**THE EFFECT OF SEX AND MENSTRUAL CYCLE PHASE ON NEUROMUSCULAR
CONTROL OF TRUNK MUSCULATURE**

by

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Submitted in partial fulfillment of the requirements

for the degree of Master of Applied Science

at

Dalhousie University

Halifax, Nova Scotia

August 2012

DALHOUSIE UNIVERSITY

SCHOOL OF BIOMEDICAL ENGINEERING

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DALHOUSIE UNIVERSITY

DATE: August 17, 2012

AUTHOR: Tara Dahn

TITLE: The Effect of Sex and Menstrual Cycle Phase on Neuromuscular Control of Trunk Musculature

DEPARTMENT OR SCHOOL: School of Biomedical Engineering

DEGREE: MASc   CONVOCATION: May   YEAR: 2013

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This thesis is dedicated to my family and friends,

for their encouragement and love

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Abstract

Women have higher rates of noncontact musculoskeletal injuries compared to men, as well as at certain times in their menstrual cycles compared to others. The purpose of this study was two-fold: i) to examine the neuromuscular activation patterns of trunk musculature between men and women and ii) within women at different times in their menstrual cycle, during the trunk stability test (TST). The TST is a dynamic lower limb exercise that challenged the trunk musculature to maintain lumbopelvic stability. Surface electromyograms for 24 muscle sites and three-dimensional pelvic motion data were collected during the TST for 18 male and 19 female subjects, as well as for nine female subjects at different times in their menstrual cycles. Through analysis of amplitude and temporal characteristics of the EMG waveforms it was determined that women respond to the TST task with a less coordinated response than men, mainly relying on more co-activation. It was further determined that women have differences in their neuromuscular control patterns during the TST at different points in their menstrual cycle.

List of Abbreviations Used

TST   Trunk stability test

EMG   Electromyography

ACL   Anterior cruciate ligament

RMS   Root mean square

N   Newton

SI   Sacroiliac

PCA   Principal component analysis

PC   Principal component

LH   Luteinizing hormone

URA   Upper rectus abdominus

LRA   Lower rectus abdominus

EO1   External oblique (anterior fibers)

EO2   External oblique (lateral fibers)

EO3   External oblique (posterior fibers)

IO   Internal oblique

MVIC   Maximum voluntary isometric contraction

FOB   Flock of birds

ECG   Electrocardiography

DSI   Daily stress index

ANOVA   Analysis of variance

BMI   Body mass index

PMS   Premenstrual syndrome

ICC   Intra-class correlation

Acknowledgements

Chapter 1: Introduction

Evidence suggests that women may be at a higher risk of a wide range of musculoskeletal injuries when compared to men. For example, it has been shown that women have a higher incidence of low back injuries (Feuerstein, Berkowitz, & Peck, 1997; Krause et al., 1997; Schneider, Schmitt, Zoller, & Schiltenwolf, 2005; Schneider, Randoll, & Buchner, 2006). Female athletes have also been shown to suffer noncontact anterior cruciate ligament (ACL) injuries at a rate 2-8 times higher than male athletes (Arendt & Dick, 1995; Arendt, Agel, & Dick, 1999; Harmon & Ireland, 2000; Hewett, Zazulak, & Myer, 2007; Wojtys, Huston, Lindenfeld, Hewett, & Greenfield, 1998). Many researchers have investigated a wide range of factors to explain these differences, but a clear explanation does not exist.

Hormone fluctuations in females have been identified as placing women at higher risks of injuries at certain times during the menstrual cycles. Women have been shown to be more likely to injure their ACL during the pre-ovulatory phase of the menstrual cycle (Hewett et al., 2007; Shultz et al., 2010). But, no work has been published examining if menstrual cycle phase has an effect on prevalence of back injuries in women. Due to the presence of estrogen receptors on passive musculoskeletal structures in the body (Liu et al., 1996), it was first hypothesized that fluctuating estrogen levels might have an effect on the mechanical properties of tendons and ligaments in the body. Long term exposure to heightened estrogen levels has been shown to significantly decrease the collagen density of passive tissue structures (Lee et al., 2004; Liu et al., 1996; Miller et al., 2007), decreasing the stiffness of the structures and decreasing the stability of the joint. However short term fluctuations, such as over the course of the menstrual cycle, have a less clear effect. Reports in the literature of the effects of the estrogen fluctuations over the menstrual cycle on passive joint restraints are equivocal, with nearly equal numbers of

studies finding no direct association (Arnold, Van Bell, Rogers, & Cooney, 2002; Beynnon et al., 2005; Bryant et al., 2008; Karageanes, Blackburn, & Vangelos, 2000; Lovering & Romani, 2005; Miller et al., 2007; Romani, Patrie, Curl, & Flaws, 2003; Van Lunen, Roberts, Branch, & Dowling, 2003; Warden, Saxon, Castillo, & Turner, 2006) as finding a direct association (Deie, Sakamaki, Sumen, Urabe, & Ikuta, 2002; Eiling, Bryant, Peterson, Murphy, & Hohmann, 2007; Heitz, Eisenman, Beck, & Walker, 1999; Romani, Curl, & Lovering, 2001; Shultz, Kirk, Johnson, Sander, & Perrin, 2004).

Other researchers have begun investigating the possibility that the fluctuating sex hormones over the course of the menstrual cycle may have more of an effect on active joint stabilizers (muscles) and/or neuromuscular control. Estrogen receptors have been identified on skeletal muscles (Lemoine et al., 2003; Wilk et al., 2005), providing a plausible mechanism for how sex hormones may affect their properties. Neuromuscular control differences, assessed through electromyography and kinematic changes, in lower limb musculature, have been identified between men and women (Ford, Myer, & Hewett, 2003; Hewett, Myer, & Ford, 2004; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2009) during athletic jumping and cutting manoeuvers similar to the manoeuvers which most commonly lead to non-contact ACL injuries (Noyes, Mooar, Matthews, & Butler, 1983). While a number of factors may be at work, they could be influenced by the disparity in sex hormone levels. Limited work has been done to examine whether acute changes in sex hormones, such as over the course of a menstrual cycle, can have altering effects on a women's neuromuscular control strategies. Friden and colleagues (Friden et al., 2003; Friden et al., 2005) looked at the changes in women's postural control over the different phases of the menstrual cycle and found that some women show higher levels of postural sway in the mid-luteal phase of their cycle. Furthermore, Dedrick et al., 2006, found

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muscle onset time changes over the course of the menstrual cycle. They hypothesized that these changes would affect joint muscle co-contraction altering joint stiffness, which in turn can affect the injury susceptibility of a joint.

While the majority of the work has focused on the lower limb, in particular the knee joint, the trunk is a portion of the body that requires intricate neuromuscular control and coordination of a large series of muscles (Brown, Vera-Garcia, & McGill, 2006; Cholewicki & McGill, 1996; Granata & Orishimo, 2001). The spine is made up of 24 separate articulating vertebrae all of which can move in 6 degrees of freedom relative to each other (three translational: anterior/posterior, medial/lateral and caudal/dorsal; and three rotational: flexion/extension, lateral flexion/extension and axial rotation) (Bergmark, 1989; McGill, 1999). Theoretical and modelling work show that even one muscle responding incorrectly could disrupt spinal stability and lead to low back injury (McGill, 1999). The effect of fluctuations in estrogen levels over the female menstrual cycle on neuromuscular control of the trunk has not been investigated.

Differences between men and women in how they control their trunk musculature during a variety of tasks has been investigated (Granata & Orishimo, 2001; Granata, Orishimo, & Sanford, 2001; Hubley-Kozey, Butler, & Kozey, 2011; Kellis, Arabatzi, & Papadopoulos, 2003; Marras, Davis, & Jorgensen, 2003). Higher levels of co-activation, defined as the simultaneous activity of various muscles acting around the same joint (Kellis et al., 2003), have been identified in women's trunk musculature compared to men's. Some studies found that women only showed higher antagonist co-activation in the trunk (Granata et al., 2001; Granata et al., 2001; Hubley-Kozey et al., 2011) while others found that women exhibited both higher agonist and antagonist co-activation compared to men (Anders, Brose, Hofmann, & Scholle, 2007; Marras et al., 2003). In all the studies listed the agonist muscles were the back extensors and the antagonist muscles

were the abdominals. This difference cannot be simply explained by a strength difference between the sexes, since differences still existed between the sexes even when loads were adjusted to each individual's strength (Da Silva, Lariviere, Arsenault, Nadeau, & Plamondon, 2009; Granata et al., 2001). Instead it is thought to represent a compensatory mechanism in women to make up for their reduced passive stiffness compared to men (Hsu, Fisk, Yamamoto, Debski, & Woo, 2006; Markolf, Graff-Radford, & Amstutz, 1978; McGill, Seguin, & Bennett, 1994; Shultz et al., 2007). Co-activation can provide active stiffness to the joint to improve its overall stability (Cholewicki, Panjabi, & Khachatryan, 1998), however it also increases the overall loading on the joint (Granata & Marras, 1995; Thelen, Schultz, & Ashton-Miller, 1995).

When performing dynamic tasks, temporal coordination of trunk musculature is considered important for maintaining stability (McGill, 1999). With the exception of the Hubley-Kozey 2011 paper, all of the papers mentioned that address sex differences in neuromuscular control of the trunk use only discrete measures, such as RMS (root mean squared) amplitudes, to perform their assessments. Using only a discrete measure of EMG waveforms to assess a dynamic task has limitations as no information can be gathered about temporal coordination of the musculature. Hubley-Kozey et al., 2011, employed pattern recognition techniques (explained in more detail in following chapters) to compare the temporal EMG responses of 24 trunk muscle sites between men and women during controlled lifting tasks (Hubley-Kozey et al., 2011). They demonstrated that women had more temporal asynchronies among ipsilateral back and abdominal muscle sites compared to men. Thus our understanding of temporal synchronies in trunk muscular activation patterns of men and women is limited. In addition, as mentioned above, none of the work on sex differences in trunk neuromuscular control has examined tasks where the abdominal muscles act as agonists. The abdominals have been shown to play a critical

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role in the development and treatment of low back pain (Bergmark, 1989; Ferreira, Ferreira, & Hodges, 2004; Gardner-Morse & Stokes, 1998; McGill, Grenier, Kavcic, & Cholewicki, 2003), and thus there is a need to examine how abdominal muscle control is effected by sex as well as by alterations over the course of the menstrual cycle.

The present study examined the amplitude and temporal characteristics of neuromuscular activation patterns of the trunk during a dynamic stability exercise that challenged the abdominals. The task examined was the trunk stability test (TST) which is a dynamic leg lifting exercise which focuses on maintaining the spine in a neutral position as well as producing coordinated coactivity among muscle sites to sustain lumbopelvic stability (Clarke-Davidson & Hubley-Kozey, 2005). Lumbopelvic stability involves minimizing pelvic and lumbar motion by engaging the trunk musculature in the proper sequence. Differences were found in muscle EMG activation amplitudes (Hubley-Kozey & Vezina, 2002; Vezina & Hubley-Kozey, 2000) and temporal EMG patterns (Hubley-Kozey & Vezina, 2002) during the TST between healthy individuals and those with low back pain. Furthermore temporal pattern differences were found with aging (Hubley-Kozey, Hanada, Gordon, Kozey, & McKeon, 2009) and for those defined as stable and unstable (Hubley-Kozey, Hatfield, & Davidson, 2010). However no work has been done assessing whether or not men and women have different amplitude or temporal EMG pattern during the TST or whether women at different points in their menstrual cycle have different responses to the task.

1.1 Purpose
The purpose of this study was to investigate whether amplitude and temporal characteristics of neuromuscular activation patterns of the trunk musculature were affected by sex and menstrual cycle phase during a dynamic stability task that challenged mainly the

abdominals to maintain lumbopelvic stability. Findings could help provide an explanation for the differences in injury rates seen between men and women as well as within women at different points in their menstrual cycles.

1.2 Study Objectives

1.2.1 Objective 1

The first study objective was to compare the amplitudes of EMG activity as well as the temporal EMG characteristics between men and women during a dynamic stability exercise, the TST.

1.2.2 Objective 2

The second objective was to examine the amplitudes of EMG activity as well as the temporal EMG characteristics during a dynamic stability exercise, the TST, performed at two different times in the menstrual cycle, once during the follicular phase, when estrogen levels are at their lowest point, and once during the ovulation phase, when estrogen levels are at their highest.

1.3 Hypothesis

1.3.1 Hypothesis 1

It was hypothesized that women would show higher overall EMG amplitudes and more agonist and antagonist co-activation than men during the TST. It was also hypothesized that women would perform the task with more temporal asynchronies between activation profiles of different muscles.

1.3.2 Hypothesis 2

  Due to fluctuating sex hormone levels it was hypothesized that women would experience changes in the neuromuscular activation patterns of their trunk musculature over the course of their menstrual cycle. It is hypothesized that on the test day where their estrogen levels are highest women will have increased agonist and antagonist co-activation and increased temporal asynchronies between activation patterns of different muscles during the TST consistent with patterns reported for a group that was unable to maintain lumbopelvic stability (Hubley-Kozey et al., 2010).

1.4 Assumptions

The assumptions of this study are:

- That minimal movement will occur between the electrodes and the skin during the EMG recordings.

- The same electrode placement will be used and hence the same motor units will be sampled on each occasion.

- That no significant learning effect will occur between trials as well as between test sessions (controlled using familiarization sessions and practice trials).

- That fatigue will not be present during the testing.

- That the subjects are able to exert their maximal voluntary isometric contraction during the normalization exercises.

Chapter 2: Literature Review

2.1 Musculoskeletal injuries and sex

Two joints with inherent instability are the spine and the knee as they rely on soft tissue for stability rather than boney articulations. Low back pain is one of the most prevalent health concerns affecting our society. Over a one month period 28.5% of the adult population will report low back pain (Macfarlane et al., 2011) and over a 3 year period that prevalence increases to 67% of the population reporting low back pain (Jarvik et al., 2005). Overall low back pain puts a huge demand on health services and leads to a significant amount of work absenteeism. Unfortunately the situation is getting worse. The prevalence of impairing chronic low back pain increased from 3.9% in 1992 to 10.2% in 2006 (Freburger et al., 2009). This rise will continue if more attention is not given to the growing problem of low back pain. As mentioned in the introduction, women have been shown to have a higher incidence of low back disorders than men (Feuerstein et al., 1997; Krause et al., 1997; Schneider et al., 2005; Schneider et al., 2006). Low back disorders are complex and can be a result of a wide range of pathologies ranging from disk problems such as degenerated discs, stress fractures in the vertebrae, and strains or sprains of the ligamentous and muscular tissue surrounding the spine. However, 80-90% of low back disorders are said to originate as a result of minor mechanical alterations instead of traumatic injuries (Cohen, Argoff, & Carragee, 2008). According to McGill, low back injuries are usually the result of long term loading, which gradually, but progressively, reduces the tolerance of the tissues to withstand perturbations (McGill, 1998). The exact cause of most back pain remains unproven and the difference in incidence rates between men and women has not been well examined or explained.

A large portion of the work in sex specific injuries has been directed at attempting to explain the difference in incidence in ACL injury. Although the sports in which ACL injuries

occur most often (soccer, basketball, lacrosse, field hockey) do involve some level of player-to-player contact, the ACL injuries themselves are often noncontact injuries that result from landing from a jump, decelerating quickly and/or rapidly changing direction (Arendt et al., 1999; Olsen, Myklebust, Engebretsen, & Bahr, 2004). These movements put higher loads on the joints than those experienced during straight running. A non-contact mechanism is the most common mechanism leading to ACL injury (Renstrom et al., 2008) and the conditions that lead to non-contact ACL injuries are experienced by both men and women. The fact that women have a 2-8 times higher incidence of ACL injuries (Arendt & Dick, 1995; Arendt et al., 1999; Harmon & Ireland, 2000; Hewett et al., 2007; Wojtys et al., 1998) has yet to be fully explained.

2.2 Defining instability – Panjabi's model

A large culprit in the injury process in both the back and the knee is instability. In the trunk, spinal instability is said to be controlled by three main subsystems which work together to allow spinal motion, carry loads and protect the spinal cord and nerves (Panjabi, 1992). Before addressing these three subsystems it is important to first define spinal instability. At present the most commonly used definition of spinal instability is: "a significant decrease in the capacity of the stabilizing system of the spine to maintain the intervertebral neutral zones within the physiological limits so that there is no neurological dysfunction, no major deformity, and no incapacitating pain" (Panjabi, 1992). The three main subsystems which make up the stabilizing system of the spine are: (1) the "passive" subsystem, which includes the vertebrae, facet articulations, intervertebral discs, spinal ligaments, and joint capsules as well as the passive mechanical properties of the muscles, (2) the active musculoskeletal subsystem, which consists of the muscles and tendons and (3) the neural and feedback system, which consists of the various force and motion transducers in the ligaments, tendons and muscles, and the neural control

centers (Panjabi, 1992). Instability in the spine could result from a deficit in any one or any combination of the three subsystems. The passive subsystem develops passive resistance at the end of movement, however in the lumbar spine its load-carrying capacity (90 N) (Crisco & Panjabi, 1991) is insignificant relative to the capacity of the active subsystem, which can provide mechanical stability for loads exceeding 1500 N (Nachemson & Morris, 1964).

The active subsystem is made up of the muscles and tendons that contribute to spinal stability. These muscles include the rectus abdominus, the transverse abdominus, the internal obliques, the external obliques, the erector spinae, the quadratus lumborum and the multifidus. Each of these muscles contributes in its own way to spinal stability. The rectus abdominus and erector spinae muscles are the main agonists controlling flexion and extension of the trunk respectively. The transverse abdominus plays an essential role in spinal stability by acting as a muscular corset, increasing tension in the thoracolumbar fascia as well as increasing intra-abdominal pressure (Hodges, 1999). The transverse abdominus muscle activity is difficult to measure with anything other than indwelling fine-wire EMG electrodes. The internal oblique, especially the inferior fibers, seem to have a synergistic function to the transverse abdominus muscle (Hodges & Richardson, 1999; Urquhart, Hodges, Allen, & Story, 2005) and therefore can be used as a surrogate measure, which can be successfully recorded using surface EMG. The external obliques stabilize the spine by controlling twisting and rotations of the trunk (McGill, 1991; Mori, 2004). The quadratus lumborum is an excellent stabilizer of the lumbar vertebrae since it attaches each transverse process to the more rigid structures of the pelvis and rib cage, creating a bilateral support for each vertebrae (Liebenson, 2000). The multifidus is the largest and most medial of the lumbar paraspinal muscles and contributes two thirds of the active stiffness at the L4-L5 junction (Wilke, Wolf, Claes, Arand, & Wiesend, 1995). Theoretical and

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modelling work show that corruption in the behavior of any of these muscles can be enough to disrupt spinal stability and lead to injury (Cholewicki & VanVliet, 2002; Kavcic, Grenier, & McGill, 2004; McGill et al., 2003).

