intensity Increased the Response of Abdominals to the Flexion Moment (PC3)	Yes Abdominal Antagonist Co-activity During Flexion Moment Increases with Task Intensity in Younger Adults Only
Continuous Activation of Back Extensors to Flexion Moment	No Increasing Task Intensity did not Consistently Change Back Extensors Response to Flexion Moment in Younger Adults	Yes Back Extensor Activity Becomes More Continuous with Increasing Flexion Moment in Older Adults Only

Age-related Difference in Trunk Neuromuscular Activation Pattern	Neuromuscular Activation Pattern Explained by Increased Task Intensity?	Age-Related Difference in Neuromuscular Activation Pattern Interaction with Task Intensity
Changes in Ipsilateral EO1/EO3 Muscle Synergy in Response to Lateral Flexion Moment	No Increasing Task Intensity Results in Ipsilateral EO1/EO3 Difference in Younger Adults	Yes Older Adult Ipsilateral EO1/EO3 Synergy with Increasing Task Intensity Younger Ipsilateral EO1/EO3 Difference
Changes in L48/L36 Muscle Synergy in Response to Lateral Flexion Moment	No Increasing Task Intensity Decreased L48 Response to Lateral Flexion Moment Relative to L36 (PC2)	No Older Adults Increased the Activity of L48 to the Lateral Flexion Moment Relative to L36 (PC2)

6.2 IMPLICATIONS

The results of this thesis have two key implications. First, although, increased agonist activation was accounted for by higher task intensity, older adults use a higher percentage of their maximum muscle activity to complete the same task as younger adults. This increased agonist activity combined with higher antagonist activity, and more continuous activation of trunk musculature increase the risk of muscular fatigue in older adults when completing tasks with similar demands [Yassierli et al, 2007]. Fatigue is a risk factor for falling and low back pain [Davidson et al, 2011; Sanchez-Zuriaga et al, 2010]. A secondary implication of the age changes in trunk neuromuscular function include increased active stiffness and stability in the trunk [Arjmand et al, 2008b; Brown and Potvin, 2005; Brown et al, 2006; Kavcic et al, 2004]. Given that age-related changes in the size and quality of trunk muscles is dependent on the muscle group analyzed [Anderson et al, 2012a; Anderson et al, 2012b; Ikezoe et al, 2012], this study could not determine if the age-related differences in trunk neuromuscular activation patterns would result in older adults having altered joint loading or stability compared to younger adults.

A secondary key finding of this thesis was the age-related differences in trunk muscle synergies. In particular, older adults had an ipsilateral EO1 and EO3 synergy where both of these muscles would produce a contralateral trunk axial rotation moment. This differed from younger adults who had an ipsilateral/contralateral EO1 and EO3 synergy where these muscle sites would produce a counter moment to balance the resultant axial rotational moment generated around the trunk. Age-related differences in these synergies suggest that older adults may have a decreased ability to balance undesired moments. In a less controlled task, these differences may result in additional trunk motion in undesired planes, in particular the transverse plane. The implications of uncontrolled trunk motion was highlighted by a recent study showing that increased trunk axial rotation during gait is a risk factor to predict falls in older adults [Doi et al, 2013].

The results of the dual-task study suggest that healthy older adults had no change in their trunk neuromuscular activation patterns compared to patterns used in a single task. Given the small sample and high, self-reported activity levels of participants recruited for this preliminary analysis, there is too little, and potentially biased data to fully understand the implications. In these healthy older adults, the null finding may also reflect that this dual-task did not require sufficient cognitive load to create a competition for resources. Observations from this study did suggest that because of the constraints of this motor task, older adults did experience some dual task cost that may have resulted in motor task prioritization. However, cognitive task performance was not measured in this experiment, so these findings will need to be confirmed in future studies.

6.3 LIMITATIONS AND FUTURE DIRECTIONS

Collectively the findings in this thesis provide objective evidence that older adults alter trunk neuromuscular activation patterns compared to younger adults to complete the same controlled dynamic functional transfer task. While, considerable care was taken in study design some limitations existed.