The neuromuscular subsystem is probably the most complex of the three subsystems. It perceives the proprioceptive signals from peripheral receptors including mechanoreceptors and pain receptors found in passive and active tissues, calculates the needs and, finally, produces and coordinates activation of the stabilizer muscles. Impairment in the neuromuscular control system can begin with mechanoreceptors. These can be disrupted or injured which causes them to produce corrupted transducer signals describing position, motion and loads. The neuromuscular control units, which respond to the transducer signals from mechanoreceptors, will be affected and will respond with corrupted muscle response patterns (Panjabi, 2006). Corrupted muscle response patterns can change which muscles are activated as well as individual muscle activation factors such as force onset, intensity and shut-off (Panjabi, 2006). Disrupted muscle control patterns can produce high stresses and strains on joint tissue and lead to pain and musculoskeletal injury (Panjabi, 2006).

This concept of stability-providing subsystems can easily be applied to joints other than the spine. For example, at the knee the ACL provides 86% of the static resistance to pure anterior tibial translation, representing a contribution to knee stability from the passive musculoskeletal subsystem (Butler, Noyes, & Grood, 1980). Research has also shown the importance of muscular activation around the knee, particularly from the hamstrings, in improving knee stability, which represents a contribution from the active musculoskeletal and neural subsystems (Goldfuss, Morehouse, & LeVeau, 1973; Hagood, Solomonow, Baratta, Zhou, & D'Ambrosia, 1990; Hirokawa, Solomonow, Luo, Lu, & D'Ambrosia, 1991; Markolf, Mensch, & Amstutz, 1976;

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Markolf et al., 1978). Early work by Goldfuss et al., 1973, reported a 48% increase in the stiffness of the medial side of the knee with contraction of the quadriceps and hamstring muscles and Markolf et al., 1976, used athletes to demonstrate a 10 fold increase in knee joint stiffness with muscle contraction. In the trunk the role of the musculature in stiffening the spine is especially critical as the passive structures of the spine provide even less stability than in the knee. As mentioned above, the osteoligamentous spine (spine with all non-passive tissues removed) can only withstand loads up to 90N (about 20 lbs), which means that without contribution from the surrounding musculature the human spine could not even support the weight of an individual's upper body.

2.3 Electromyography and the action potential

Changes in the active and neuromuscular subsystems can be monitored using electromyography (EMG). Electromyography monitors myoelectric signals produced in muscles with electrodes placed either over or inside a muscle. As a muscle is activated it generates action potentials which flow through the muscular tissues producing voltage gradients which can be recorded as the myoelectric signal. For researchers this signal is a useful method by which the mechanical contribution of muscles to gross anatomical movements can be estimated. The relationship between EMG and muscle force has been extensively studied during both isometric and dynamic contractions. During isometric contractions the EMG-muscle force relationship has been reported to be linear (Bouisset & Goubel, 1971; Moritani & deVries, 1978) or slightly non-linear (Heckathorne & Childress, 1981; Woods & Bigland-Ritchie, 1983). During dynamic contractions the relationship is more complicated as change in muscle force has been shown to be dependent on change in muscle fiber length (Gordon, Huxley, & Julian, 1966) and fiber velocity (Edman, 1978; Edman, 1979). Different models have been developed to estimate muscle

force based on EMG data (Buchanan, Lloyd, Manal, & Besier, 2004; Buchanan, Lloyd, Manal, & Besier, 2005; McGill, 1991).

Though the signals collected using EMG are not directly proportional to the tension and displacement generated by individual muscles, EMG signals provide valuable knowledge about strategies used by the central nervous system to control the musculature. EMG can be recorded with minimal interference to the movement in progress and from a large number of muscles simultaneously without significantly impacting the behavior of the system (Loeb & Gans, 1986). The process of muscular contraction and eventual movement begins in the frontal motor cortex of the brain where impulses from large neural cells, called motorneurons, travel downwards through the corticospinal tract in the spinal cord and out towards peripheral muscles. The impulses generated by the neural cells are called action potentials and are a result of a voltage depolarization-repolarization phenomenon that occurs across the neural cell membrane. The membranes' resting ionic potential is disrupted by a surrounding stimulus which allows sodium ions to briefly rush into the cell causing the membrane potential to spike and become positive. An active transport mechanism, called a sodium-potassium pump, quickly begins to pump sodium ions out of the cell, returning the membrane potential back to resting. This ion transfer sequence which makes up the action potential lasts about 1ms and stimulates a succession of similar events in a wave that eventually reaches the muscle tissue at the end of the nerve cell. When the impulse reaches the muscle it continues along the muscle tissue as a motor unit action potential (Kutz, 2009). A single fiber does not act alone; instead one motor neuron can have branches to several different muscle fibers which act together. All together the neuron and the fibers that it connects to are referred to as a motor unit.

The motor unit action potential generated in the muscle fiber causes the release of calcium ions within the fiber. The presence of calcium fibers allows the formation of cross bridges between actin and myosin filaments which make up the muscle fibers. The actin filaments slide inwards along the myosin filaments, causing the muscle fiber to contract (Kutz, 2009). The electromyogram measures the summation of all motor unit action potentials in a given location. Given the link between number of motor units firing and number of muscle fibers contracting, the electromyogram represents the overall strength and timing of a muscular contraction.

2.4 Sex differences in the stabilizing subsystems

2.4.1 Sex differences in the passive stabilizing subsystem
Since the three stabilizing subsystems play such an important role in the injury process, it is likely that the explanation for the disparity seen between male and female injury incidence can be found in differences between the sexes in one or a combination of these subsystems. Clear sex differences have been demonstrated in the passive subsystem. Women exhibit decreased tendon and ligament stiffness compared to men (Hsu et al., 2006; Markolf et al., 1978; Shultz et al., 2007). It has been suggested that decreased tendon stiffness will decrease the overall stability of a joint, therefore increasing the risk of musculoskeletal injury at that joint (Granata, Padua, & Wilson, 2002; Granata, Wilson, & Padua, 2002; Markolf et al., 1976). This results in women having less stable joints and higher risks of musculoskeletal injuries compared to men (Arendt & Dick, 1995; Arendt et al., 1999; Feuerstein et al., 1997; Harmon & Ireland, 2000; Hewett et al., 2007; Krause et al., 1997; Schneider et al., 2005; Schneider et al., 2006; Wojtys et al., 1998). It is suggested that this may occur due to the effects sex hormones have on the passive joint restraints (tendons and ligaments), since men and women physiologically have very different

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concentrations of the main sex hormones: estrogen, progesterone and testosterone (Tortora & Anagnostakos, 1987).

Estrogen and progesterone receptors have been identified on the fibroblasts of the human ACL (Liu et al., 1996) which leads to the belief that these hormones may have a direct influence on the ligamentous tissue. Progesterone has been observed to have no effect on soft tissue properties (Bell et al., 2011). Early studies using rats demonstrated that the administration of estrogen resulted in a decrease in both total collagen and collagen synthesis rates in rat tendons and fascia (Dyer, Sodek, & Heersche, 1980; Fischer, 1973). Estrogen has been shown to attenuate fibroblast activity, decreasing tendon collagen density (Lee et al., 2004) and synthesis rate in humans (Liu et al., 1996; Miller et al., 2007). Collagen is produced by fibroblasts and is responsible for the major load-bearing of the ACL (Smith, Livesay, & Woo, 1993), therefore a reduction in its density or synthesis rate would seriously impact the stability of the knee joint by reducing the mechanical strength of the tissue. Lee et al., 2004, demonstrated that in an estrogen-free environment mechanical loading of ACL fibroblasts resulted in an increased production of type 1 collagen mRNA; whereas, when estrogen was added to the environment the increase of type 1 collagen in response to loading was lessened. In the original study in 1997 by Liu et al. it was demonstrated that in fibroblasts treated for 2 weeks with physiological estradiol, the most common estrogen, concentrations of 0.025ng/ml showed collagen synthesis rates which were only 60% that of control fibroblasts in an estradiol-free environment. This would indicate that higher estrogen levels cause a decrease in collagen concentrations and an overall decrease in tendon/ligament stiffness. This explains why women, with their increased levels of estrogen, exhibit decreased tendon and ligament stiffness when compared to men (Hsu et al., 2006; Markolf et al., 1978; Shultz et al., 2007).   In the trunk specifically, McGill, Segiun and Bennett,

1994, demonstrated that women have a reduced level of passive stiffness in their lumbar torsos in multiple directions of movement compared to men (McGill et al., 1994). Although differences in passive tissue properties between men and women offer a partial explanation for the discrepancy seen in injury rates between the sexes, they are likely not the sole cause of these differences, especially in the trunk since, as has been mentioned previously, the passive system has a very small load bearing capacity.

2.4.2 Sex differences in the active stabilizing subsystem

Recent work has reported the presences of alpha and beta estrogen receptors in skeletal muscles (Lemoine et al., 2003; Wilk et al., 2005), providing a plausible tissue-based explanation for how sex differences in estrogen levels can influence the active subsystem. Sex differences in the rate of muscle force production have been documented. In general women require more time than men to produce the same relative muscle force level (Bell & Jacobs, 1986; Hakkinen, 1991; Huston & Wojtys, 1996; Winter & Brookes, 1991). Winters et al., 1991, suggested that the lower rate of force development in women may be a result of structural differences in the elastic components of their muscles, which could be caused by the discrepancy in estrogen levels between the sexes. Sex-based studies have also demonstrated that women have smaller tendon cross-sectional areas (Pichler et al., 2008), different muscle fiber type composition (Thorstensson & Carlson, 1987) and therefore less muscle strength compared to men. Women have been shown to have about 60% of the absolute torque production capacity of men (Linde et al., 1997; Sale, MacDougall, Always & Sutton, 1987).

2.4.3 Sex differences in the neural and feedback stabilizing subsystem

Differences in how men and women control their musculature have also been identified. Hewett et al., 2004, evaluated neuromuscular control strategies during landing both before and

after puberty in both men and women. Up until puberty men and women have similar sex hormone levels. Alterations in hormone levels, especially estrogen levels, are one of the primary changes that occur over the course of puberty (Sizonenko, 1978). Preceding puberty, men and women showed no differences in knee kinematics, inferring no differences in neuromuscular control patterns (Hewett et al., 2004). After puberty women appeared to show a change in their neuromuscular control pattern during landing resulting in an increased knee valgus alignment that places the ACL at a greater risk of injury (Ford et al., 2003; Hewett et al., 2004; Hewett & Myer, 2011). Women also exhibited increased rectus femoris activity and altered kinematic profiles during the early stance phase of an unanticipated cutting maneuver compared to men (Landry et al., 2009). Huston et al., 1996, also demonstrated that women use different muscle recruitment strategies than men. Women relied on their gastrocnemius and quadriceps to resist anterior tibial translation whereas men relied more on their hamstrings (Huston & Wojtys, 1996). These studies indicate that there are differences in neuromuscular control between men and women and these differences could help explain the higher incidence of musculoskeletal injuries seen in women.

2.5 Sex differences in trunk neuromuscular control

Examining trunk muscle control is important because not only can it help explain causes of low back pain, but it can also help explain causes of other injuries such as falls or lower limb musculoskeletal injuries. The trunk is the largest segment in the body and the location of the trunk center of mass affects the moments and forces on the lower limbs as well as body stability as a whole. Variability in trunk kinematics is thought to be associated with the decrease in stability while moving and increase in falls seen with age (Grabiner et al., 2008; Hurt, Rosenblatt, Crenshaw, & Grabiner, 2010; Menz, Lord, & Fitzpatrick, 2003). Deficits in

neuromuscular control of the trunk can lead to uncontrolled lateral trunk motion that may also increase knee abduction motion and torque through mechanical (lateral GRF motion) and neuromuscular (increased hip adductor torque) mechanisms (Hewett et al., 2005; Hewett, Myer, & Ford, 2005). Either or both of these mechanisms may increase strain on the ACL and lead to injury.

Granata and Orishimo, 2001, reported differences between the sexes in co-activation of the abdominals during a static trunk extension effort (Granata et al., 2001; Granata et al., 2001) . Muscle co-activation is defined as the simultaneous activity of various muscles acting around the same joint (Kellis et al., 2003). In the trunk co-activation increases joint stability and protects against damaging movement following perturbations (Gardner-Morse & Stokes, 1998). Higher mean normalized EMG amplitude values in the rectus abdominus and external obliques in women than men indicated that women recruited 32% more abdominal co-activation than men (Granata & Orishimo, 2001). Granata, Orishimo and Sanford, 2001, added to this finding, reporting that women also demonstrated increased back extensor muscle co-activation following static fatiguing exertions. Marras et al., 2003, also reported a higher level of abdominal co-activation in women compared to men during a dynamic lifting task and Anders et al., 2007, reported a higher level of co-activation in women compared to men in all trunk musculature measured (both abdominals and back extensors) during a whole body tilt. Other researchers, however, have found that in some situations men and women do not differ in their levels of co-activation (Nelson-Wong & Callaghan, 2010). Nelson-Wong and Callaghan, 2010, evaluated co-activation patterns over two hours of standing and found no differences between men and women. Their findings likely differed from previous studies because the intensity of their task investigated (standing) was significantly lower than the other tasks evaluated (static and dynamic

fatiguing exertions). For the sex differences in co-activation to become evident it seems a certain minimum level of demand on the musculature needs to be met.

All four studies mentioned above further support that women have reduced passive stability compared to men. By Panjabi's model, when one of the three main stability-providing subsystems is disadvantaged the other subsystems must increase their contribution to maintain the overall stability of the joint (Panjabi, 1992). Therefore the consequence of reduced passive stability is a higher demand on the musculature to compensate and produce stability through co-activation (active stability). Co-activation of muscles around a joint will help increase that joint's active stiffness (Cholewicki, Juluru, Radebold, Panjabi, & McGill, 1999) which in turn will improve the joint stability (Cholewicki et al., 1998). It is also argued that the co-activation of trunk musculature may be needed to account for the muscular strength differences between sexes (Mannion et al., 1997; Marras, Jorgensen, Granata, & Wiand, 2001). However differences between sexes still existed in back extensor activation amplitudes when tasks are adjusted to the individual's strength (Da Silva et al., 2009; Granata et al., 2001). A reduced level of passive stiffness would explain the necessity of higher levels of co-activation in women as compared to men.

The improved stability in the trunk generated by co-activation of the surrounding musculature comes at a cost. Co-activation of trunk musculature increases lumbar spine loads (Granata & Marras, 1995; Thelen et al., 1995). Muscles pull on their respective insertions and origins to generate trunk moments and therefore muscles that cross, insert on or originate from the lumbar spine can load it in compression and shear. A greater number of active agonist and antagonist muscles, that is, increased co-activation, can generate greater total spinal loading. Antagonist co-activation has an increased effect on spinal loading as it not only directly increases

the loading from its own contractile force, but by producing a moment in the opposite direction of the desired movement, it creates a demand on the agonist muscles to increase their muscle contractile forces to offset the higher antagonist moment (Granata & Marras, 1995). It has been estimated that the level of co-activation measured in EMG studies increases the compressive load on the spine during static activity by 16-19% for extension efforts and by a greater amount in lateral bending efforts (Thelen et al., 1995). For dynamic movements the increase in compressive loads is estimated to be even higher. Granata and Marras, 1995, calculated that the compressive loads and shear forces acting on the spine during dynamic lifting exertions increased by 45% and 70% respectively due to co-activation. Therefore, although co-activation is able to provide added stability to the spine, it would be more ideal to have a spinal system that was naturally stable without needing co-activation and the increased spinal loads that accompany it. The higher levels of co-activation seen in the female spine may put women at greater risk of back injuries due to the increased compressive and shear forces experienced.

Co-activation is often measured by activation amplitudes (Anders et al., 2007), co-activation indices (Kellis et al., 2003) and onset and offset times (MacDonald, Moseley, & Hodges, 2010), all of which are discrete measures. It is important to acknowledge the limitations of using a discrete measure of muscle activity to describe the relationships between muscles responding to a dynamic task. In an attempt to better relate muscle activation synergies and co-activations to dynamic neuromuscular control, studies have applied correlations (Sirin & Patla, 1987), which allow for the temporal patterns to be compared between muscles. However, correlations between muscles can only be applied to two muscles at a time, limiting the ability to perform a reasonable data analysis on a large group of muscles. Recent studies using pattern recognition techniques (explained in more detail below, as well as in Appendix A) on the

temporal EMG waveforms, allow the examination of both amplitude and temporal characteristics of muscle activation for a wide group of muscles (Butler, Hubley-Kozey, & Kozey, 2010; Hubley-Kozey et al., 2010; Ivanenko, Poppele, & Lacquaniti, 2004; Lamoth, Meijer, Daffertshofer, Wuisman, & Beek, 2006). Hubley-Kozey et al., 2011, employed these pattern recognition techniques to compare the temporal EMG responses of 24 trunk muscle sites between men and women during controlled lifting tasks (Hubley-Kozey et al., 2011). Consistent with the previous work they found higher relative activation amplitudes or increased co-activation. However in addition to these amplitude differences they demonstrated that women had more temporal asynchronies among ipsilateral back and abdominal muscle sites compared to men. These asynchronies may arise from changes in the neuromuscular control system over the course of a women's menstrual cycle creating more variability in the female system. To our knowledge no one has examined temporal EMG patterns of the trunk over the course of the menstrual cycle in women to see if this is true. If so, it could help explain the higher prevalence of musculoskeletal injuries in women compared to men.

2.6 Effects of the menstrual cycle on injury risk

It has not only been suggested that the hormone differences between men and women lead to differences in injury rates but also that a woman's own fluctuations in hormone levels can result in her being more susceptible to injuries at one time in her cycle compared to another. A woman's hormone levels fluctuate significantly over the course of her monthly menstrual cycle. During the early follicular phase (day 1-6) serum estrogen and progesterone levels are low. Estrogen levels are then elevated during the late follicular phase (day 7-11) with an estrogen spike occurring at the time of ovulation (day 14). During the luteal phase (day 15-28)

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progesterone levels elevate and estrogen levels remain somewhat elevated and slowly return to

baseline (Wojtys et al., 1998) (See Figure 2.1).