6.3.1 Limitations

1) The cross-sectional design in Chapter 4 is limited in determining cause and effect. There is the potential that altered trunk neuromuscular activation patterns were caused by factors not controlled in this study. By matching younger and older adults by anthropometric characteristics, and using the exclusion criteria in this study, potential confounders in trunk neuromuscular activation patterns such as sex, physical task intensity due to limb mass, and history of low back injuries [Butler et al, 2010; Hubley-Kozey et al, Submitted; Hubley-Kozey et al, 2012; Jacobs et al, 2011] were minimized. However, longitudinal studies are better able to determine changes associated with aging, however this was not feasible given the time frame of this study.

2) The sample size of older adult participants in the dual-task experiment was small, although it is consistent with similar studies published in the literature [Van Impe et al, 2011; Simoneau et al, 2008; Tsang et al, 2009]. This occurred from low volunteer levels during the time it was feasible to collect data for this thesis. Thus, this thesis is limited to understand the implications of the dual-task experiment. A second limitation with the sample of older adults in both experiments is their relatively high activity level. It would be ideal to get participants that are more representative of the general population to

provide a larger spectrum of older adults. However, conclusions were made within the limitation of this sample and were not extrapolated beyond.

3) To explain if increasing task intensity could account for age-related differences in trunk neuromuscular activation patterns, altered reach distance increased the external moment of this task. The comparison of only two loading conditions is a potential limitation of this study since at neither reach did the EMG activation in younger adults approach the activation amplitudes of the older adults. Thus, whether higher loading intensities in younger adults would result in similar neuromuscular patterns as those used in older adults is not known. However, as the directionality of neuromuscular pattern changes explained by increasing physical task intensity were often the opposite of the differences found by between age groups, it is unlikely that increasing loads would alter the general findings of this study.

4) To assess if trunk neuromuscular patterns used by older adults could be changed by increasing the cognitive load, a dual-task paradigm was used to challenge the cognitive resources. However, in this study only a single level of cognitive dual-task was performed. There was no measurement of dual-task performance to the cognitive task. Despite this delimitation, participants did perform a correct dual-task trial. Measuring response time would add additional metrics to aid in interpretation of this study. With a measure of cognitive task performance, it would be possible to determine if the cognitive task was insufficient to evoke competition for cognitive resources. Secondly, this measure could account for features such as task prioritization.

6.3.2 Future Directions

The following sections will suggest five key future directions.

Future Direction 1

For feasibility of experimental design, some data were not recorded between participants such as trunk muscle strength. Instead, it was assumed that existing cross-sectional and longitudinal data could account for age-related declines in strength. However given an EMG to force relationship [Brown and McGill, 2008b] interpretation of trunk neuromuscular activation patterns would have been aided if a measure of trunk torque in various planes were included. Secondly, self-reports such as activity level could be biased. Future studies should use objective measures for the strength and aerobic characteristics of participants. Given the interaction between physical activity and age-related decline in tissues, these data could be used to correct data, or allow for future studies to compare if trunk neuromuscular activation patterns used by older adults differ for participant activity level.

Future Direction 2

EMG activation amplitudes are associated with force. While this relationship exists in both younger and older adults, age-related changes in the size and quality of trunk muscles are not uniform across trunk muscle groups [Anderson et al, 2012a; D'Antona et al, 2007; Ota et al, 2012]. A consequence of these changes is that properties such as joint stiffness and joint compression that are linked with trunk neuromuscular activation patterns in younger adults [Kavcic et al, 2004; Vera-Garcia et al, 2006] cannot be assessed in the current older adult participants. Future studies should determine if the age-related changes in trunk neuromuscular function are associated with increasing spinal stability or joint forces. To answer these questions one would have to either directly measuring joint stiffness, or attempt to calculate joint stiffness and joint compression in older adults in modeling studies.

Future Direction 3

A predominant theme in this thesis is that age-related changes in trunk neuromuscular function would compensate for: i) decreased passive stiffness while the spine is in neutral position, and/or ii) decreased joint position sense. To date it is unknown if older adults have decreased in vivo passive stiffness while the trunk is in a neutral position. This theory is built on in vitro data that older adults spines have more strain for a given stress [Iida et al, 2002a] and radiographic studies supporting age-related decreases in vertebral joint space [Hangai et al, 2008; Hicks et al, 2009]. However, it is also unknown whether joint space narrowing is associated with decreased passive stiffness. Secondly, healthy aging is associated with reduced trunk joint position sense [Goldberg et al, 2005] . However, no study has confirmed if joint position sense is associated with altered trunk neuromuscular activation patterns. Future studies should attempt to establish these relationships.

Future Direction 4

The final objective of this thesis was to establish whether age-related changes in neuromuscular function arise in the neural system, and if these changes serve as a compensatory mechanism. There is debate whether altered cortical recruitment in older adults represent a compensatory mechanism or are a function of reduced cortical inhibition [Bernard and Seidler, 2012; Seidler et al, 2010; Ward, 2006]. The findings from this thesis imply that older adults use more feed forward control of the trunk represented by more continuous trunk muscle activity. Using a dual-task paradigm to compete for cortical resources involved in feed forward activation of trunk muscles had no effect on trunk neuromuscular activation patterns in the participants of this study. In addition to feed forward recruitment, this thesis showed that older adults had altered recruitment of trunk muscle synergist and antagonist, a finding that suggests older adults have diffuse recruitment of the motor cortex. Since dual-task did not alter these recruitment patterns, it is possible that greater motor cortex diffusion is the result of an age-related change in reduced cortical inhibition. However, as electromyography measures muscle action potentials generated by both cortical and sub-cortical inputs, age-related differences in trunk neuromuscular activation patterns may be explained by changes to sub-cortical structures. Future studies, should attempt to determine what structures participate in age-related changes in trunk neuromuscular activation patterns.

Future Direction 5

All of these suggestions for future studies provide evidence that healthy aging is a complex process associated with changes to numerous systems. With an aging population it is increasingly important to determine to what extent age-related changes in the systems involved in maintaining joint control participate in common problems such as falls, and low back disorders. Both costly and are associated with decreased independence and function [Hicks et al, 2008; Rudy et al, 2007]. Despite work suggesting that age-related differences in trunk function (strength, motion, and position sense) participate in falls and low back injuries, no study has established if age-related changes in trunk neuromuscular activation patterns are associated with either problems. Thus, a long-term goal is to design a study that can assess this relationship, to be used as a metric to assess individual risk, clinical success, or treatment design for these age-related injuries.

6.4 CONCLUDING REMARKS

The purpose of this thesis was to determine if there were age-related differences in trunk neuromuscular activation patterns in order to complete a controlled dynamic transfer task. A secondary purpose was to determine if the neuromuscular activation patterns used by older adults were altered by an increased cognitive demand. Given that, the time varying recruitment of trunk muscles is essential to maintain trunk motion it is important to identify if there are age-relate changes in trunk neuromuscular activation patterns, as it may provide insight on why older adults have increased incidence of falls, and low back injuries.

In conclusion, this thesis provides a comprehensive analysis of age-related changes in trunk neuromuscular activation patterns while performing a functional transfer task. Given the methodological limitations of previous studies, the present work used a highly controlled experimental task to conclude that older adults have altered trunk neuromuscular activation patterns. Specifically, consistent with hypothesis one, it was found that older adults compared to younger adults had increased activation for agonist, increased antagonist co-activation, more continuous activation of trunk muscle sites despite changing external moments, and altered synergistic relationships between the numerous trunk muscle sites. Furthermore, consistent with the second hypothesis, it was shown that all age-related differences, except overall agonist activation, were not explained by increased relative physical task intensity.

The secondary purpose of this thesis was inconclusive. A single cognitive load in a dual-task experiment did not result in age-related differences in trunk neuromuscular activation patterns. The results of this study did reject the third hypothesis that a dual-task condition would modify the trunk neuromuscular patterns used by the active older adults that participated in this study. However, the results of this study do suggest numerous ways to adapt future studies to examine the effects of cognitive demand on physical task performance.

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APPENDIX A AGE-RELATED CHANGES IN THE NEUROMUSCULAR RESPONSE OF TRUNK MUSCLES DURING A DYNAMIC LEG LOADING PERTURBATION TASK

This appendix contains a modified version of the abstract published for the Canadian Society for Biomechanics, Burnaby, British Columbia, Canada. June 6-9, 2012.

INTRODUCTION

Healthy aging is associated with greater risk of falling or developing musculoskeletal injuries. It is speculated that this increased injury rate is due to altered neuromuscular patterns during dynamic loading. The purpose of this study was to determine if there are age-related differences in trunk musle response to a pelvis stabilization exercise using principal component analysis (PCA).

METHODS

Twenty-eight healthy adults volunteered for this study: 16 younger (20-45) adults were matched with 12 older (65+) adults based on mass, height, and sex. All participants signed an informed consent form approved by Dalhousie University Health Sciences Research Ethics Board.