The most recent ACL consensus statement concluded that the risk of suffering a

noncontact ACL injury is not equal across the menstrual cycle and is greatest during the pre-

ovulatory phases, which consists of menses through ovulation (Shultz et al., 2010). A 2007

review of the literature by Hewitt, Zazulak & Myer came to a similar conclusion, stating that

there is an increase in noncontact ACL injuries surrounding the pre-ovulatory phase of the

menstrual cycle (Hewett et al., 2007).There is some disagreement on this topic, with some

studies showing an increased risk of ACL injury at ovulation (Adachi, Nawata, Maeta, &

Kurozawa, 2008; Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002; Wojtys et al., 1998).
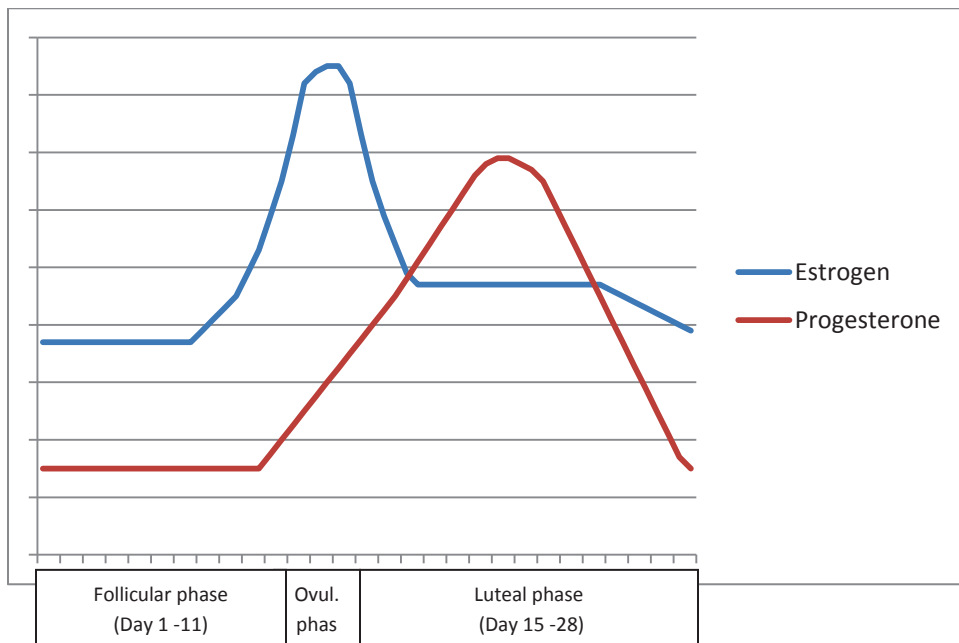
Figure 2.1: Estrogen and Progesterone levels over the menstrual cycle. The ovulation phase is abbreviated as Ovul. phase and lasts from day 12-14. (adapted from Dzugan & Scipione, 2006).

Some discrepancies between methods within the studies examining ACL risk over the

menstrual cycle may be partially responsible for the disagreement in results. Not all the studies

mentioned above excluded participants taking oral contraceptives. Oral contraceptives greatly alter the natural hormone fluctuations seen within the female body, significantly attenuating and leveling out estrogen levels (Bryant et al., 2008). Having subjects on oral contraceptives within a subject pool makes it difficult to draw conclusions about results. Also there is no consensus between these studies on menstrual cycle phase labels and lengths. Studies divide the cycle into anywhere from 2-4 phases, often of varied lengths and often using different names for the same time period between studies. Nevertheless, these studies do conclude that menstrual cycle may be an important factor in ACL injuries. No work has been done yet examining if menstrual cycle phase has an effect on prevalence of back injuries in women.

A potential mechanism leading to the varied injury rates across the menstrual cycle is the effect of fluctuating estrogen on mechanical properties of tendons and ligaments. Fluctuations in sex hormones, which are unique to the female body, are commonly included by researchers as a potential factor leading to an increased risk of soft tissue injuries. Since women have higher levels of estrogen at specific points in their menstrual cycles it is suggested that the stiffness of their ligaments, and thus their risk of sustaining musculoskeletal injuries, may change in accordance with their hormone levels.

Studies differ on whether acute changes in plasma estrogen levels over the course of the menstrual cycle can affect soft tissue mechanical behaviour. The majority of the work has focused on the ACL, although one study examined the Achilles tendon (Bryant et al., 2008). Several studies have identified acute estrogen-induced changes in soft tissue mechanical behavior (Deie et al., 2002; Eiling et al., 2007; Heitz et al., 1999; Romani et al., 2001; Shultz et al., 2004). Other studies have demonstrated no direct association between acute estrogen fluctuations and soft tissue mechanical behavior (Arnold et al., 2002; Beynnon et al., 2005;

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Bryant et al., 2008; Karageanes et al., 2000; Lovering & Romani, 2005; Miller et al., 2007; Romani et al., 2003; Van Lunen et al., 2003; Warden et al., 2006). A review of the literature by Zazulak et al., 2006, stated that 6 out of 9 prospective cohort studies did not shown any correlation between ACL laxity and menstrual cycle phase (Zazulak, Paterno, Myer, Romani, & Hewett, 2006).

`   Several of the studies that did find changes in soft tissue mechanical properties over the course of the menstrual cycle indicate that lower magnitudes of anterior knee laxity are found during the early follicular phase, when both estrogen and progesterone are at their lowest point, and higher magnitudes of laxity are found near ovulation and the early luteal phase, when estrogen levels are higher but progesterone levels have not yet risen (Deie et al., 2002; Park, Stefanyshyn, Loitz-Ramage, Hart, & Ronsky, 2009).

       Shultz et al., 2004, argue that estradiol, progesterone and testosterone and their interactions account for 63% of the changes in knee laxity across the cycle when a 3-4 day time delay was considered. But that there is a lot of variability in the length of time delay between changes in hormone levels and changes in knee laxity (Shultz et al., 2004). Shultz et al., claim that this variability makes it difficult to identify a single test day in each phase that will represent the same point in the cycle for all women and that this could explain why studies using a single test day per cycle phase often did not find cyclic changes in knee laxity across the menstrual cycle. Daily tracking of hormone levels and knee laxity measures were used in an attempt to get a better picture of individual variability (Shultz, Gansneder, Sander, Kirk, & Perrin, 2006). However, they were unable to replicate their previous results and were not able to explain with hormone changes why some women saw changes in knee laxity while others did not.

Shultz et al., 2011, attempted to complete the picture of knee ligament laxity variation by examining varus-valgus and internal-external rotational laxity in addition to the conventionally measured anterior knee laxity. They found that the cyclic variations in anterior knee laxity over two time points in the menstrual cycle were not accompanied by changes in varus-valgus or internal-external rotational laxity. They concluded that there may be ligament-specific responses to hormone changes over the menstrual cycle and that further work would be needed to investigate this.

Overall the reports of the effect of menstrual cycle phase on the mechanical properties of soft tissues such as the ACL are equivocal, with nearly equal numbers of reports supporting both sides. It seems that alterations to soft tissue mechanical properties alone do not explain the effect that the menstrual cycle seems to have on injury susceptibility.

2.7 Effect of the menstrual cycle on neuromuscular control

The reports of the effects of the menstrual cycle on passive joint restraints are vague and do not alone explain the differences in injury rate seen over the course of the menstrual cycle. Recent work has reported the presence of estrogen receptors alpha and beta in skeletal muscles (Lemoine et al., 2003; Wilk et al., 2005) providing another plausible tissue-based explanation for how estrogen fluctuations over the course of the menstrual cycle can influence the active subsystem. However, it may be that the menstrual cycle has a larger effect on the neuromuscular subsystem, as opposed to the mechanical properties of the muscular subsystem itself, since muscular fatigability and muscle strength in women do not seem to change over the course of the menstrual cycle (Janse de Jonge, Boot, Thom, Ruell, & Thompson, 2001; Lebrun, McKenzie, Prior, & Taunton, 1995).   Janse de Jonge et al., 2001, examined muscle function at three

different phases of the menstrual cycle with significantly different concentrations of female sex hormones. They found no significant change for any strength parameters including maximal isometric quadriceps strength, isokinetic knee flexor and extensor strength and hand grip strength. They also found that the quadriceps contractile properties and knee flexor and extensor fatigue were not affected by the menstrual cycle. Lebrun et al., 1995, also determined that the isokinetic strength of the quadriceps and hamstrings were not influenced by the menstrual cycle phase. Changes in the neuromuscular subsystem alone can still easily lead to injury. In fact, impaired coordination of trunk musculature has been shown to be associated with low back pain (Akuthota & Nadler, 2004; Hodges & Richardson, 1996; Hodges & Richardson, 1998), whereas improved strength of the trunk musculature does not seem to be associated with the prevention of low back pain (Nadler, Malanga, DePrince, Stitik, & Feinberg, 2000; Nadler et al., 2001; Nadler et al., 2002).

Limited work has examined whether acute changes over the menstrual cycle can have altering effects on a women's neuromuscular control strategies. Early studies found that fine motor activity and reaction times can fluctuate over the course of the menstrual cycle (Posthuma, Bass, Bull, & Nisker, 1987; Stocker, 1974). More recently, Friden and colleagues (Friden et al., 2003; Friden et al., 2005) have looked at the changes in women's postural control over the different phases of the menstrual cycle and found that some women show higher levels of postural sway in the mid-luteal phase of their cycle. Dedrick et al., 2006, found that women performing drop jumps had delayed onsets of their semitendinosis muscles during the luteal phase of the menstrual cycle as well as altered hip muscle onset timing differences compared to during the follicular phase. No kinematic differences in lower limb alignment accompanied these neuromuscular differences. Work by Chaudhari et al., 2007, also demonstrated no kinematic

differences during several demanding lower limb actions performed by women at different phases of their menstrual cycles. Dedrick et al., 2006, suggest that the muscle timing changes over the course of the menstrual cycle may affect joint co-contraction and result in altered joint stiffness and the ability to distribute force through bony and soft tissues. These changes would in turn affect the injury susceptibility of a joint.

It seems that changes in sex hormones can have an effect on how an individual's neuromuscular system reacts to different tasks. Large, long-term changes in sex hormones, such as the developmental changes that differentiate men from women, create clearly demonstrated alterations in neuromuscular control patterns (Ford et al., 2003; Hewett et al., 2004; Landry et al., 2009). Shorter term changes in sex hormones, such as over the course of the menstrual cycle, have less clear effects on neuromuscular control. Preliminary evidence suggests that there may be some base differences in neuromuscular control over the menstrual cycle (Dedrick et al., 2006; Friden et al., 2003; Friden et al., 2005). However, the lack of evidence of kinematic differences between tasks performed at different points in the menstrual cycle suggests that the differences may be subtle.

2.8 Effect of the female menstrual cycle on control of trunk musculature

Few studies have examined whether trunk muscle control changes over the course of the menstrual cycle. The work by Friden et al. in 2003 and 2005 is likely the closest to evaluating neuromuscular control of the trunk over the menstrual cycle. They examined how postural sway, which can be thought of as a partial outcome of trunk muscle control, changed over the menstrual cycle. As stated earlier, they found that women showed higher levels of postural sway in the mid-luteal phase of their cycle. There has been quite a lot of work examining the

difference in trunk muscle control between men and women as has been discussed in previous sections. Since hormone levels are one of the biggest physiological differences between men and women it can be argued that they may play a role in any differences observed between men and women in their trunk muscle control.

2.9 Experimental procedures

2.9.1 Experimental tasks to evaluate neuromuscular control of the trunk

Neuromuscular control of the trunk relies on the integration and coordination of a wide number of separate muscles. Studies of the trunk suggest that all these muscles must be coordinated with strict muscle activation patterns in order to maintain stability during motion while minimizing joint loads and risk of low back pain (Brown et al., 2006; Cholewicki & McGill, 1996; Granata & Orishimo, 2001). It has been suggested that even one muscle responding inappropriately could provide the necessary force to disrupt stability (McGill, 1999). Neuromuscular control of the trunk can be assessed by examining both amplitude and temporal characteristics of the electromyogram (EMG) recorded during the performance of a dynamic task, which can provide information about potential differences in the timing, recruitment order, and co-activation patterns of the trunk musculature.

Lumbopelvic stabilization exercises aim at reprogramming spine stabilizing muscles to improve their motor control skills, as well as their ability to compensate for any weakness in the passive stabilization structures. Dynamic stability exercises are widely used to build muscle strength and endurance as well as develop neuromuscular control strategies required to maintain dynamic trunk stability. The trunk stability test (TST) has been investigated in the literature

(Clarke-Davidson & Hubley-Kozey, 2005; Hubley-Kozey & Vezina, 2002; Hubley-Kozey et al., 2009; Hubley-Kozey et al., 2010; Vezina & Hubley-Kozey, 2000) because it is a dynamic stability exercise that focuses on maintaining the spine in a neutral position as well as producing coordinated co-activity among muscle sites to sustain lumbopelvic stability. The task involves leg-loading and is dynamically challenging to the abdominal musculature due to the altered loading on the lumbopelvic region throughout the exercise. The TST task is conducted in a supine position. As the legs are lifted they create an external hip extensor moment, which is countered by contraction of the hip flexors, mainly the rectus femoris and psoas. The psoas originates from lumbar vertebrae at levels L4 and L5 and therefore its contraction imposes forces on the spine that challenge lumbopelvic stability.

Lumbopelvic stability is defined as the ability to maintain the pelvis in a physiologically safe position relative to the lumbar spine. The link between the pelvis and lumbar spine is composed of the sacroiliac (SI) joint. The SI joint has to transfer large loads between the lower limb and torso and has a relatively flat joint surface that is favourable for the transfer of compressive forces and bending moments (Snijders, Vleeming, & Stoeckart, 1993). However the flat SI joint is vulnerable to shear forces (Snijders et al., 1993). The SI joint is somewhat protected from these shearing forces by anatomical features of the joint such as the wedge shape of the sacrum and rougher articular cartilage relative to other joints. A coarser texture of the cartilage and the presence of ridges and depressions enhances friction at the joint and consequently increase stability (Bowen & Cassidy, 1981; Pool-Goudzwaard, Vleeming, Stoeckart, Snijders, & Mens, 1998; Vleeming, Stoeckart, Volkers, & Snijders, 1990). Men have more prominent cartilage modifications (Vleeming, Volkers, Snijders, & Stoeckart, 1990), indicating that women may have less natural protection against shear forces acting on the SI

joint. Further resistance to shear forces on the SI joint is created by muscle forces that provide compression on the joint. Inadequate muscle strength or improper coordination between muscles can cause insufficient protection of the SI joint against shear forces which could lead to lumbopelvic instability, pain and low back disorders (Pool-Goudzwaard et al., 1998).

The TST task consists of 5 levels of varying difficulty. Level 3 of the task involves an unsupported single leg extension (see Figure 2.2) and is the only level with both increased demands on the abdominal musculature and increased asymmetric loading (Clarke-Davidson & Hubley-Kozey, 2005). Hubley-Kozey et al., 2010, identified neuromuscular patterns during the TST task that divided individuals into two groups: those who were able to perform the exercise without excessive lumbar and pelvic motion (stable group) and those who were unable (unstable group) (Hubley-Kozey et al., 2010). Both groups showed an overall mean activation pattern that was similar for all abdominal wall muscle sites, indicating an underlying synergy, with the rectus abdominus being slightly more responsive to leg loading perturbations. The stable group showed higher relative activation in their lower rectus abdominus sites, which represent the fibers that attach directly to the pelvis, than their upper rectus abdominus sites, whereas the unstable group showed no difference in activation between the two rectus abdominus sites (Hubley-Kozey et al., 2010). The stable group also exhibited higher relative activation in the lateral and posterior fibers of the external oblique compared to the anterior fibers. In contrast, the unstable group showed higher relative activation in the anterior and lateral fibers of the external oblique compared to the posterior fibers. The physiological role of the anterior fibers of the external oblique in maintaining lumbar-pelvic stability is unclear as they do not attach to the pelvis or the lumbar spine. This may have contributed to the inability of the unstable group to minimize lumbar-pelvic motion (Hubley-Kozey et al., 2010). Also subjects in the unstable group displayed an

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exaggerated burst of activity associated with lifting the second leg off the table during the task which has been previously reported in subjects with low back pain performing level 1 of the TST task (Hubley-Kozey & Vezina, 2002). Other researchers have identified neuromuscular control differences between subjects with 'fit' motor systems who are able to easily meet the stability demands of a task and subjects with 'less fit' motor systems. Subjects with 'less fit' motor systems demonstrate more variability in their ability to stabilize their spine, sometimes producing more co-activation of surrounding musculature to keep stability high, which consequently can add more compression to the spine (McGill, Sharratt, & Seguin, 1995).
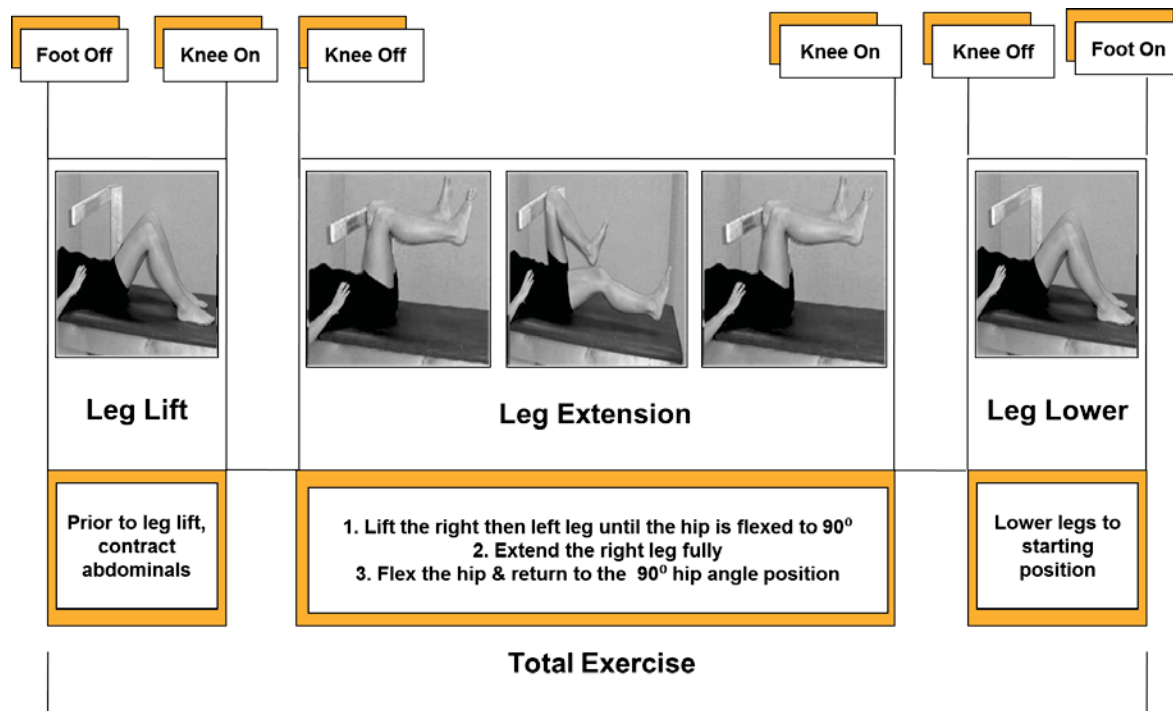
Figure 2.2: The TST began with the participant lying supine on the bed. Prior to beginning the task the participant performs an abdominal hollowing manoeuver. The test consisted of the participant lifting their right leg and then their left to 90º hip flexion (thighs met a crossbeam), extending their right leg out to full knee extension, gently tapping the right heel on the table, returning the right thigh to the crossbeam, and lowering the left then right foot to the bed. The movement was divided into three phases: leg lifting, leg extension and leg lowering, which were defined by event markers generated by pressure sensitive switches located on the right foot, the top of the right thigh and the crossbeam. The task was completed to an 8 second count (1 – right thigh to crossbeam, 2- left thigh to crossbeam, 3- right leg extending, 4 – right leg fully extended, right heel tapped on bed, 5- right leg flexing, 6- right thigh to crossbeam, 7- left foot on bed, 8- right foot on bed) (modified with permission from Hubley-Kozey et al., 2010).

2.9.2 Factors effecting EMG measurement

EMG crosstalk

One of the biggest drawbacks of using surface electromyography is the potential for surface electrodes to pick up myoelectric signals from muscles other than the muscle of interest. This signal-contaminating phenomenon is known as crosstalk (Winter, Fuglevand, & Archer, 1994). The presence of crosstalk can make the interpretation of the EMG signal more difficult.

How large an influence crosstalk has on measured signals is hard to quantify. One study measured surface EMG of several maximally stimulated feline muscles before and after the nerves running to specific muscles that could be contributing to crosstalk were cut (Solomonow et al., 1994). Following cutting the nerves they concluded that in the 'worst-case scenario' cross-talk was contributing to only 5% of the signal value. When this result is considered together with the fact that surface electrodes have relatively shallow pick-up depths (18mm) it is suggested that the effect of cross-talk may be negligible in large muscles (Fuglevand, Winter, Patla, & Stashuk, 1992; Solomonow et al., 1994). Some studies using other analysis techniques, such as cross-correlation techniques have concluded that 13-16% of an EMG signal can be due to contributions from neighboring muscles (De Luca & Merletti, 1988; Winter et al., 1994). However, upon investigation it seems that using cross-correlation techniques to assess cross-talk may be somewhat flawed as cross-correlations could misinterpret muscles acting synergistically as cross-talking onto each other due to their similar activation patterns. Additionally, since signals change as they pass through different volumes of tissue, cross-talk signals may have different components from the muscle of interest, which would reduce the reliability of using cross-correlations for evaluating cross-talk (Farina, Merletti, Indino, & Graven-Nielsen, 2004; Lowery, Stoykov, & Kuiken, 2003). Nonetheless it seems that a signal measured from one muscle cannot

be assumed to originate only from that muscle and none of the other nearby muscles. To account for this, methodological procedures can be adapted to reduce the influence of crosstalk in EMG measurements. For example, smaller electrodes with reduced spacing can be used to decrease the pick-up volume and the potential for cross-talk (Fuglevand et al., 1992).

Number of sites to measure

Another important concern when evaluating neuromuscular control is how many muscle sites to measure. The number needs to be high enough to capture the key neuromuscular control patterns, but still low enough to keep the volume of data manageable. Experimental and biomechanical modelling studies show that all trunk muscles have an important role (Cholewicki & VanVliet, 2002; Kavcic, Grenier, & McGill, 2004) and that even different segments of the same muscle can respond differently to a perturbation (Jonsson, 1970; Mirka, Kelaher, Baker, Harrison, & Davis, 1997; Vink, van der Velde, & Verbout, 1987), thus justifying measuring a comprehensive number of sites. However, most studies in this area measure a maximum of 14-trunk muscle sites with the majority measuring much less (Cholewicki & McGill, 1996; Cholewicki, Panjabi, & Khachatryan, 1997; Granata & Orishimo, 2001; Hodges & Richardson, 1997; Lariviere, Gagnon, & Loisel, 2000; Marras & Davis, 1998; van Dieen, Cholewicki, & Radebold, 2003). While there are redundancies for some tasks, all trunk muscles, and even different sections of the same muscles, have been shown to have different responses to varying perturbations, making a fully comprehensive protocol of measuring 24 muscle sites ideal (Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010) (See figure 2.3).
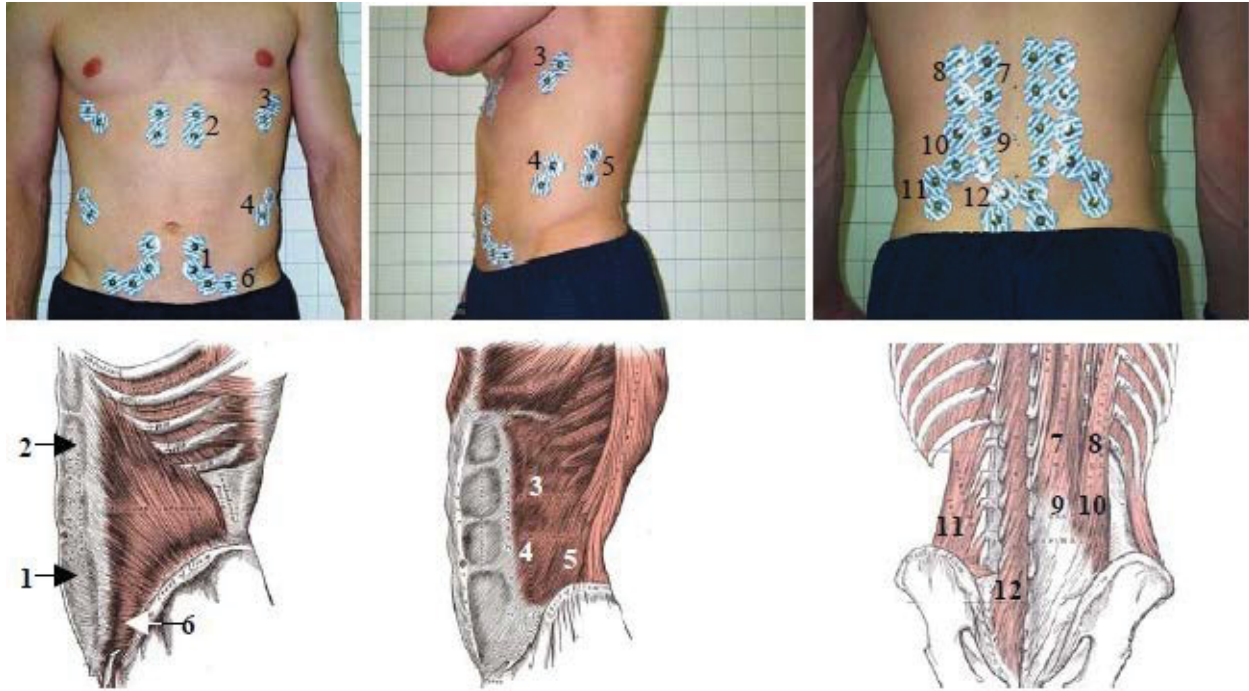
Figure 2.3: Electrode placement and anatomical representation of the muscles sites. 1=lower rectus abdominis; 2=upper rectus abdominis; 3=anterior fibres of the external oblique; 4=lateral fibres of the external oblique; 5=posterior fibres of the external oblique; 6=internal oblique; 7=longissimus at L1; 8=iliocostalis at L1; 9=longissimus at L3; 10=iliocostalis at L3; 11=quadratus lumborum; 12=multifidus (Reproduced with permission from Butler, 2007).

Managing large EMG data sets – Principal Component Analysis

To manage large data sets, data reduction techniques need to be imposed, preferably in a way that minimizes the amount of data loss. Techniques based on principal component analysis (PCA) have been shown to be an effective method to reduce the large quantity of data and make comparisons in waveform characteristics to understand co-activation and temporal synchronies among muscles (Butler, Hubley-Kozey, & Kozey, 2009b; Hubley-Kozey & Vezina, 2002). Unlike averaging and using ratios, PCA is able to maintain the salient features of the original data, and thus is an attractive alternative for data reduction (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Hubley-Kozey, Deluzio, Landry, McNutt, & Stanish, 2006; Wrigley, Albert, Deluzio, & Stevenson, 2005). A diagram of PCA is shown in Figure 2.4. Briefly, PCA works by

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transforming the EMG waveform data so that it can be explained by a limited number of principal components (PCs) that explain the majority of the variation in the data (Hubley-Kozey et al., 2006; Jackson, 1991). The waveform data (time-normalized to 101 data points) forms the matrix X (n x 101), where n is the number of subjects. A covariance matrix, S (101,101) is formed using the variances and covariances of X. Extracting the eigenvectors from the covariance matrix (S) yields an orthogonal transformation matrix, U(101,101). The eigenvectors which make up U are transformed variables that each describes a feature of variation in the data. Eigenvectors are also called principal components. Each waveform is given a Z score (also called a PC score) based on how closely it corresponds to a specific PC. PC scores can then be used in statistical analysis to compare EMG waveforms. The PCA process is explained in more detail in Appendix A.
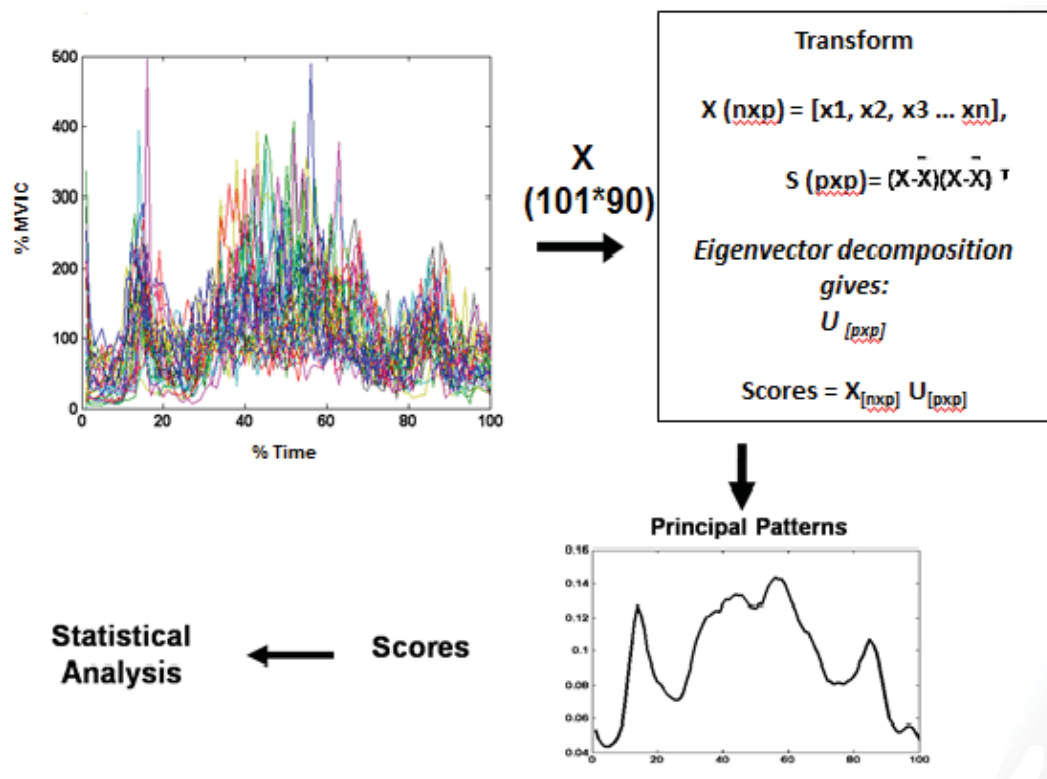
Figure 2.4: Principal Component Analysis (modified with permission from Hubley-Kozey et al., 2010).

2.9.3 Experimental procedures for tracking menstrual cycle phase

We are interested in evaluating the effect of acute fluctuation in estrogen levels over the menstrual cycle on neuromuscular control of the back. An important part of our procedure will be centered on properly selecting our test days so that this effect can be examined. It is expected that the greatest changes in neuromuscular control will occur as a result of changes in estrogen concentration in the female body. Therefore, the most appropriate times to perform our tests are when estrogen levels are at their lowest point and when they are at their highest. As previously mentioned, estrogen levels are at their lowest point during the early follicular phase (days 1-6) and at their highest point at ovulation (~day 14) (Wojtys et al., 1998). Therefore participants in this study will need to be tested in the middle of their follicular phase (~day 3-5) and as close to their ovulation as possible (~day 14).

The average menstrual cycle length for a woman between the ages of 20 and 30 is approximately 28 days long (Cole, Ladner, & Byrn, 2009; Fehring, Schneider, & Raviele, 2006). However there is still a significant amount of variability among cycle lengths. Cole et al., 2009, demonstrated that in 167 women aged 18-36, the 95% confidence interval for menstrual cycle lengths was 23-32 days. In order to determine proper test day timing, study participants need to be questioned about the timing of their last menses as well as asked about their typical cycle length. Their average cycle lengths can be used in determining the timing of the follicular phase test sessions. A woman with a 28 day cycle would have their follicular phase test session on day 2-7 of their cycle. A woman with a significantly shorter or longer cycle will have this test day adjusted accordingly to fall within 7-25% of the way through their cycle. The ovulation phase test day can be determined independently of a participant's normal cycle length. Ovulation prediction kits, which are commercially available and widely used by research groups in this

field (Abt et al., 2007; Bryant et al., 2008; Shultz et al., 2006), can be used by participants to determine when ovulation occurs. Ovulation prediction kits have been shown to be highly accurate and easy to use (Gudgeon, Leader, & Howard, 1990; Leader, Russell, Clifford, & Stenning, 1991). The ovulation prediction kits detect the LH (luteinizing hormone) surge that occurs just prior to ovulation. The test day can be scheduled for shortly after this surge (24-48hrs).

Several studies that examine the effects of sex hormone levels on soft tissue behaviour use blood samples collected on test days to confirm that subjects are being measured on the correct day (Eiling et al., 2007; Friden et al., 2003; Friden et al., 2005; Janse de Jonge et al., 2001; Lebrun et al., 1995; Shultz et al., 2006). This allows researchers to be more lenient with their scheduling of test days as they are able to confirm or reject a test day after the fact based on expected versus measured hormone levels. There are, however, some limitations of using blood samples for this purpose. Most of the studies mentioned above analyse the blood samples collected using commercially available immunoassay kits, though these kits are convenient and widely used they do have some inaccuracies and limitations associated with them. Many kits lack the proper validation and assay sensitivity and specificity necessary for reliable test results (Stanczyk et al., 2003). Stanczyk et al., 2003 recommend that because of the limited validations performed by manufacturers, researchers planning on using immunoassay kits should carry out extensive validation of each kit prior to use. Also it is important to consider the risks using phlebotomy will cause to the patients. The act of venipuncture required for drawing blood samples is acutely painful and distressing to patients (Cason & Grissom, 1997; Seemann & Reinhardt, 2000). Other researchers have published respected, well-cited work using alternative methods for determining menstrual cycle phase, choosing to omit the use of blood samples to

check hormone levels at each test date (Belanger et al., 2004; Karageanes et al., 2000). In these studies normal menstrual cycle length was used in estimating the timing of the follicular test day and basal body temperature was monitored to indicate when ovulation occurred to determine the timing of the ovulation test day. These methods have proved reliable and effective. The use of ovulation prediction kits, which are more accurate than monitoring basal body temperature (Gudgeon et al., 1990), should further improve the accuracy of choosing of a test days. This makes the use of blood samples unnecessary as they would only add undue harm to the subjects and potentially generate inaccurate results.

When performing hormone analysis it is important to take into account other factors that can influence the hormonal environment of the body. The nutritional intake of a participant prior to testing can greatly influence the hormones associated with energy substrate mobilization and use, such as insulin, glucagon, epinephrine, growth hormone, insulin like growth factor and cortisol (Tremblay, Chu, & Mureika, 1995). However, since the hormone being investigated in this study will be estradiol (the most common estrogen) fasting prior to test sessions will not be necessary. Factors such as sleep and stress can also greatly disrupt highly circadian hormones, such as sex hormones like luteinizing hormone and follicle stimulating hormone. Therefore when evaluating sex hormones, participants should be well rested prior to testing; 8 hours of sleep prior to testing should be recommended. Stress levels of participants should also be monitored between test sessions. If inadequate recovery time following physical activity has elapsed the hormone profiles of individuals can also be affected (Hackney & Viru, 2008). Participants should be asked to refrain from physical activity for 24 hours prior to testing.

Chapter 3: Methodology

3.1 Research design

The thesis was divided into two related but independent studies. The first study was a cross-sectional design with two independent groups. The second study was a repeated measure design. Approval for these studies was obtained from the Health Sciences Research Ethics Board, Dalhousie University. Methodology details specific to individual objectives of the thesis will be covered in their respective sections. Below is a description of the general methodology relevant to both studies.

3.2 Subjects

Subjects were recruited by word of mouth and through emails, posters and notices around Dalhousie University. All subjects interested in participating took part in an initial telephone interview where they were asked a series of health screening questions as well as questions concerning their eligibility for inclusion in the study. The health screening questions determined if the subject was in generally good health and whether they had any health related conditions that could have been exacerbated by participation in the study. Individual objective inclusion criteria are included in their respective sections.

3.3 Ethical issues

There were minimal risks associated with this protocol. A low risk for a rash developing from the electrode paste and /or cleaning the skin with alcohol/water mixture exists. If this occurs, it would last no more than a day. There is low risk associated with the surface EMG procedure; however, there is always a small risk of electric shock when using any electrical device. The equipment used reduces this risk since the patient unit is battery operated with a mechanism to decrease current flow, the lab has hospital grade grounds and the EMG system

meets CSA standards. There is a low risk for some mild post-exercise muscle soreness. If this occurs, it should not last more than two days.

Overall, there have been more than 150 participants including asymptomatic controls, those with low back pain and older adults tested in the lab for various studies using similar protocols with no major adverse effects reported. If a subject experiences any discomfort during testing, they are asked to report this immediately to the tester, and the activity is stopped. All test activities are directly under the participant's control and he/she was asked to push as hard as he/she could without feeling any discomfort. The subjects were permitted to rest at any time, or to withdraw from the study at any time even after testing had begun. So far after testing over 150 subjects with similar tasks there have been no negative consequences, only mild skin irritation in less than 1% of participants.

3.4 Data acquisition

A surface EMG system (3-8 channel, Bortec Inc., Calgary, Alberta) was used for recording the myoelectric signals from 24 muscle sites on the back and abdominals. Collected signals were used to determine EMG amplitudes and activity patterns. The EMG signals were pre-amplified (500x) and then amplified differentially (Bandpass 10-1000Hz; input impedance > 10GΩ; CMRR 115dB) with three AMT-8 EMG systems (Bortec Inc. Calgary, Alberta). The analog signals were sampled at 2000Hz using a 16-bit analogue-to-digital (A/D) converter (National instruments, CA-1000) using LABVIEW and were stored on an IBM Pentium computer for subsequent off-line processing.

Three battery operated patient isolation units each powered by two 9V batteries were used. Triggering signals from various sources during the tests were collected as well and were recorded on channels 24, 25 and 26 of the analogue to digital converter. The triggers consist of

simple circuits which are able to indicate when a specific event takes place that breaks the circuit. The signals collected from the triggers were synchronized with the EMG signals so that the EMG signals could be analysed in terms of timing phases (See Figure 2.2 for more info on triggers).

Twenty-four pairs of silver/silver chloride electrodes (Meditrace or Red Dot, both 10mm) were placed over the back and abdominal muscle sites (30 mm inter-electrode distance). Standardized landmarking procedures were used for placing the electrodes properly over six bilateral abdominal sites and six bilateral back sites. All electrodes were placed along the muscle fibers of the underlying muscles. Prior to electrode placement to reduce skin resistance, standard skin preparation methods were used, which include shaving hair if necessary and abrading the skin with alcohol swabs (Vezina & Hubley-Kozey, 2000). Skin impedance was checked with a multi-meter (Fluke 77) to ensure it was below 200KΩ before beginning testing. This is well below the acceptable skin/electode to amplifier impedance ratio of 1% (Soderberg, 1992).