Following a series of maximum isometric voluntary contractions (MVCs), participants performed three trials of a dynamic stabilization exercise involving a series of leg lifts from a supine crook lying position. To a 4-second count, they raised the right then left thigh to 90º, followed by left then right lowering to the starting position (Figure A.1). Angular displacement of the pelvis was captured using a Flock of Birds™ motion tracker system (Ascension Technologies Corp.) sampled at 50 Hz. Electromyographic (EMG) data collected from 24 trunk muscle sites (abdominal and back extensors) at 1000Hz were full wave rectified and low pass filtered (6 Hz), time normalized to 100% (entire exercise) and amplitude normalized to the appropriate MVC.

EMG ensemble-average waveforms for three trials for each muscle site and subject were entered into a principal component (PC) analysis model, with abdominal and back sites entered separately [Hubley-Kozey et al, 2009a]. Mixed model ANOVA (group, muscle) were conducted for each PC score ($\alpha=0.05$).

RESULTS

Peak pelvis angular displacement (5.5±2.7°) and time to complete movement (3.6±0.3s), were not different between the two groups. Five PCs explained 92% of the variance of abdominal sites (Figure A.2) with significant group and/or group* muscle interactions in PCs 4-5 (Figure A.3de). For the back sites, two PCs explained 91.6% of the variance (Figure A.4). With older adults having a significant group*muscle interaction for PC 1 (Figure A.5a). All PCs demonstrated significant muscle effects ($p < 0.001$).

DISCUSSION & CONCLUSIONS

For the abdominal muscle sites PC4 captured group difference where older adults compared to younger adults had continuous activation (higher scores) from 30-55% corresponding to no leg support (Figure A.2hi & Figure A.3d). PC5 accounted for a phase shift where older adult activity occurred earlier (higher scores) in ipsilateral oblique sites prior to left leg lift (Figure A.2jk & Figure A.3e). For the back muscle sites group differences were found in PC1, which captured the overall higher amplitude response of back muscles sites (Figure A.3b) in older adults (Figure A.5a).

Consistent with previous work [Hubley-Kozey et al, 2009a], this study found that older adults had an altered neuromuscular response for abdominal sites even for this low-demand task. Since the amplitude alterations were not systematic throughout the exercise, but were phase specific, they are not likely indicative of age-related strength deficits (Figure A.6bcd). In particular, it was found that older adults had more continuous activation of abdominals through the no leg support phase of the dynamic leg lifting task despite a reduced external moment during this period suggesting more continuous activation (Figure A.6abc). This continuous activation response along with the temporal phase shift (Figure A.6a) may represent an age-related change in amplitude response before leg loading perturbation reflecting increased anticipatory control to stabilize the lumbar spine. A novel finding of this study is the increased back muscle activation for specific back sites (Figure A.6f) in older adults suggesting an age-related antagonist co-activation strategy to stiffen the spine from perturbations. Together these changes illustrate that older adults have altered neuromuscular response that are not consistent between abdominals and back extensors during this stabilizing exercise.

Table A.1: Significant main effects and interactions (p-values) for PC scores for abdominal and back muscle sites

Comparison	PC1 abs	PC2 abs	PC3 abs	PC4 abs	PC5 abs	PC1 back	PC2 back
Group	0.252	0.963	0.963	< **0.001**	**0.120**	**0.007**	0.355
Group*Msl	0.377	0.646	0.599	0.286	**0.021**	**0.036**	0.131

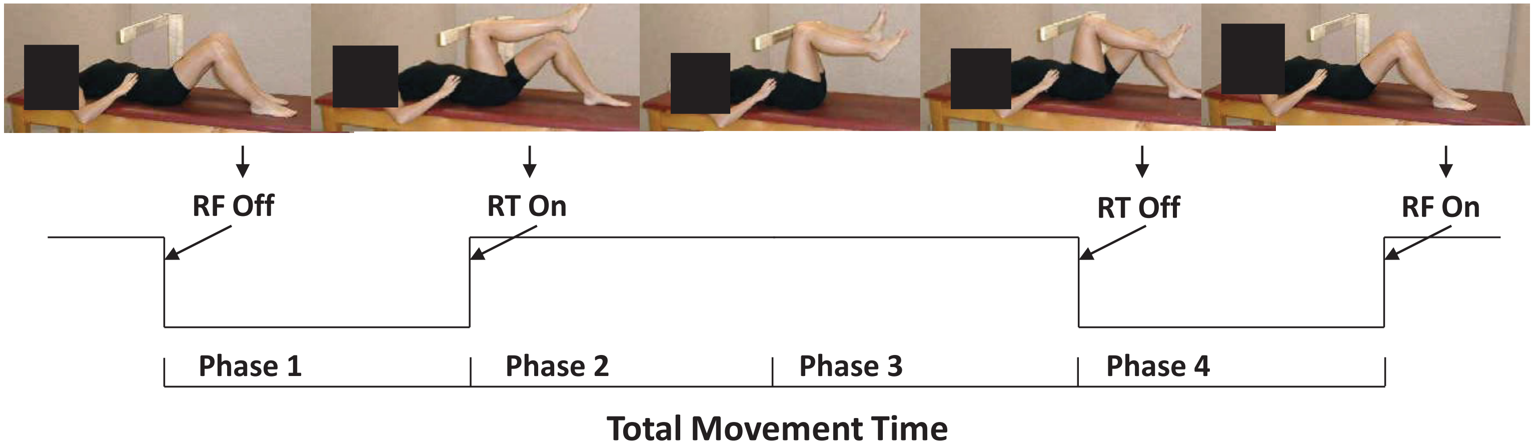
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Figure A.1: Exercise protocol utilize in experiment. From a supine position with the knees to remain bent at 90° participants were instructed to abdominally hollow "pull their navel up and towards their chest as if you were to tuck your stomach under your rib cage". Participants were then instructed to perform a continuous leg lifting exercise task timed to an external count with each second corresponding with a particular phase of the task. To ensure proper timing a metallic circuit was completed using a right foot (RF) or thigh (RT). Phase 1 right leg lift (right foot (RF) off to right thigh (RT) on), Phase 2 left leg lift (starting on RT on), Phase 3 left leg lower (ending on RT off), and Phase 4 right leg lower (RT off to RF on).