Consistent with previous work (Butler, Hubley-Kozey, & Kozey, 2009a; Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010; Hubley-Kozey et al., 2010) the abdominal muscles that were monitored include the rectus abdominus, the external obliques and the internal obliques. The rectus abdominus was monitored at two sites, the lower and upper rectus abdominus. Electrodes were placed on the muscle belly halfway between the sternum and the umbilicus for the upper rectus abdominus (URA) and halfway between the umbilicus and pubis for the lower rectus abdominus (LRA) (Gilleard & Brown, 1994). The external oblique was measured at three different sites representing the anterior, lateral and posterior fibers. The electrodes for the anterior fibers (EO1) were placed over the 8th rib in line with the rib and adjacent to the costal cartilage (Ng, Kippers, & Richardson, 1998). The electrodes for the lateral

fibers (EO2) were placed approximately 15 cm lateral to the umbilicus at a 45 degree angle (McGill, 1991). The electrodes for the posterior fibers (EO3) were placed halfway between the lower border of the ribcage and the iliac crest on at an 80 degree angle (Nouwen, Van Akkerveeken, & Versloot, 1987). The internal obliques (IO) was measured at one site with electrodes placed at the center of the triangle formed by the inguinal ligament, lateral border of the rectus sheath and the line between the two anterior superior iliac spines (Ng et al., 1998). Though the electrode placements were standardized, minor location adjustments were made based on individual anthropometrics and palpations.

Consistent with previous work (Butler, Hubley-Kozey, & Kozey, 2009a; Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010) the back muscles that were monitored included the erector spinae (longissimus and iliocostalis) at different lumbar levels, the quadraus lumborum and the multifidus. Four sites were located over the erector spinae. Electrodes for the longissimus and iliocostalis were placed 3 and 6 centimeters lateral to the spinous process, respectively, at both the L1 and L3 lumbar levels (L13, L16, L33, L36) (Vink et al., 1987). The quadratus lumborum was measured by two sites located approximately 8.5 cm lateral to the L4 spinous process (L48). The final site, the multifidus muscle, was placed adjacent to the midline at the L5 level (L52) (Kavcic et al., 2004). As with the abdominal sites, though the electrode placements were standardized, minor location adjustments were made based on individual anthropometrics and palpations. Three ground electrodes were placed over the iliac crest as a reference for each amplifier

Validation tests were used to isolate specific muscles to ensure proper electrode function and placement for each muscle site consistent with our standard protocol (Butler, Hubley-Kozey, & Kozey, 2009a; Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010). Trunk flexion and

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abdominal hollowing exercises were used to recruit the rectus abdominis and internal obliques (Richardson, Jull, Hodges, & Hides, 1999) respectively. Isometric axial rotation and lateral flexion were performed to recruit the external obliques (McGill, 1991). Isometric trunk extension was used to recruit the back extensors (Butler, 2007). During the validation exercises the signals were checked for quality to ensure low noise and good signal with the gains on each channel adjusted to ensure maximum signal without the signal being clipped.

3.5 Normalization procedures

A series of exercises were performed to produce a maximal voluntary isometric contraction from the muscles being monitored. The purpose of these maximal voluntary isometric contractions was to obtain bench mark EMG curves to which all data is normalized in order to remove the effect of variance in EMG signals due to adipose tissue and to allow for better subject-to-subject comparisons. Adipose tissue can greatly reduce the recorded amplitude of an EMG signal (Solomonow et al., 1994). It has been estimated that adipose tissue can account for 81% of the variance seen in EMG signals (Hemingway, Biedermann, & Inglis, 1995). This method of MVIC normalization has been shown to be effective and reliable (Burden & Bartlett, 1999; Kavcic et al., 2004).

For the normalization exercises the subjects were secured with resistive straps and manual resistance was provided in the opposite direction of the intended motion to aid with the correct performance of the task (Butler, 2007). Motivation and verbal feedback was also provided to help the subject reach maximal exertion.

A total of eight exercises previously used for normalization (Butler, 2007) were performed to recruit MVICs from all muscle sites. The first exercise was a supine sit up which is performed to maximally recruit the rectus abdominus (Vezina & Hubley-Kozey, 2000). To

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activate the oblique muscle sites left and right sided exertions were performed. These included seated axial rotations to the right and left (Richardson et al., 1999) as well as right and left lateral flexion coupled with an ipsilateral hip hike in side-lying (Butler, 2007). Prone back extension (McGill, 1991) and back extension coupled with axial rotation to both the right and left (Butler, 2007) were performed to maximally recruit the back musculature. All normalization exercises were held for 3 seconds with 2 minute rest interval between each contraction. After the normalization trials baseline muscle activity (subject bias) was recorded for 3 seconds while the subject was lying supine and relaxed. The system bias was recorded for 1 second at the end of the full experimental session. Subject and system biases are recorded to allow baseline noise present in the subject or system to be filtered out during processing.

3.6 Motion measurement

The Flock of Birds ^{TM} (FOB) motion system (Ascension Technology Inc., Burlington, Vermont) was used to monitor motion of the pelvis during the TST task. One electromagnetic sensor was placed on the subject's left iliac crest (See Figure 3.1). The sensor provided 6 degrees of freedom (x,y,z displacement, yaw, pitch and roll rotations) with respect to a global coordinate axis system located in the FOB source. The measurements were not directly related to the anatomical reference frame however, yaw most closely captured anterior/posterior tilt of the pelvis, pitch most closely captured horizontal rotation (side to side) of the pelvis and roll most closely captured up and down tilt of the pelvis in the frontal plane. This system of 3-dimensional motion capture has been used in the literature (Butler, Hubley-Kozey, & Kozey, 2009a; Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010; Silfies, Bhattacharya, Biely, Smith, & Giszter, 2009) and provides information regarding pelvis motion during the trials. Minimal motion was desired and thus any trials with excess motion were excluded from analysis.
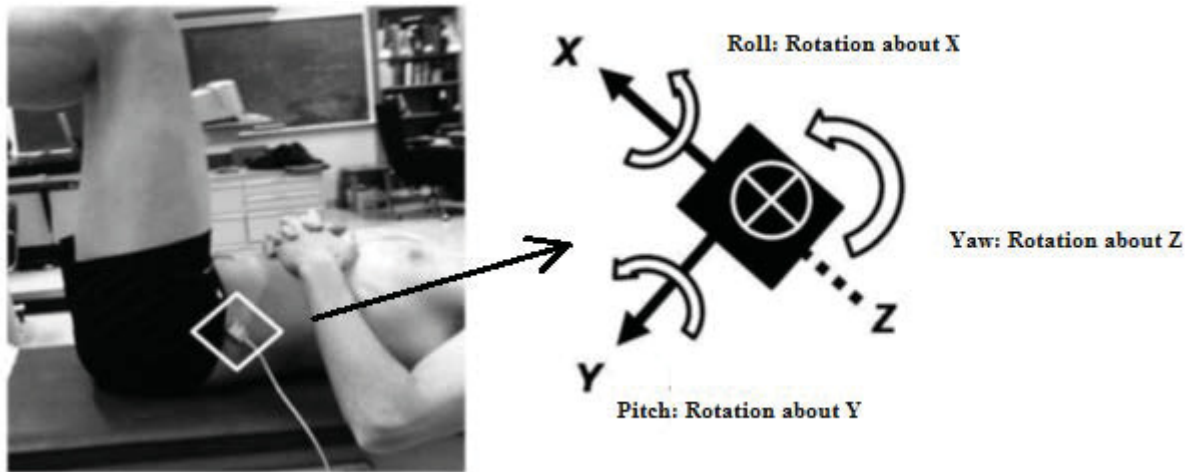
Figure 3.1: Flock of birds sensor placement on the iliac crest. Yaw describes motion around the Z axis, pitch describes motion around the Y axis and roll describes motion around the X axis (modified with permission from Hubley-Kozey et al., 2009).

3.7 Experimental task

The experimental task evaluated was the trunk stability test (TST). The rationale for selecting this task was explained in the review of literature. The TST has 5 different levels or variations (Clarke-Davidson & Hubley-Kozey, 2005; Hubley-Kozey et al., 2010). Subjects only performed TST level 3 (See Figure 2.2). This level was selected because of its asymmetrical nature. The subject begins lying supine on the bed. The test consists of the subject lifting their right leg and then their left to 90º hip flexion (thighs will meet a crossbeam), extending their right leg out to full knee extension, gently tapping the right heel on the table, returning the right thigh to the crossbeam, and lowering the left then right foot to the bed. The movement was divided into three phases: leg lifting, leg extension and leg lowering, which were defined by event markers generated by pressure sensitive switches located on the right foot, the top of the right thigh and the crossbeam.

3.8 Data processing

All raw EMG signals were visually inspected for quality and noise levels or artefacts (eg. spikes, DC offsets). Custom programs in Matlab® (MathWorks, Inc., Natick, MA. Version 7.3) were used to process the data. First a recursive fifth order Butterworth high pass filter at 30Hz was used to remove the ECG signal from any low amplitude EMG signals in which it was present (Butler, Newell, Hubley-Kozey, & Kozey, 2009). The power spectrum was calculated for each EMG signal and if low level noise from the FOB system was detected it was removed with an inverse Fast Fourier transform. The raw EMG was corrected for the system and subject bias, adjusted for the true channel gain and full wave-rectified. The signals was then filtered with a 6 Hz second order recursive Butterworth low pass filter to create a linear envelope profile. The normalization trials were used to determine the maximum amplitudes for each of the 24 muscle sites using a 500 msec moving window (Vezina & Hubley-Kozey, 2000). The maximum amplitudes were used to normalize the EMG signals for all trials resulting in EMG waveforms reported in percent maximal voluntary isometric contraction (%MVIC). All EMG signals were time normalized to 100% of the movement using a linear interpolation algorithm.

The motion data from the FOB was processed using standardized methods (Butler et al., 2010; Hubley-Kozey et al., 2009). The data was low-pass filtered at 1-Hz using a recursive second order Butterworth filter using a customized program in Matlab® (MathWorks, Inc., Natick, MA. Version 7.3). The angular displacement of each marker was calculated as the difference in degrees between the maximum and minimum angular positions during a task.

3.9 Data analysis

Normalized EMG waveforms were compared using Principal Component Analysis (PCA). PCA was done using a customized Matlab® (MathWorks, Inc., Natick, MA. Version 7.3)

program. As described in the preceding chapter as well as Appendix A, an eigenvector decomposition was performed on the covariance matrix of all of the ensemble averaged EMG profiles from each individual trunk muscle. Principal components (PCs) are extracted from the matrix which represents overall trends seen in the data. Each PC captures a different feature of the EMG waveform data. Each measured waveform was given a PC score for each PC based on how similar the waveform is to the PC curve. PC scores can then be used in statistical analysis to compare temporal synchronies between waveforms (Hubley-Kozey & Vezina, 2002; Hubley-Kozey et al., 2009).

Chapter 4: Sex Differences in Neuromuscular Control Patterns during the Trunk Stability Test

4.1 Introduction

Women have higher incidences of a wide range of musculoskeletal injuries compared to men, including low back injuries (Feuerstein et al., 1997; Krause et al., 1997; Schneider et al., 2005; Schneider et al., 2006). Physiological and anatomical differences exist between the sexes, women have smaller tendon cross-sectional areas (Pichler et al., 2008), different muscle composition (Thorstensson & Carlson, 1987) and lower musculoskeletal stiffness (Granata et al., 2002a; Granata et al., 2002b). Differences also exist between the sexes in how they control their lower limb musculature during athletic cutting and jumping manoeuvers (Ford, Myer, Toms, & Hewett, 2005; Hewett et al., 2004; Landry et al., 2009), however minimal work has been done on the trunk musculature.

In terms of low back injury prevention, neuromuscular control of the trunk is particularly important as impaired coordination of trunk musculature is associated with low back pain (Akuthota & Nadler, 2004; Hodges & Richardson, 1996; Hodges & Richardson, 1998). Having just one of the many trunk muscles responding inappropriately can disrupt spinal stability and lead to injury (McGill, 1999). Impairment in the neuromuscular control system can begin with mechanoreceptors in the ligaments and muscles which can send corrupted signals to neuromuscular control units. Neuromuscular control units will then respond with corrupted muscle response patterns which can produce higher stresses or strains on tissues leading to pain and injury (Panjabi, 2006).

Overall there has been a limited amount of research examining sex differences in neuromuscular activation patterns of trunk muscles. Higher levels of co-activation, defined as the simultaneous activity of various muscles acting around the same joint (Kellis et al., 2003), have

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been identified in women's antagonist trunk musculature compared to men's (Granata et al., 2001; Granata et al., 2001) as well as in both women's agonist and antagonist trunk musculature (Anders et al., 2007; Marras et al., 2003). It is suggested that the co-activation of trunk musculature is partially accounted for by muscular strength differences between sexes (Mannion et al., 1997; Marras et al., 2001). However, differences have been shown to exist between sexes in back extensor activation amplitudes even after tasks were adjusted to the individual's strength (Da Silva et al., 2009; Granata et al., 2001). Co-activation could instead be more prevalent in women to compensate for their reduced passive stiffness (Hsu et al., 2006; Markolf et al., 1978; Shultz et al., 2007). In that reduced stiffness of the passive tissues around a joint decreases the overall joint stability and can increase the risk of musculoskeletal injury (Granata et al., 2002; Granata, Wilson et al., 2002; Markolf et al., 1976; McGill et al., 1994), increased co-activation of muscles around a joint, both agonist and antagonist, will increase the joint's active stiffness (Cholewicki et al., 1999), thus improving joint stability (Cholewicki et al., 1998).

Co-activation can be measured by activation amplitudes (Anders et al., 2007), co-activation indices (Kellis et al., 2003) or onset and offset times (MacDonald et al., 2010). All of these measures, however, are discrete measures which have a limited capacity when attempting to describe dynamic tasks. Recent studies have instead started using pattern recognition techniques on the temporal EMG waveforms, allowing the examination of both amplitude and temporal characteristics of muscle activation for a wide group of muscles (Butler et al., 2010; Hubley-Kozey et al., 2010; Ivanenko et al., 2004; Lamoth et al., 2006; Hubley-Kozey & Vezina 2002). Principal component analysis is explained in more detail by Ivanenko et al., 2004, as well as in Appendix A, but briefly involves scoring each EMG waveform as to how similar it is to a set of principal patterns generated that describe overall trends seen in the data. These scores can

be used in statistical analysis to compare differences in EMG amplitude and temporal characteristics.

A recent study (Hubley-Kozey et al., 2011) employed these pattern recognition techniques to compare the temporal EMG responses of 24 trunk muscle sites between men and women during a controlled lifting task. In addition to higher amplitudes, they demonstrated that women had more temporal asynchronies among ipsilateral back (agonist) and abdominal (antagonist) muscle sites compared to men, indicating that women performed the task in a less coordinated fashion. The lifting task examined by Hubley-Kozey et al. recruits back extensor muscles as the main agonists for the exercise. Abdominal muscles have been shown to play a significant role in the development and treatment of low back pain (Bergmark, 1989; Ferreira et al., 2004; Gardner-Morse & Stokes, 1998), but no work has examined if men and women show similar differences in their neuromuscular control patterns during a task that primarily recruits the abdominals.

The present study examined a dynamic stability exercise that challenged the abdominal muscles. The trunk stability test (TST) is a dynamic leg lifting exercise focusing on maintaining the spine in a neutral position as well as producing coordinated activity among muscle sites to sustain lumbopelvic stability (Clarke-Davidson & Hubley-Kozey, 2005). Lumbopelvic stability involves minimizing pelvic and lumbar motion by engaging the trunk musculature in the proper sequence. Differences in muscle EMG activation amplitudes (Hubley-Kozey & Vezina, 2002; Vezina & Hubley-Kozey, 2000) and temporal EMG patterns (Hubley-Kozey & Vezina, 2002) during the TST between healthy individuals and those with chronic low back pain have been described. Furthermore temporal patterns differences were found with aging (Hubley-Kozey et al., 2009) and for those defined as physiologically stable and unstable (Hubley-Kozey et al.,

2010). However, no research has analyzed whether men and women have different amplitude or temporal EMG patterns during the TST.

The primary purpose of this study was to examine differences in muscular activation patterns of trunk musculature in men and women during a dynamic leg-lifting task, the TST. It was hypothesized that women would recruit higher overall EMG amplitudes in both abdominal and back extensors, thus more co-activation than men during the TST. It was also hypothesized that women would perform the task in a less coordinated fashion than men, thus, having more temporal asynchronies between activation profiles of different muscles during the task. A secondary purpose of this study was to assess the level of inter-subject variability within a group of women compared to that found in a male group, to examine the overall level of variability between the sexes.

4.2 Methodology

The sample used in this study was an age and BMI (body mass index) matched subset from a pre-existing database of a larger group of healthy controls. Eighteen men and 19 women aged 20 to 50 years participated after signing an informed consent form approved by the Health Sciences Research Ethics Board at Dalhousie University. A schematic of the study design is shown in Figure 4.1. All subjects were healthy as determined by a health screening questionnaire. None had any history of low back pain or cardiovascular, neurological or other orthopaedic conditions that would affect their ability to properly and safely participate in the study. Age and anthropometric measures (mass, height, waist girth) were collected. Descriptive statistics can be found in Table 4.1. Surface EMG and 3-dimensional pelvic motion was collected during the TST from 6 bilateral abdominal muscle sites and 6 bilateral back muscle sites.
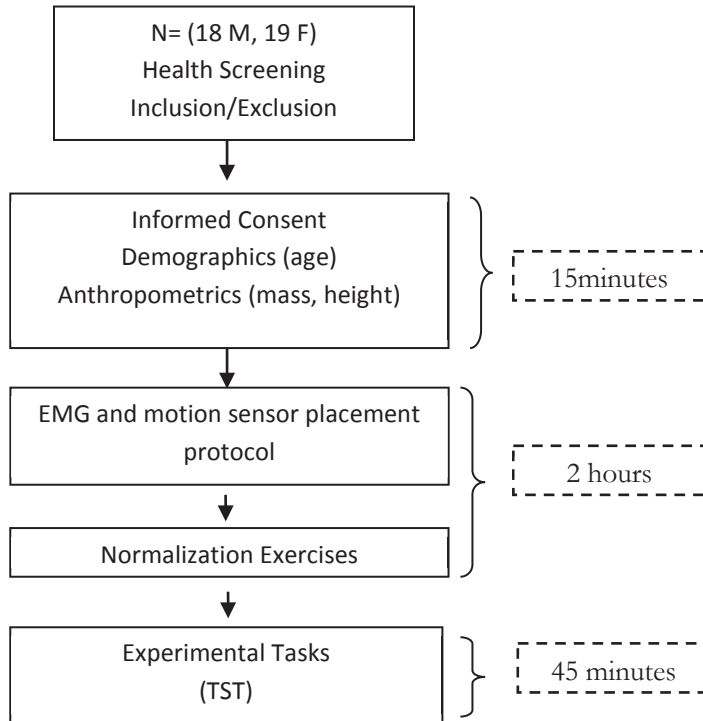
Figure 4.1: Schematic of study flow.