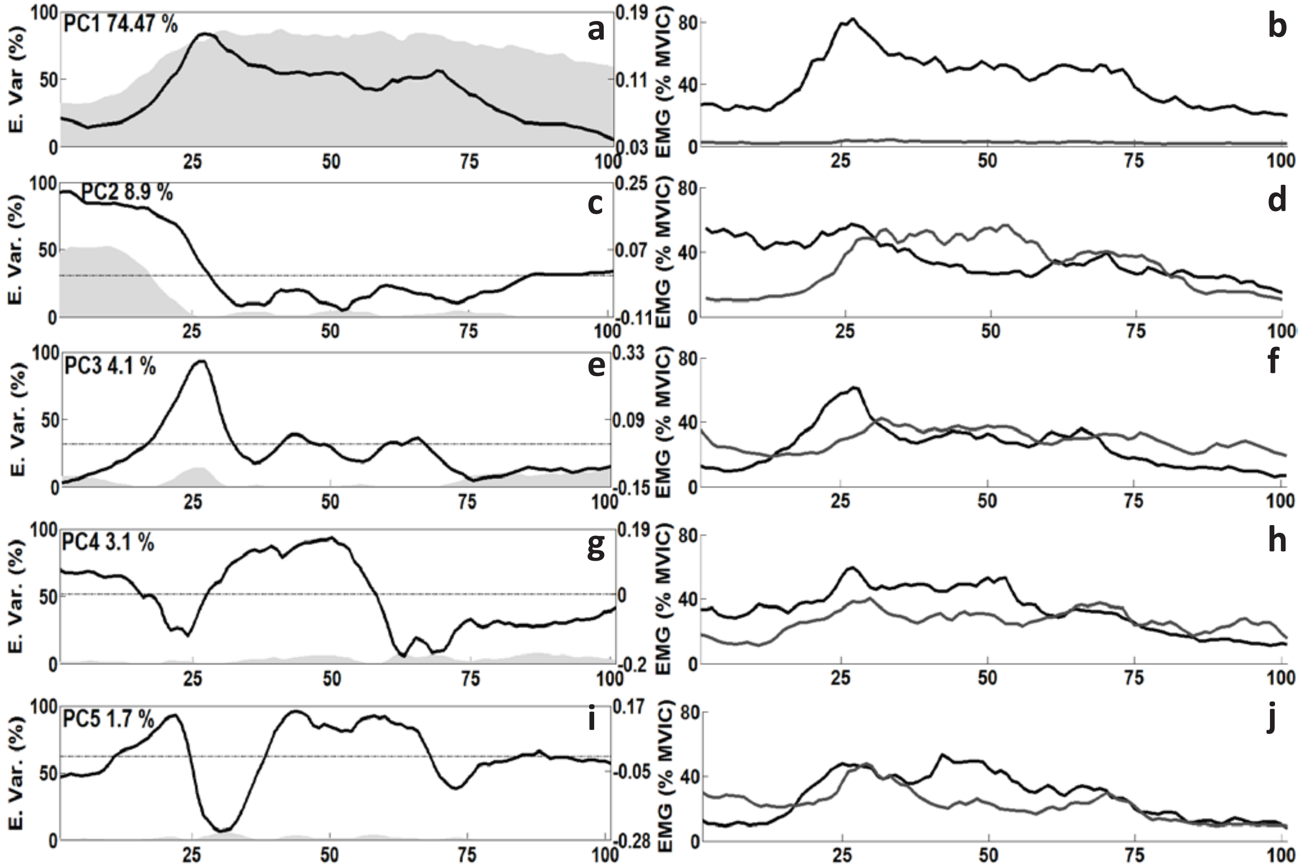

Figure A.2: For principal components (1-5) for the abdominal sites plots a, c, e, g, and i show the principal component (black line), and explained variance relative to total movement time depicted by grey shading. Total explained variance for each principal component is shown in the top left corner of these plots. To assist with interpretation, for each principal component plots b, d, f, h, and j show ensemble average waveforms of abdominal site EMG activation patterns representing the 5 highest (black line) and 5 lowest (grey line) PC scores.

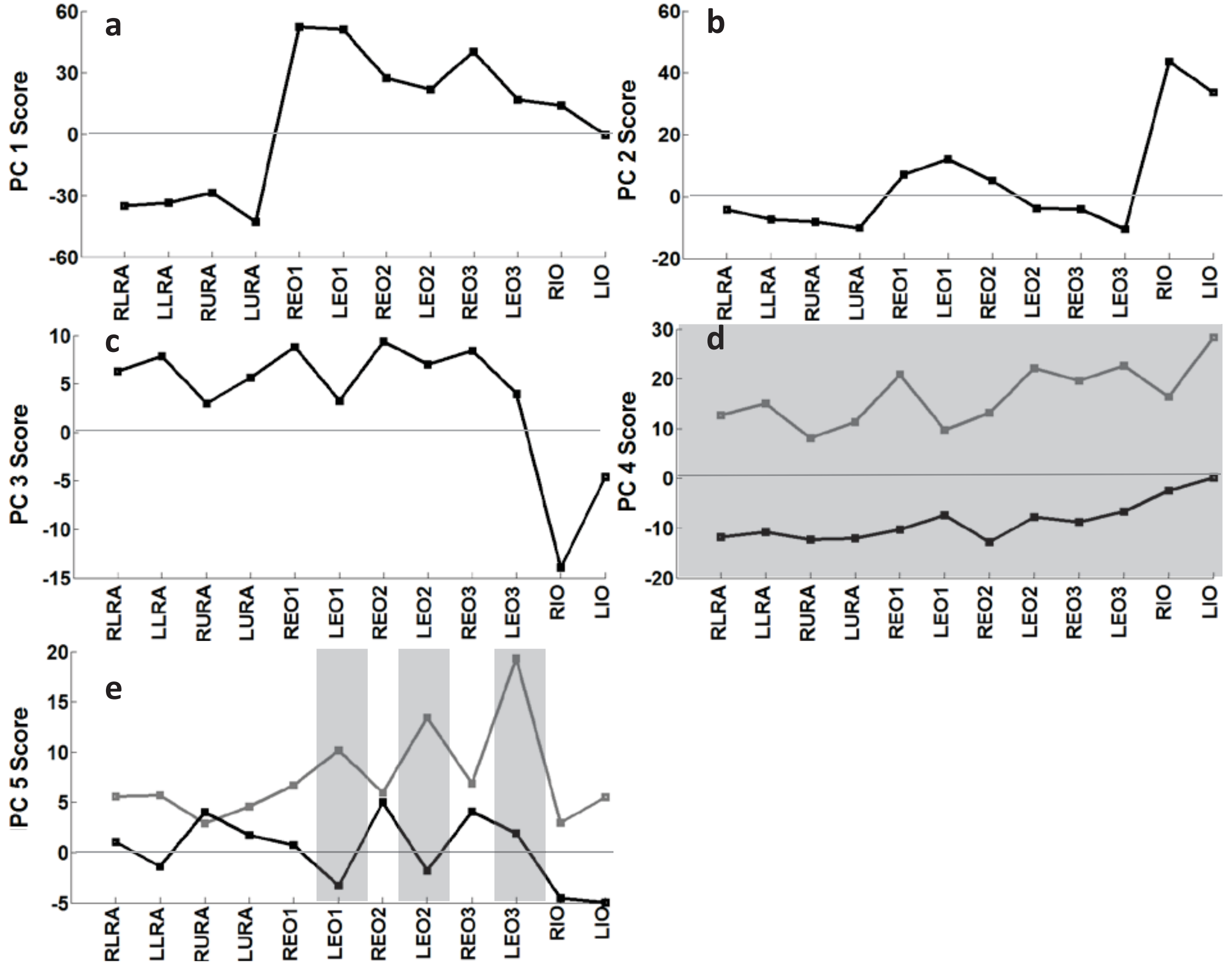

Figure A.3: Interaction plots for principal component (PC1-5 as indicated by the y-axis label) scores for abdominal muscle sites. For group*muscle interactions (d & e), grey lines = older adults, and black lines= younger adults. Grey shading indicates group differences.

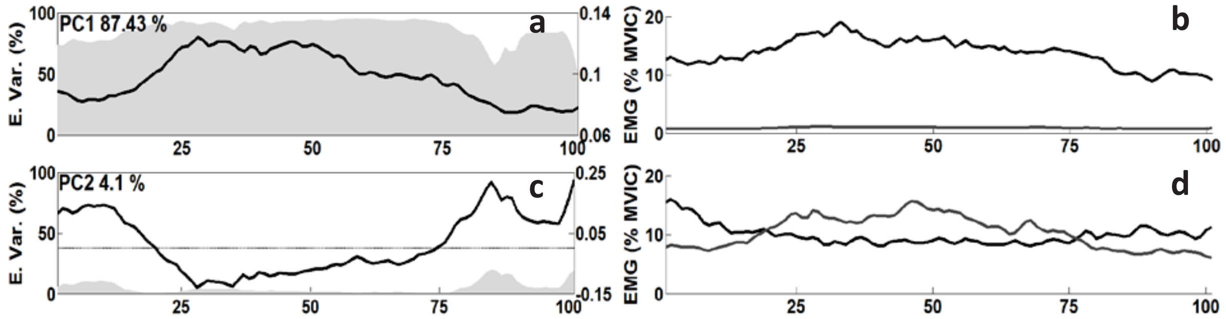

Figure A.4: For principal components (1-2) for the back sites plots a and c show the principal component (black line), and explained variance relative to total movement time depicted by grey shading. Total explained variance for each principal component is shown in the top left corner of these plots. To assist with interpretation, for each principal component plots b and d show ensemble average waveforms of back site EMG activation patterns representing the 5 highest (black line) and 5 lowest (grey line) PC scores.

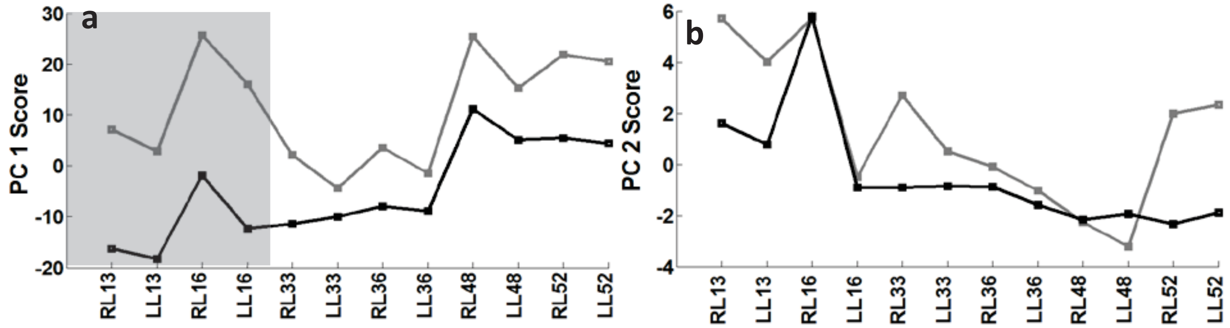

Figure A.5: Interaction plots for principal component (PC1-2 as indicated by the y-axis label) scores for back muscle sites. Line colour corresponds to older (grey line), and younger (black line) adults. Grey shading indicates significant group differences.