4.2.1 Surface EMG

The skin was prepared prior to electrode placement by shaving excess hair and abrading with aqueous alcohol swabs. Twenty four pairs of Ag/AgCL surface electrodes (.79 cm^2) (Meditrace, Graphics Control Canada Ltd.) were placed in a bipolar configuration (30 mm inter-electrode distance) over the 12-bilateral trunk muscle sites. Briefly, electrodes were placed over the upper and lower rectus abdominis (LRA & URA) (Gilleard & Brown, 1994), external oblique over anterior (EO1), lateral (EO2) and posterior fibers (EO3) (McGill, 1991; Ng et al., 1998; Nouwen et al., 1987), internal oblique (IO) (Ng et al., 1998) and lumbar levels L1 and L3 at approximately 3 and 6 cm from the midline (L13, L16, L33, L36) which correspond to the longissimus and iliocostalis muscle sites (Vink et al., 1987). In addition, electrodes were placed over the quadratus lumborum at L4 approximately 8 cm from the midline (L48) and multifidus at

L5 approximately 1-2cm from the midline (L52) (Kavcic et al., 2004). The electrode placements were standardised, with individual adjustments made where necessary based on variations in anthropometrics. In addition, a set of validation exercises were performed in an attempt to isolate individual muscles and ensure proper electrode function and placement (Kendall & McCreary, 1983) including trunk flexion and extension, rotations, hip hiking, and lateral bending.

4.2.2 Motion analysis

Angular motion of the pelvis was monitored throughout the task to ensure minimal pelvic motion using the Flock of Birds[TM] (FOB) motion system (Ascension Technology Inc., Burlington, Vermont). A magnetic sensor was placed just inferior to the mid-point of the left iliac crest (See Figure 3.1). This sensor recorded 6° of freedom (x,y and z displacements; and roll, pitch and yaw rotations) with respect to a global coordinate axis system located in the FOB source. The measurements were not directly related to the anatomical reference frame, however, yaw corresponds to anterior/posterior tilt of the pelvis, pitch corresponds to horizontal rotation (side to side) of the pelvis and roll corresponds to side flexion of the pelvis in the frontal plane.

4.2.3 Experimental task

The TST is divided into 5 separate levels. During this study participants performed three trials of level 3 of the TST, which was the only level performed (See Figure 2.2). Prior to beginning the task, subjects were instructed to perform an abdominal hollowing manoeuvre. The timing of the task was monitored using event markers generated by pressure sensitive switches located on the bottom of the right foot, the top of the right thigh, and the crossbeam.

4.2.4 Normalization exercises

Following the experimental task, eight different maximum voluntary isometric contraction (MVIC) exercises were performed (Butler, Hubley-Kozey, & Kozey, 2009b)

including a supine sit-up, sitting axial rotation to the right and left, side-lying lateral flexion to the right and left with contralateral hip hike, prone back extension and prone back extension coupled with axial rotation to the right and to the left. These exercises were designed to selectively recruit each of the 12 bilateral muscle sites. By performing a wide range of exercises it is ensured that each muscle has the opportunity to be recruited maximally. Standardized verbal feedback was provided for all MVIC exercises. Each exercise was repeated twice and a two-minute rest period was given between trials to minimize fatigue.

4.2.5 Data acquisition and processing

The EMG signals were pre-amplified (500x) and then amplified differentially (Bandpass 10-1000Hz; input impedance > 10GΩ; CMRR 115dB) with three AMT-8 EMG systems (Bortec Inc. Calgary, Alberta). The analog signal was sampled at 2000Hz using a 16-bit analogue-to-digital (A/D) converter (National instruments, CA-1000) using LABVIEW^{TM}. FOB data was simultaneously collected using LABVIEW^{TM} on a separate computer and sampled at 50Hz using a 12-bit analogue-to-digital converter (National Instruments, CA-1000). EMG and FOB data were synchronized using the event markers.

Custom programs in Matlab® (MathWorks, Inc., Natick, MA, Version 7.3) were used to process the EMG and FOB data. The details of processing are published elsewhere (Butler, Hubley-Kozey, & Kozey, 2009b). Briefly, the EMG data were filtered with a recursive fifth order Butterworth high pass filter at 30Hz to remove ECG artifact (Butler et al., 2009). The power spectrum was calculated for each EMG signal and if low level noise from the FOB system was detected it was removed with an inverse Fast Fourier transform. All EMG data were corrected for bias and gain, full wave-rectified and low pass filtered at 6Hz with a second order Butterworth recursive low pass filter to create a linear envelope profile. The linear envelope data

was time normalized to 100% movement using a linear interpolation algorithm and then amplitude normalized to the highest EMG activity from the MVIC exercises (Vezina & Hubley-Kozey, 2000). FOB motion data was filtered at 1Hz with a recursive second order Butterworth low pass filter and the maximal angular displacements for roll, pitch and yaw relative to the global coordinate system were calculated for each leg extension phase.

4.2.6 EMG data analyses

Principal component analysis (PCA) was used to extract characteristics from the EMG waveform data (Hubley-Kozey & Vezina, 2002; Jackson, 1991). PCA was applied to the time normalized EMG data to reduce the multi-dimensional nature of the data and extract important features of variability. PCA models were constructed separately for the back extensor and abdominal EMG waveforms. This was done because the temporal characteristics of the abdominal and back muscles differ, and the higher amplitude waveforms of the abdominals would have dominated the variance in the data, resulting in the back muscles being poorly described. Details of the PCA process are described in Appendix A as well as elsewhere (Hubley-Kozey & Vezina, 2002; Hubley-Kozey et al., 2009) but briefly a covariance matrix, S (101x101) was calculated from the variances and covariances of time-normalized EMG waveforms (444 x 101, 37 subjects x 12 muscles = 444) upon which an eigenvector decomposition was performed. This resulted in the formation of an orthogonal transformation matrix composed of a series of eigenvectors, also called principal components (PCs), which describe the principal patterns of variation seen in the EMG waveforms. Each EMG waveform was given a Z score, also called a PC score, based on how closely it matched each of the PCs. PC scores could then be used in inferential statistics, allowing quantitative statistical analysis of the patterns seen in the EMG data.

4.2.7 Statistical analysis

Descriptive statistics were calculated for all variables and t-tests were performed to test for sex differences. A mixed model two factor (sex, muscle) general linear ANOVA model (α=0.05) tested for differences in PC scores for the back extensors and abdominal muscles separately. Assumptions of normality and homogeneity of variance were examined with data transformations performed where necessary, using a Johnson transform. Post-hoc Tukey pair-wise comparisons were performed on any significant findings. All statistical analyses were performed using Minitab™ (Minitab Inc., State College, PA, Version 15). Inter-subject variability was assessed using standard deviations of PC scores. The standard deviations between PC scores for each muscle site were calculated for men and women separately. This resulted in 12 standard deviation values for each sex (one per muscle site) for each of the 6 PCs used (three abdominal PCs and three back extensor PCs). Differences in standard deviations in PC scores for each PC between sexes were assessed using T-tests ($\alpha = 0.05$), comparing the 12 standard deviation values for the female muscle sites to the 12 standard deviation values for the male muscle sites.

4.3 Results

Descriptive data for men and women are found in Table 4.1. No significant differences were found in age or BMI between the groups; however, men had significantly higher mass, height and waist girth compared to women.

Table 4.1: Subject demographics. Bolded values are significantly different between sexes.

	Age (yrs)	Mass (kg)	Height (cm)	BMI	Waist girth (cm)
Men (n=18)	30.1 ± 7.4	**78.8 ± 11.3**	**179.1 ± 6.5**	24.5 ± 2.8	**82.6 ± 7.3**
Women (n=19)	31.2 ± 8.6	**61.6 ± 12.0**	**164.5 ± 4.3**	22.7 ± 4.0	**73.4 ± 9.8**

Data from the FOB motion sensor indicated that pelvic motion was less than 5° in all three directions (See Table 4.2). The greatest motion was seen in roll (4.3 ± 1.7 degrees in men and 3.5 ± 1.8 degrees in women), which corresponds to side flexion of the pelvis in the frontal plane (See Figure 3.1 for visual of yaw, pitch and roll orientations). There was no significant difference between the sexes in motion in any of the directions ($\alpha = 0.05$), thus confirming that similar pelvic motion occurred during the testing and that pelvic motion did not likely add to any sex differences observed.

Table 4.2: Pelvic motion data. No significant difference ($p < 0.05$ between sexes).

	Ant/Post Tilt (Yaw)	Hor Rotation (Pitch)	Side Flexion (Roll)
Men	3.0 ± 1.9	2.3 ± 1.3	4.3 ± 1.7
Women	2.9 ± 0.9	2.7 ± 1.1	3.5 ± 1.8

Note: Ant/Post = Anterior/Poster and Hor = Horizontal.

Women performed the task in 7.2 ± 0.4 seconds, which was significantly faster than the 7.4 ± 0.4 seconds demonstrated by the men. Despite being only 3.5% faster (less than half a second), significance was achieved due to the low variability in this highly constrained task. When each phase of the motion was examined separately women performed each phase slightly faster than men, significantly so in the leg lift phase (See Table 4.3).

Table 4.3: Motion time for the entire task as well as each separate phase of the task (leg lift, leg extension and leg lower) separated by sex. Significant differences are bolded.

	Total time (s)	Leg lift (s)	Leg extension (s)	Leg lower (s)
Men	**7.4 ± 0.4**	**2.1 ± 0.2**	3.1 ± 0.3	2.2 ± 0.2
Women	**7.2 ± 0.4**	**1.9 ± 0.2**	3.1 ± 0.3	2.2 ± 0.3

4.3.1 Abdominal EMG waveform analysis

Average EMG waveforms for all abdominal muscle sites and both sexes during the TST task are depicted in Figure 4.2. Approximately 88% of the variance in the abdominal waveform

data is explained by three PCs generated by the PCA for the abdominal data (See Figure 4.3).

PC1 captured the overall amplitude and shape characteristics of the waveforms (See Figure

4.3A) and the PC1 *score* is highly correlated to RMS amplitude (Hubley-Kozey et al., 2009).

PC2 captured a difference in activity between the early part of the movement and mid-

movement. Muscles with high PC2 scores increased their activity in response to the leg

extension portion of the task, whereas muscles with low PC2 scores pre-activated to a high level

and then decreased activity and maintained a consistent level of activity throughout the entire

task (See Figure 4.3B). PC3 captured a spike in activity at about 15% movement time

corresponding to when the left leg was lifted off the bed (See Figure 4.3C) compared to activity

at the end of the task. Representative high and low scoring curves for each of the 3 PCs are also

depicted in Figure 4.3D, 4.3E and 4.3F.
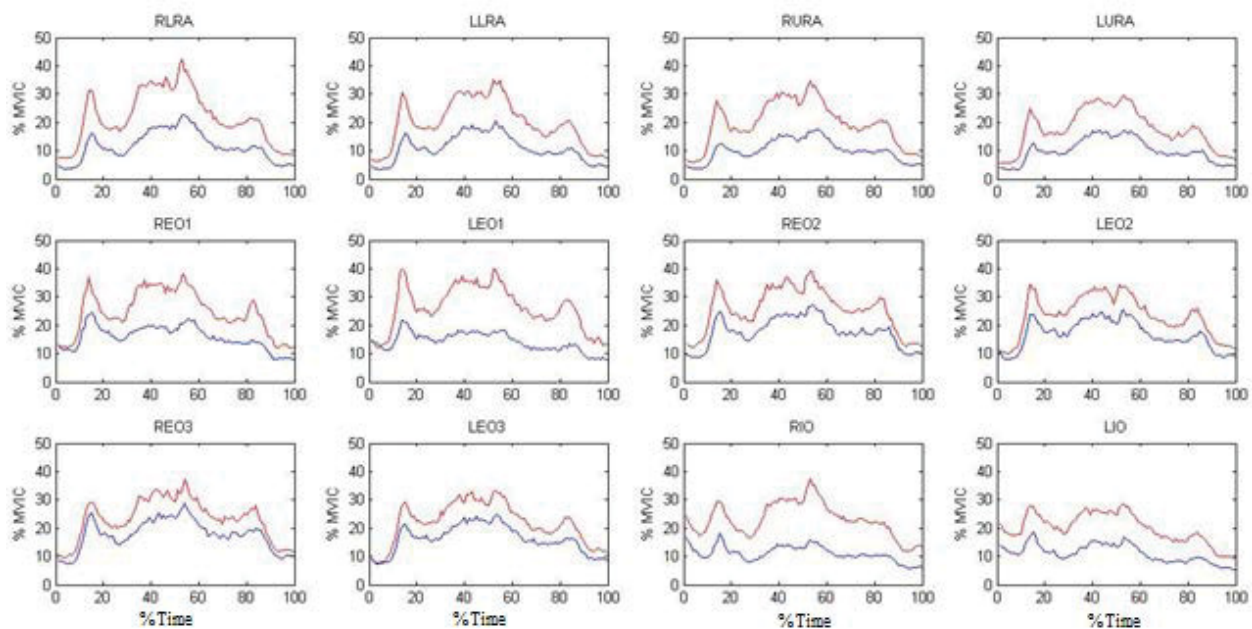
Figure 4.2: Ensemble average EMG waveforms for each of the 12 abdominal muscle sites. Averaged waveforms for **women are shown in red** and for **men are shown in blue**. RLRA = right lower rectus abdominus; LLRA = left lower rectus abdominus; RURA = right upper rectus abdominus; LURA = left upper rectus abdominus; REO1= right external oblique site 1; LEO1 = left external oblique site 1; REO2= right external oblique site 2; LEO2 = left external oblique site 2; REO3= right external oblique site 3; LEO3 = left external oblique site 3; RIO = right internal oblique; LIO = left internal oblique

Figure 4.3: First 3 principal components for the abdominal muscles. Blue shading on curves A, B and C indicate portions of the curve where the most variance in the data is described by the PC . A) PC1 which explains 79.2% of the waveform variance; B) PC2 which explains 6.1% of the variance; C) PC3 which explains 2.4% of the variance; D-F) High and low scoring curves for PCs 1-3 respectively, with high scores shown in blue and low scores shown in red, and the average high and average low scores in bold.

The ANOVA for the abdominal PC1 scores revealed significant main effects for sex ($p < 0.001$) and muscles ($p < 0.001$) as well as a significant sex and muscles interaction effect ($p=0.033$). Post hoc results for the significant interactions showed that women had significantly higher PC1 scores than men for 8 of the 12 abdominal muscle sites (RLRA, LLRA, RURA, LURA, REO1, LEO1, RIO and LIO). Women also had fewer inter-muscle differences between abdominal PC1 scores than men did. The interaction effects are shown in Table 4.4.

PC2 results indicate significant main effects for sex ($p=0.025$) and muscles ($p < 0.001$). Women have significantly higher PC2 scores in their abdominal muscle sites (1.27 ± 34) compared to men (-1.41 ± 25). The significant between muscle site differences are depicted in Table 4.4.

PC3 results indicate significant main effects for sex (p=0.004) and muscles (p < 0.001).

Women have significantly higher PC3 scores in their abdominal muscle sites (0.77 ± 21)

compared to men (-0.86 ± 17). The significant between muscle site differences are depicted in

Table 4.4.

Table 4.4: Average PC scores and standard deviations for significant interactions and muscle main effects for all 12 abdominal muscle sites for PC1-3. Statistically significant sex differences in PC1 scores bolded. Inter-muscle differences for men and for women are indicated with symbols (see Table caption).

	PC1		PC2	PC3
	Men	Women		
(1) RLRA	**-65.1 (71.9)**	**39.2 (100.4)**	20.1 (20.6) #	0.33 (21.8)
(2) LLRA	**-72.7 (62.5) #**	**20.8 (93.1)**	16.8 (17.2) #	3.38 (15.4)
(3) RURA	**-83.4 (62.5) ε**	**13.8 (103.1)**	13.4 (19.2) α	-3.60 (14.9)
(4) LURA	**-82.8 (58.8) ε**	**-3.3 (83.5)**	15.6 (14.7) #	-0.97 (12.5)
(5) REO1	**-30.0 (85.9)**	**67.1 (129.6)**	-8.2 (38.8)	4.67 (21.5) δ
(6) LEO1	**-48.1 (59.5)**	**77.6 (151.6)**	-12.5 (28.9) β	5.76 (22.1) β
(7) REO2	-7.1 (50.0)	76.2 (134.1)	-2.8 (18.6) δ	-4.86 (19.1)
(8) LEO2	-16.0 (60.7)	55.8 (101.5)	-0.46 (19.4) φ	1.21 (20.58)
(9) REO3	-5.7 (82.9)	55.2 (86.3)	0.91 (19.3) φ	-8.93 (20.15) α
(10) LEO3	-21.4 (77.3)	36.8 (121.4)	4.04 (16.8) φ	-2.88 (13.9)
(11) RIO	**-80.5 (63.5) ε**	**47.9 (91.1)**	-21.6 (31.5)	-1.35 (18.03)
(12) LIO	**-79.0 (60.0) ε**	**45.6 (163.4)**	-25.3 (51.2)	7.25 (22.0) #

Bolded numbers indicate a significant sex effect.
For PC1: For *men*: #LLRA is significantly different than REO2 and REO3; εRURA, LURA, RIO and LIO are significantly different than REO2, LEO2 and REO3. For *women:* No inter-muscle differences.
For PC2: #RLRA, LLRA and LURA are significantly different from REO1, LEO1, REO2, LEO2, REO3, LEO3, RIO and LIO; αRURA is significantly different from REO1, LEO1, REO2, LEO2, REO3, RIO and LIO; β LEO1 is significantly different from REO3 and LEO3; δREO2 is significantly different from RIO; φLEO2, REO3and LEO3 are significantly different from RIO and LIO.
For PC3: #LIO is significantly different from RLRA, RURA, LURA, REO2, REO3, LEO3 and RIO; αREO3 is significantly different from LLRA, REO1, LEO1 and LEO2; βLEO1 is significantly different from RURA, REO2 and LEO3; δREO1 is significantly different from RURA and REO2.

4.3.2. Back extensor EMG waveform analysis

EMG waveforms for all back muscle site and both sexes during the TST task are depicted

in Figure 4.4. 93.1% of the variance in the back waveform data was explained by three PCs

generated by the PCA (See Figure 4.5). As in the abdominal data, PC1 captured the overall

amplitude and shape characteristics of the waveforms (See Figure 4.5A) and PC1 *score* is highly

correlated to RMS amplitude (Hubley-Kozey et al., 2009). PC2 was a difference operator between the phases in the first 50% of the motion, it captured a lower amplitude during the first 25% and an increased amplitude during the middle of the motion (See Figure 4.5B). PC3 captured a pattern of varied amplitude in the signal over the first 20% of the movement with increased initial amplitude (See Figure 4.5C).
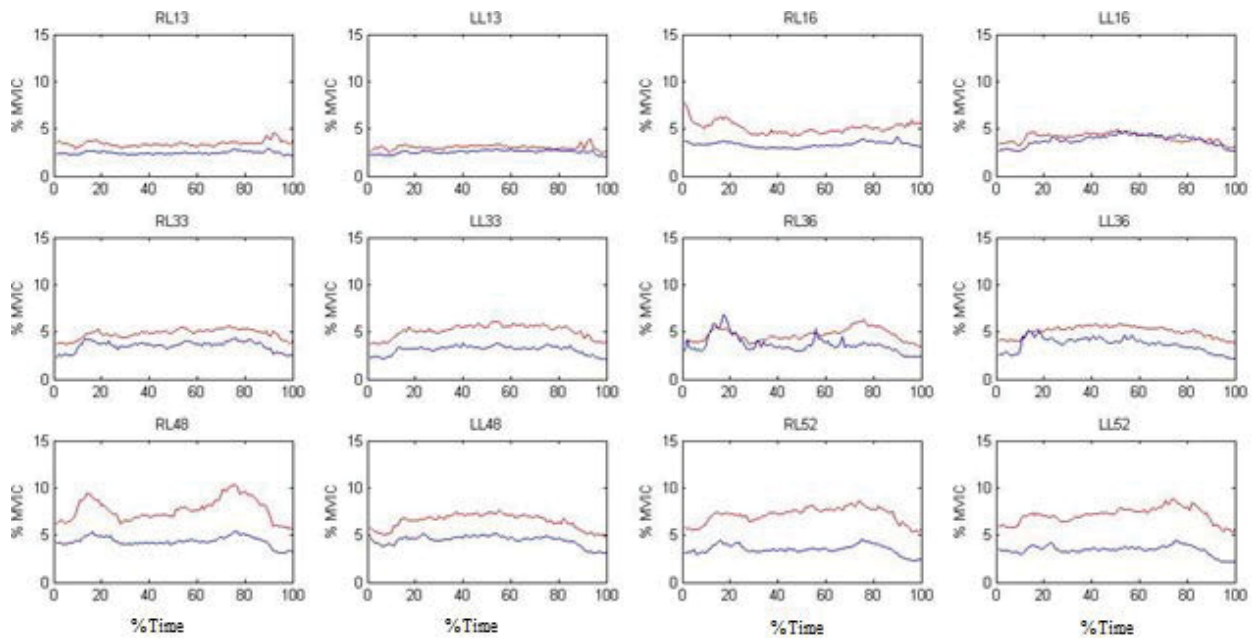
Figure 4.4: Ensemble average EMG waveforms for each of the 12 back muscle sites. Averaged waveforms for **women are shown in red** and for **men are shown in blue**. RL13= right medial erector spinae (level L1); LL13= left medial erector spinae (level L1); RL16= right lateral erector spinae (level L1); LL16= left lateral erector spinae (level L1); RL33= right medial erector spinae (level L3); LL33= left medial erector spinae (level L3); RL36= right lateral erector spinae (level L3); LL36= left lateral erector spinae (level L3); RL48 = right quadratus lumborum; LL48 = left quadratus lumborum; RL52 = right multifidus; LL52 = left multifidus.

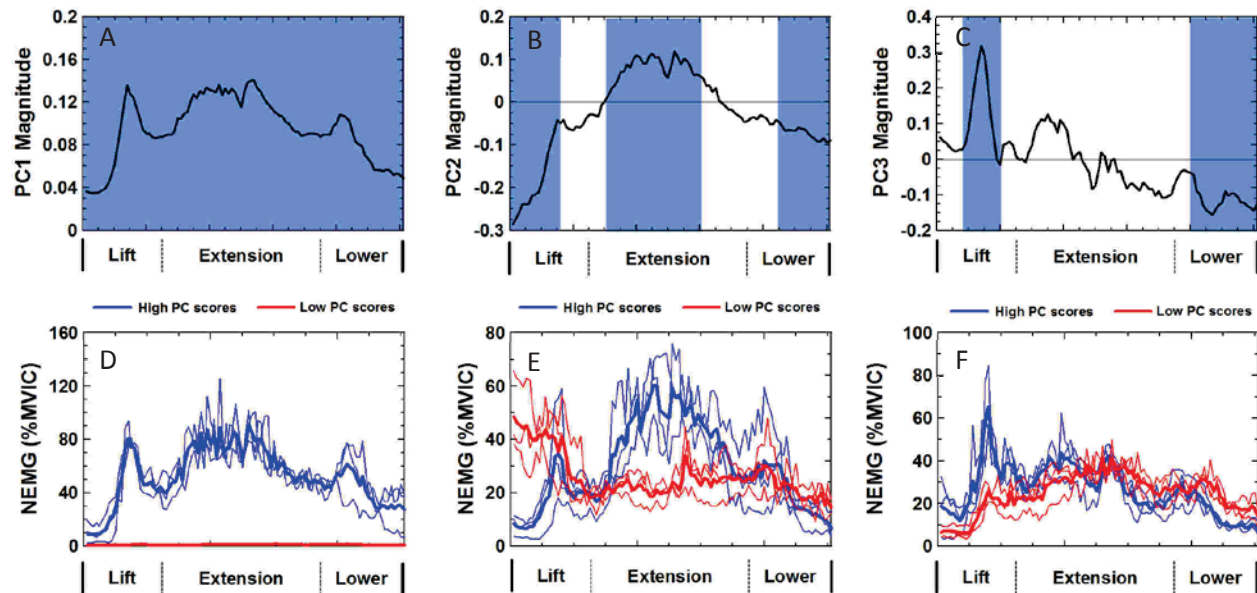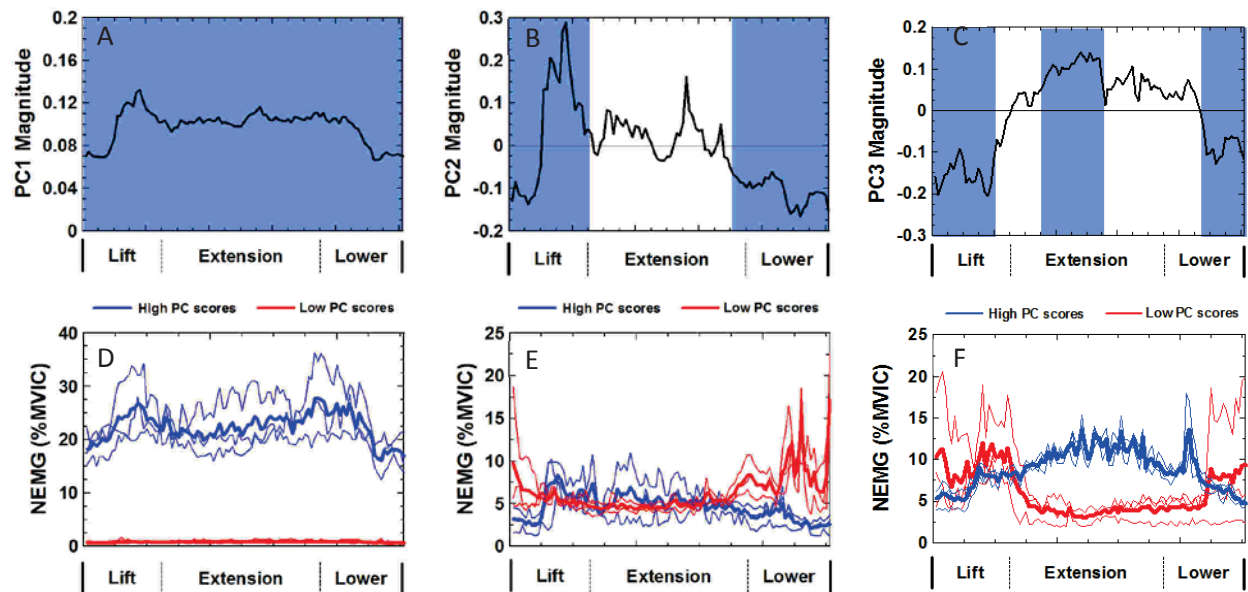**Figure 4.5:** First 3 principal components for the back muscles. Blue shading on curves A, B and C indicate portions of the curve where the most variance in the data is described by the PC. A) PC1 which explains 86.2% of the waveform variance; B) PC2 which explains 4.2% of the variance; C) PC3 which explains 2.7% of the variance; D-F) High and low scoring curves for PC1-3 respectively with high scores shown in blue and low scores shown in red, and with the average high and average low scores bolded.

PC1 results indicate significant main effects for sex (p=0.000) and muscles (p=0.000) and a significant sex and muscles interaction effect (p=0.049). Post hoc of the interaction found that women had higher PC1 scores for 5 or the 12 back muscle sites (LL33, RL36, RL48, RL52 and LL52). Women showed more inter- muscle variability in PC1 scores than men. PC1 scores for men and women for each muscle site are depicted in Table 4.5.

PC2 results indicate significant main effects for sex (p=0.000) and muscles (p=0.007).Women had significantly lower PC2 scores in their back muscle sites (-1.28 ± 5.3) compared to men (1.43 ± 9.3). The significant between muscle site differences are depicted in Table 4.5.

PC3 results indicate significant main effects for sex (p=0.028) and muscles (p=0.000) with the sex by muscles interaction effect showing a trend (p=0.061).   Women have

significantly lower PC3 scores in their back muscle sites (-0.038 ± 6.5) compared to men (0.001 ± 5.76). The significant between muscle site differences are depicted in Table 4.5.

Table 4.5: Average PC scores and standard deviations for significant interactions and muscle main effects for all 12 back muscle sites for PC1-3. Statistically significant sex differences in PC scores are bolded. Inter-muscle differences for men and for women are indicated with symbols (See caption below Table).

	PC1		PC2	PC3
	Men	Women		
(1) RL13	-19.34 (12.0) #	-10.55 (17.2) α	-0.84 (4.5)	-1.78 (4.2) #α
(2) LL13	-18.32 (13.3) #	-14.20 (11.9) αβ	0.31 (3.3)	-0.069 (3.5) δ
(3) RL16	-10.73 (15.4)	5.74 (23.9)	-2.44 (5.7)	-5.21 (8.4) β
(4) LL16	-5.73 (31.9)	-4.09 (16.7) δ	0.86 (3.7)	1.98 (7.6)
(5) RL33	-8.02 (38.8)	3.08 (36.7) δ	0.020 (4.7)	0.89 (4.6)
(6) LL33	**-12.21 (27.7) ε**	**6.17 (49.0) δ**	0.65 (3.4)	2.46 (5.7)
(7) RL36	**-6.20 (47.2) ε**	**1.63 (26.7) φ**	2.56 (18.0) #	-2.20 (8.5) #
(8) LL36	-6.17 (39.6)	4.88 (47.7) δ	2.64 (12.3) #	1.71 (3.6)
(9) RL48	**-0.12 (26.9)**	**31.18 (48.1)**	-1.71 (4.9)	-1.59 (6.0) φ
(10)LL48	0.89 (22.7)	18.90 (39.0)	0.25 (4.0)	2.06 (4.9)
(11)RL52	**-8.63 (18.3)**	**25.63 (42.9)**	-0.91 (5.8)	0.86 (5.6)
(12)LL52	**-9.32 (19.56)**	**25.12 (40.3)**	-1.39 (4.7)	0.89 (4.6)

Bolded numbers indicate a significant sex effect.
For PC1: For *men*, #RL13 and LL13 sites are significantly different from RL48 and LL48 sites; εLL33 and RL36 sites are significantly different from LL48 site. For *women*, αRL13 and LL13 sites are significantly different from RL48, LL48, RL52 and LL52 sites; βLL13 is significantly different from RL16 site; δLL16, RL33, LL33, LL36 sites are significantly different than RL48, RL52 and LL52 sites; φRL36 site is significantly different than RL48 and RL52.
For PC2: #RL36 and LL36 are significantly different from RL16 and RL48.
For PC3: #RL13 and RL36 are significantly different than LL16, RL33, LL33, LL36, LL48 and RL52; α RL13 is significantly different than LL52; βRL16 is significantly different from all muscle sites except RL13; δLL13 is significantly different from LL36 and LL48; φRL48 is significantly different from LL16, LL33, LL36, LL48 and RL52.

4.3.3 Inter-subject differences

The amount of variability between men and women was assessed by calculating the standard deviations of PC scores generated for each of the 6 PCs evaluated (see Table 4.6). While women had higher standard deviations for all PC scores compared to men, only PC1 and PC3 scores from the abdominal muscles were significantly higher (p<0.05). The differences in standard deviations for PC2 in the abdominals and PC1 in the back extensors showed a trend that nearly reached significance (p-values of 0.076 and 0.077 respectively).

Table 4.6: Average standard deviations of PC scores (PC1-3) between subjects for men and women for the abdominal and back. * equals significant at α = 0.05.

	Abdominals			Back		
	Women	**Men**	**P-Value**	**Women**	**Men**	**P-Value**
PC1	113.0 (26.3)	66.5 (10.8)	0.0000 *	38.6 (13.5)	27.8 (11.1)	0.077
PC2	27.7 (12.7)	21.0 (9.0)	0.076	6.9 (1.2)	5.8 (4.7)	0.397
PC3	20.6 (4.5)	15.5 (5.1)	0.008 *	6.6 (1.9)	5.7 (3.1)	0.245

4.4 Discussion

The dominant amplitude and temporal features in the EMG waveforms were captured by three principal patterns from the two PCA models (abdominals and back extensors). In the abdominal EMG waveforms approximately 88% of the variation was captured by these three patterns and in the back EMG waveforms the three principal patterns captured 93% of the variation (Figures 4.3 & 4.5). PC1 scores from both models captured amplitude differences (Hubley-Kozey et al., 2009) between waveforms, representing overall level of recruitment of various muscles. PC2 and PC3 from both models identified unique temporal characteristics within the waveform data set. The purpose of this study was to determine whether both amplitude and temporal differences existed in trunk muscle response patterns between men and women while performing the TST.

4.4.1 Amplitude characteristics and co-activation

First, when examining the abdominal muscles amplitude (PC1 scores), differences were found between sexes for all 12 muscle sites examined, with women showing higher PC1 scores. In 8 of the 12 abdominal muscles (RURA, LURA, RLRA, LLRA, REO1, LEO1, RIO and LIO) women had significantly higher PC1 scores compared to men (see Table 4.4). This confirms the first hypothesis that women would have higher amplitude responses to the task than men, but suggests that it is muscle specific, with all four EO2 and EO3 muscle sites not being recruited to higher levels in women compared to men. The fact that the increase in activation amplitude in

women's abdominal muscles compared to men's is not consistent across all muscle sites indicates that differences could not all be attributed to strength difference between the sexes or by a difference between the sexes in their abilities to maximally recruit their muscles through the normalizations only. If this were true, then a consistent difference in activation amplitudes between the sexes would be expected. Instead these results indicate that in addition to strength differences between the sexes there are also muscle-specific recruitment and activation amplitude differences.

Women also demonstrated fewer differences in PC1 scores among abdominal muscles than men did (see Table 4.4). Men had a total of fourteen significant differences between PC1 scores of the different abdominal muscles whereas women had no significant differences. Since the PC1 score represents the overall amplitude and shape of the EMG response this indicates that women recruited their abdominal muscles to higher yet more similar amplitudes than men, who were better able to selectively recruit specific abdominal muscles as needed. This similar amplitude has been referred to as abdominal bracing (Allison, Godfrey, & Robinson, 1998) and was found in healthy asymptomatic individuals previously for only the highest and most demanding level (level 5) of the trunk stability test (Clarke-Davidson & Hubley-Kozey, 2005). This indicates that the abdominal bracing response may only be recruited when lumbopelvic stability is most challenged. Clarke-Davidson et al., 2005, included both men and women in their subject pool and did not separate for sex, thus it is possible that had they separated their results by sex they may have found similar bracing patterns in the women's data for level 3 of the TST.

The back extensors represent the antagonist muscles during the TST task. Again, women had higher amplitudes (PC1 scores) for all 12 back muscles examined compared to men, although only 5 of the differences were statistically significant between the sexes (LL33, RL36,

RL48, RL52 and LL52). Only low levels of back extensor activations (3% MVIC) are necessary to maintain spinal stability (Cholewicki et al., 1997; Cholewicki & VanVliet, 2002), therefore, this higher level of back extensor activation in women may be adding to stability by increasing the total muscle activity, or co-activation, around the spine.

Previous authors have reported similar results of higher muscle activation amplitudes in women when compared to men, both in their agonist and antagonist muscle groups, during tasks that challenge the trunk musculature (Anders et al., 2007; Marras et al., 2003). Other authors, however, describe women having higher activation amplitudes in their antagonists only (Granata & Orishimo, 2001; Granata et al., 2001; Hubley-Kozey et al., 2011). However, not all work agrees and it has also been reported that no differences in activation amplitude levels exist between the sexes during standing (Nelson-Wong & Callaghan, 2010). Increased EMG amplitudes in women are not only explained by the sex difference in muscular strength (Mannion et al., 1997; Marras et al., 2001; Miller, MacDougall, Tarnopolsky, & Sale, 1993; Wust, Morse, de Haan, Jones, & Degens, 2008), as it has been shown that even when a task was adjusted to an individual's strength, differences still exist between activation amplitudes of men and women (Da Silva et al., 2009; Granata et al., 2001). In the present study the load was imposed on the subject's trunk musculature via force vectors from lifting their own legs, creating an external hip extensor moment. This moment is countered by contraction of the hip flexors, mainly the rectus femoris and psoas. The psoas originates from lumbar vertebrae at levels L4 and L5 and inserts on the lesser trochanter of the femur, therefore its contraction imposes downward anterior forces onto the spine. The rectus femoris originates on the anterior inferior iliac spine and inserts into the patellar tendon, therefore its contraction during hip flexion imposes forces that act to anteriorly tilt the pelvis. This moment of force magnitude will be relative to the mass of the

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subject's legs as well as the moment arm length of the legs and the speed at which the legs are lifted. Therefore, while not standardized to a set torque level, it was relative to the individual's body mass and size. In this sense, the absolute demand for the women was less than that for men, as they had significantly lower mass and height (see Table 4.1). This strengthens the possibility that sex differences in activation amplitudes seen in this study are due to more than just strength differences between the sexes since women still displayed higher activation levels even though the load they were lifting was on average less.