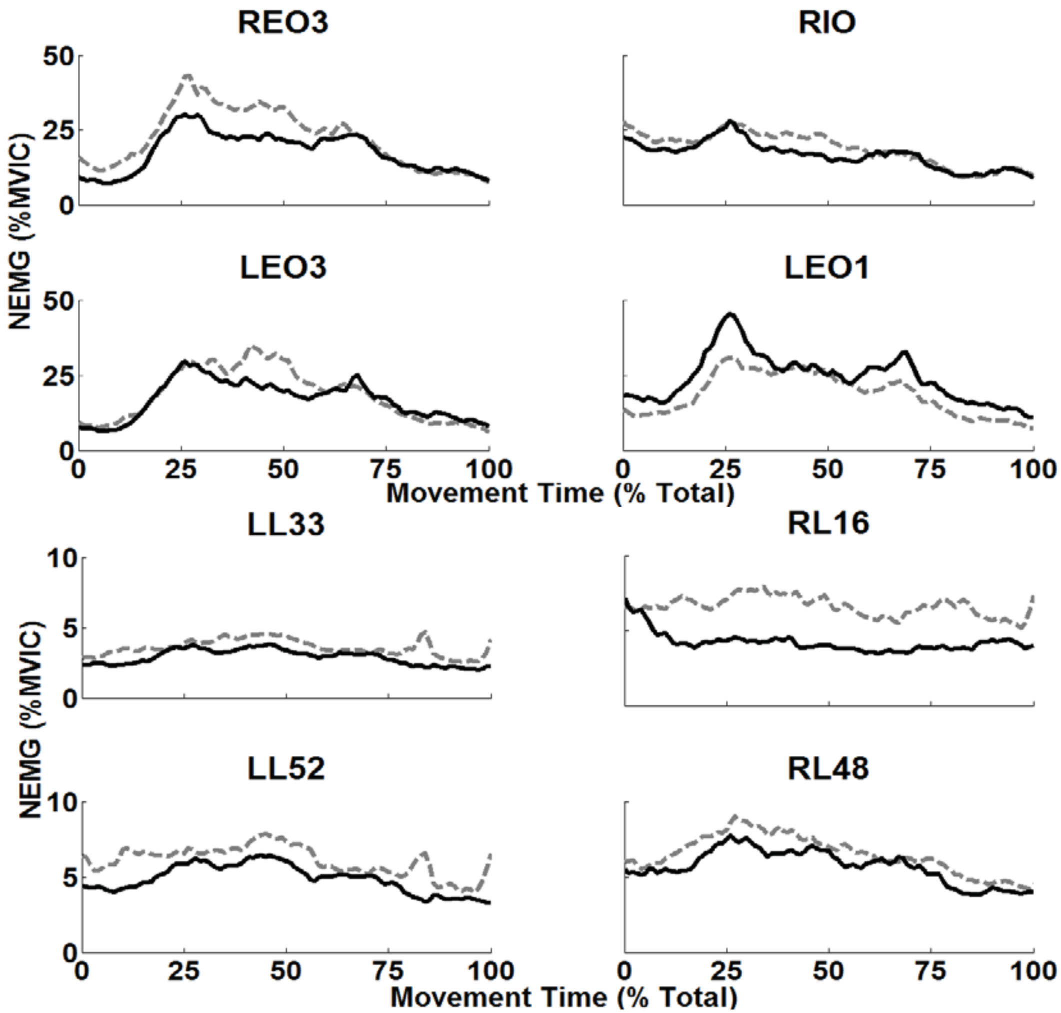

Figure A.6: Representative ensemble average EMG waveforms (abdominals and back sites) for older adults (grey dashed line) and younger adults (solid black line) while performing the dynamic leg loading perturbation task. Respective muscle site is denoted in the title of the subplot.