It has been demonstrated that women have decreased passive tissue stiffness compared to men (Hsu et al., 2006; Markolf et al., 1976; Shultz et al., 2007). Decreased stiffness of passive tissues around a joint can challenge the stability of a joint. Women have also been shown to have smoother articular cartilage in their sacroiliac (SI) joint compared to men who have rougher cartilage with more grooves and divots (Vleeming, Volkers et al., 1990). Rougher cartilage provides more friction in the SI joint which improves its stability (Bowen & Cassidy, 1981; Pool-Goudzwaard et al., 1998; Vleeming et al., 1990). This reduction in women's passive stabilizing subsystem would result in the other subsystems needing to compensate, in this case, resulting in a greater contribution to stability coming from the active and neural subsystems through muscular activation (Panjabi, 1992). Increased simultaneous muscular activation around a joint is referred to as co-activation (Kellis et al., 2003). As previously mentioned, co-activation can come in the form of agonist co-activation or antagonist co-activation. In this study, simultaneous higher activation of the abdominal muscles is referred to as agonist co-activation and simultaneous higher activation of the back extensor muscles is referred to as antagonist co-activation. Our results demonstrate that women show both higher agonist and antagonist co-

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activation compared to men, which is in agreement with results published by Anders et al., 2007, and Marras et al., 2003.

Co-activation has been shown to increase the stability of a joint, protecting it against damage caused by perturbations (Gardner-Morse & Stokes, 1998). As women have been shown to have reduced passive stiffness and less stable SI joints compared to men, it follows that they would also demonstrate more co-activation. However even though co-activation is able to benefit the spinal system through increased stability it comes with a trade-off of increasing the compressive and shear loads on the spine (Granata & Marras, 1995; Thelen et al., 1995). Antagonist co-activation increases the extra load on the spine even more than agonist co-activation, since not only is it imposing its own muscle forces, but also the presence of antagonist moments will create the necessity for increased agonist moments to produce the desired resultant force.

Co-activation, though it does increase the loading on the spine, can be beneficial if it is the only way for the spinal system to generate enough stability to protect itself from excessive movement following a perturbation. If instability were to occur it could result in injuries due to movement of the spinal tissues outside of their physiological limits (Panjabi, 1992). The higher long-term loading from co-activation may be necessary to avoid these acute, instability-related injuries. However, if a system is able to generate enough stability to prevent instability-related injuries without co-activation, through more stable passive structures etc., this may be the best alternative. Men are able to perform the TST task with less co-activation than women and still maintain the same level of low pelvic motion throughout the task (see Table 4.2). Therefore, it seems that women may have a passive stability disadvantage compared to men, supported by findings related to passive stiffness (Hsu et al., 2006; Markolf et al., 1976; Shultz et al., 2007)

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and SI joint mobility (Vleeming, Volkers et al., 1990). This reduced natural stability results in women requiring more active stability to perform the TST with minimal pelvic motion. This active stability is obtained through increased agonist and antagonist co-activation, as shown by our results as well as those of others (Anders et al., 2007; Marras et al., 2003), which increases the overall spinal loads on the female spine and may help explain the higher back injury rates seen in women compared to men.

4.4.2 Temporal characteristics

The second and third PCs captured temporal characteristics within the waveforms. For the abdominal data, PC2 captured a pattern of responsiveness to the leg lifting task reflected by low activity at the beginning and end of the task relative to in the middle of the task when the right leg was extending. The right leg extension is the most challenging part of the task due to the increased moment arm generated by the fully extended leg. Women showed significantly higher PC2 scores in their abdominal muscles compared to men. So in addition to the overall increase as illustrated by PC1, women had a selective increase in abdominal muscle activity, and therefore agonist co-activation, during the leg extension portion of the movement when there was an increased relative demand to maintain lumbopelvic stability.

PC3 captured a spike in activity seen at 15% of the movement time corresponding to when the second (left) leg was lifted off the bed compared to activity later in the task. Women had significantly higher PC3 scores in their abdominal muscles than men and this initial spike in activity corresponds with women completing this phase faster than men. Hubley-Kozey et al., 2010, reported that an increased level of activity at this time point was characteristic of individuals who were unable to maintain lumbopelvic stability while performing the TST. Therefore the higher PC3 scores seen in women may indicate that it is harder for women to

maintain lumbopelvic stability during the leg lift portion of the task compared to men. This results in women responding with a muscle response that does not fully match the demands of the task, having either misaligned feed-forward (from motor cortex) or feed-back mechanisms (from reflex pathways) resulting in an over-recruitment of motor units with women lifting their leg in a slightly faster and possibly more uncontrolled fashion.

Temporal coordination of the back extensor sites are captured with PC2 and PC3. PC2 captures responsiveness over the first 20% of the movement corresponding to when the second (left) leg is lifted off the bed. Men had significantly higher PC2 scores in their back extensors than women, indicating that their back extensor muscle activation is more responsive to the leg lifting demand. Men had significantly higher PC3 scores in their back extensor muscle sites than women, indicating that they had an increase in activity of their back extensors in the middle of the task relative to the beginning and end. This increase in activity in the middle of the task is in response to the demand on the trunk during leg extension. This finding coupled with the finding of higher EMG amplitudes in the female back extensors compared to the male back extensors indicates that women, consistent with the abdominal findings, may be using their back extensors in more of a continuous 'bracing' fashion compared to men, who demonstrate a more selective response in their back muscle activation patterns.

4.4.3 Inter-muscle variability

When comparing differences in muscle responses within women and within men, some interesting trends emerge. As mentioned earlier, women showed fewer differences among PC1 scores for individual abdominal sites than men did (see Table 4.4). This supports the idea that women have higher levels of agonist co-activation than men as they recruit all their abdominal muscles to similar amplitudes to achieve increased stability, whereas men are able to selectively

recruit abdominal muscles as needed. However when examining PC1 scores in the back extensors this trend is reversed, more inter-muscle differences are seen in women compared to men (see Table 4.5). The second hypothesis of the study was that women would have more temporal asynchronies between activation profiles of different muscles during the task compared to men. PC2 and PC3 represent temporal variations in the EMG waveforms as opposed to the overall amplitude captured by PC1. However, no significant sex and muscle interaction effects were found for any of the temporal PCs, indicating that both men and women demonstrated similar inter-muscle differences. This contrasts the significant interaction found for sex and trunk muscles reported during a lift and replace task (Hubley-Kozey et al., 2011). Women showed more EMG temporal asynchronies between ipsilateral agonist muscles (back extensors) than men. Women also showed more significant differences between sides and among ipsilateral antagonist muscles (abdominals). It is unclear why the results of these studies are different. One possible explanation is that the lift and replace task (Hubley-Kozey et al., 2011) imposed less demand on the musculature (no muscles recruited over 30% MVIC) compared to the TST, which recruited some muscle sites up to 40% MVIC. A higher demanding task could possibly increase the need for stability through co-activation, which could lead to a decrease in inter-muscle temporal differences seen as muscles would instead be recruited in a 'bracing' fashion.

4.4.4 Inter-subject variability

Another interesting finding is that women showed more inter-subject variability than men as depicted by the higher standard deviations in PC scores among women compared to among men for two abdominal characteristics (PC1 and PC3) (See Table 4.6). Women also showed higher standard deviations in their PC2 scores in the abdominals and their PC1 scores in the back extensors compared to men, though the differences did not attain significance. A similar feature

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to the spike in activity seen in PC3 in the abdominal muscles in this study was identified in another study as a main differentiator between those who could maintain lumbopelvic stability throughout the TST and those who could not (Hubley-Kozey et al., 2010). Therefore, the larger variability seen in PC3 scores in women could indicate that within the population of women tested some are able to maintain lumbopelvic stability throughout the task and some are not. As a group the abdominal muscle responses of women to the TST task were more varied than men. It is unclear why this variability exists more in women than in men. A potential explanation may be that increased variability may arise as a result of changes in the neuromuscular control system over the course of a women's menstrual cycle. Recent studies have found that neuromuscular control can be affected by the menstrual cycle in women (Dedrick et al., 2006; Friden et al., 2003; Friden et al., 2005). This suggests that the variability seen in women's data may be due to women's neuromuscular activation patterns of the trunk musculature being better suited to maintain lumbopelvic stability at one time in their menstrual cycle compared to others.

4.4.5 Conclusion

In conclusion, the results of this study partially confirm the first hypothesis showing that women recruited several of their abdominal muscles to higher amplitudes than men to perform the TST. Overall women had higher back extensor activity, thus responded to the TST task with higher amounts of agonist and antagonist co-activation compared to men. There were fewer differences among muscle sites indicative of women using a bracing strategy to complete the task and maintain stability. Women had a greater relative increase in abdominal activation during the leg extension portion of the task (PC2 scores) indicative of a greater challenge to maintain lumbopelvic stability compared to men.

Women demonstrated an increased burst in abdominal muscle activation in response to the second leg being lifted during the task compared to men, which is similar to a pattern seen in individuals who were unable to maintain lumbopelvic stability throughout the TST (Hubley-Kozey et al., 2010). Women did not show more temporal asynchronies in muscular activation patterns compared to men, which rejects the second study hypothesis. Women did show more inter-subject variations in muscle response patterns to the task than men. This suggests that as a population women respond to the task with a wider range of muscle recruitment strategies than men.

Chapter 5: Effect of the Female Menstrual Cycle on Neuromuscular Control of Trunk Musculature.

5.1 Introduction

Women sustain noncontact musculoskeletal injuries at higher rates than men. A variety of explanations have been provided, such as physiological and anatomical differences between the sexes like smaller tendon cross-sectional areas in women (Pichler et al., 2008), different muscle composition (Thorstensson & Carlson, 1987) and lower musculoskeletal stiffness in women (Granata et al., 2002; Granata, Wilson et al., 2002). One other potential mechanism that has recently been examined is that women have altered neuromuscular control patterns compared to men (Ford et al., 2003; Granata & Orishimo, 2001; Granata et al., 2001; Hewett et al., 2004; Hewett & Myer, 2011; Hubley-Kozey et al., 2011) which could increase their risk of injury. Furthermore, women, compared to men, have been shown to have a greater amount of variability in neuromuscular responses between individuals, as was shown in Chapter 4. It was hypothesized that this variability in neuromuscular responses between women may result from effects that the menstrual cycle has on neuromuscular control.

Over the course of the menstrual cycle, women experience large fluctuations in estrogen and progesterone concentrations with estrogen peaking at ovulation and progesterone rising over the second half of a woman's cycle. Women have been shown to sustain injuries at a higher rate at specific times in their menstrual cycle. ACL injuries are more prevalent during the pre-ovulatory phase of their menstrual cycle, where estrogen levels are elevating to their highest point (Hewett et al., 2007; Shultz et al., 2010). It has been suggested that the fluctuations in sex hormone levels may play a role in why women are more susceptible to injuries at specific times in their cycles. However, the effect of menstrual cycle phase on prevalence of back injuries in women is unclear.

Estrogen concentrations have been shown to be linked to mechanical tissue properties. Longer term elevated estrogen concentrations results in decreased density and synthesis rate of the stability-providing molecule collagen (Lee et al., 2004; Liu, Al-Shaikh, Panossian, Finerman, & Lane, 1997; Miller et al., 2007), and thus decreased stiffness and stability of a joint (Smith et al., 1993). Women have consistently higher levels of estrogen than men, providing a plausible explanation why women exhibit decreased tendon and ligament stiffness when compared to men (Hsu et al., 2006; Markolf et al., 1978; Shultz et al., 2007). However, the effect of shorter term changes in estrogen levels, such as over the course of the menstrual cycle, is less clear.

A large number of studies have examined the effect that the menstrual cycle has on mechanical properties of passive tissue. About equal numbers of studies have found no association between menstrual cycle phase and mechanical properties of passive tissues behavior (Arnold et al., 2002; Beynnon et al., 2005; Bryant et al., 2008; Karageanes et al., 2000; Lovering & Romani, 2005; Miller et al., 2007; Romani et al., 2003; Van Lunen et al., 2003; Warden et al., 2006) as have found an association (Deie et al., 2002; Eiling et al., 2007; Heitz et al., 1999; Romani et al., 2001; Shultz et al., 2004). A review of the literature by Zazulak et al., 2006, specifically stated that 6 out of 9 prospective cohort studies did not show any correlation between ACL laxity and menstrual cycle phase (Zazulak et al., 2006). Research has also been conducted on how the menstrual cycle could affect the injury susceptibility of a joint through systems other than passive tissues. The stability of a joint is said to be controlled by three main subsystems; the passive musculoskeletal subsystem, the active musculoskeletal subsystem and the neural feedback subsystem (Panjabi, 1992). Therefore, if the menstrual cycle does not have a clear effect on the passive musculoskeletal subsystem, it may be affecting joint stability and

injury susceptibility through either the active musculoskeletal subsystem or the neural feedback subsystem.

Previous work has reported the presences of estrogen receptors alpha and beta in skeletal muscles (Lemoine et al., 2003; Wilk et al., 2005), providing a possible link between hormone fluctuations over the menstrual cycle and the active neuromuscular subsystem. However, it may be that the menstrual cycle has a larger effect on neuromuscular control, as opposed to the specific mechanical properties of the muscular subsystem. While fluctuations in sex hormones do not appear to affect muscular fatigability or muscle strength in women (Janse de Jonge et al., 2001; Lebrun et al., 1995), a limited amount of work has examined whether acute changes in sex hormones, such as over the course of a menstrual cycle, affect neuromuscular control strategies in women.   Friden and colleagues (Friden et al., 2003; Friden et al., 2005) found that a women's postural control has the most sway in the mid-luteal phase of their cycle. Dedrick et al., 2006, found that women performing drop jumps had delayed onsets of their semitendinosus muscles during the luteal phase of the menstrual cycle as well as altered hip muscle onset timing differences compared to during the follicular phase.

Neuromuscular control of the trunk is particularly important as trunk muscle activation patterns must be coordinated to optimize dynamic trunk stability while also minimizing joint loads and risk of low back pain (Brown et al., 2006; Cholewicki & McGill, 1996; Granata & Orishimo, 2001). It has been suggested that even one muscle responding inappropriately could provide the necessary force to disrupt stability (McGill, 1999). Differences between sexes in the trunk neuromuscular activation patterns have been evaluated for a variety of tasks. Women show higher levels of co-activation compared to men during tasks that challenge the stability of the trunk (Anders et al., 2007; Granata & Orishimo, 2001; Granata et al., 2001; Marras et al., 2003).

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Co-activation is defined as the simultaneous activity of various muscles acting around the same joint (Kellis et al., 2003).

A recent study (Hubley-Kozey et al., 2011) evaluated temporal differences in EMG patterns between men and women during a controlled lifting task. Temporal characteristics, as well as amplitude characteristics of EMG waveforms, can be evaluated using a procedure called principal component analysis. Principal component analysis is described in more detail in Appendix A as well as elsewhere (Hubley-Kozey et al., 2009; Hubley-Kozey & Vezina, 2002) but briefly it involves scoring each EMG waveform on how similar it is to a set of principal patterns generated that describe sources of variability seen in the data. These scores can be used in statistical analysis to compare differences in EMG amplitude and temporal characteristics.

Hubley-Kozey et al., 2011, demonstrated that women had more temporal asynchronies among ipsilateral back and abdominal muscle sites compared to men during a controlled lifting task, indicating that women performed the task in a less coordinated fashion than men. Women had more variability between subjects in both amplitude and temporal characteristics of their muscle activation patterns in response to a dynamic leg-lifting task than men in Chapter 4 of this thesis. The task evaluated was the trunk stability test (TST), which specifically challenges lumbo-pelvic stability (Clarke-Davidson & Hubley-Kozey, 2005). Lumbo-pelvic stability is defined as the ability to maintain the pelvis and lumbar spine in physiologically safe positions and is largely controlled by moments and active stiffness created by muscular activation. Inadequate muscle strength or improper coordination between muscles can lead to lumbopelvic instability, pain and low back disorders (Pool-Goudzwaard et al., 1998). Patterns for stable and unstable responses to the TST task have been identified previously (Hubley-Kozey et al., 2010).

A potential explanation for the discrepancies between women and men in how they control their trunk musculature, as well as the greater variability seen among the female muscle activation patterns, may stem from alterations in female neuromuscular control strategies over the course of their menstrual cycle, possibly as a result of fluctuating estrogen levels. The purpose of this study was to evaluate the amplitude and temporal characteristics of EMG waveforms generated by the trunk musculature during the TST at two different times in the menstrual cycle; during the follicular phase when estrogen levels are low, and during the ovulation phase when estrogen levels are high. It was hypothesized that performing the TST during the ovulation phase would result in increased co-activation and increased temporal asynchronies between activation patterns of different muscles consistent with patterns reported for a group that was unable to maintain lumbopelvic stability (Hubley-Kozey et al., 2010).

5.2 Methodology

5.2.1 Subjects

A total of twenty-two women were screened for participation in the study, of these, twelve fit the criteria for the study. None of these subjects were included in the data set used in Chapter 4, they were all newly-recruited. Three subjects did not continue with the entire study due to menstrual cycle abnormalities (n=2) and time constraints (n=1). Nine subjects completed the study. All subjects signed an informed consent approved by the Health Sciences Research Ethics Board at Dalhousie University. All subjects were healthy, with no history of cardiovascular, neurological or musculoskeletal disorder, as determined by a health screening questionnaire. All subjects were between the ages of 20 and 30 as the prevalence of anovulatory cycles is highest in women under 20 and over 30 (Chiazze, Brayer, Macisco, Parker, & Duffy, 1968; Collett, Wertenberger, & Fiske, 1954). Subjects were included if they had a self-reported

consistent menstrual cycle for the past 3 months (ie. no missed menses, consistent length of cycle, consistent length of menses). Subjects were excluded if pregnant or breast feeding, or if they were within one year following a pregnancy. Subjects were excluded if they had used any form of oral or hormonal contraceptives (ie. contraceptive patch, vaginal hormone ring, contraceptive injections, etc.), in the past 6 months. Subjects were excluded if they had any low back pain episodes within the past 12 months or if they had ever experienced a low back injury severe enough to require medical assistance. Athletes performing more than moderate volumes of aerobic conditioning (over 3 hours of aerobic activity per week) were excluded from the study as they have been shown to have higher rates of menstrual cycle oddities (Vescovi, 2011). Postural assessments were performed on all subjects under the direction of a certified physiotherapist to ensure no postural abnormalities (kyphosis, scoliosis etc.). Age and anthropometric measures (mass and height) were collected. Descriptive statistics for the nine participants who completed the study can be found in Table 5.1. All subjects were tested at two separate phases over one menstrual cycle corresponding to their follicular phase (~day 2-6) and their ovulation phase (~day 14). Other studies have tested over several menstrual cycles to account for intra-subject variability in cycles but no differences were found between any of the cycles measured (Friden et al., 2003; Friden et al., 2005). Since testing was occurring on two separate days the test-retest reliability of the study protocol was assessed prior to beginning the study with a small reliability study, details are included in Appendix B. The reliability study revealed acceptable reliability and provided guidance to further improve reliability.

5.2.2 Procedure

Participants attended an initial familiarization session, during which informed consent was obtained and they received instruction in the experimental tasks that would be performed on

test days, in an attempt to eliminate a training effect. The subject's demographics (age and sex) were recorded as well as structural anthropometric parameters: body mass (kg), height (cm), and standing elbow height (cm). A brief postural assessment was performed by the principal investigator, under the supervision of a physiotherapist, to ensure there was no evidence of spinal abnormalities such as scoliosis. Outline of procedures are depicted in Figure 5.1.
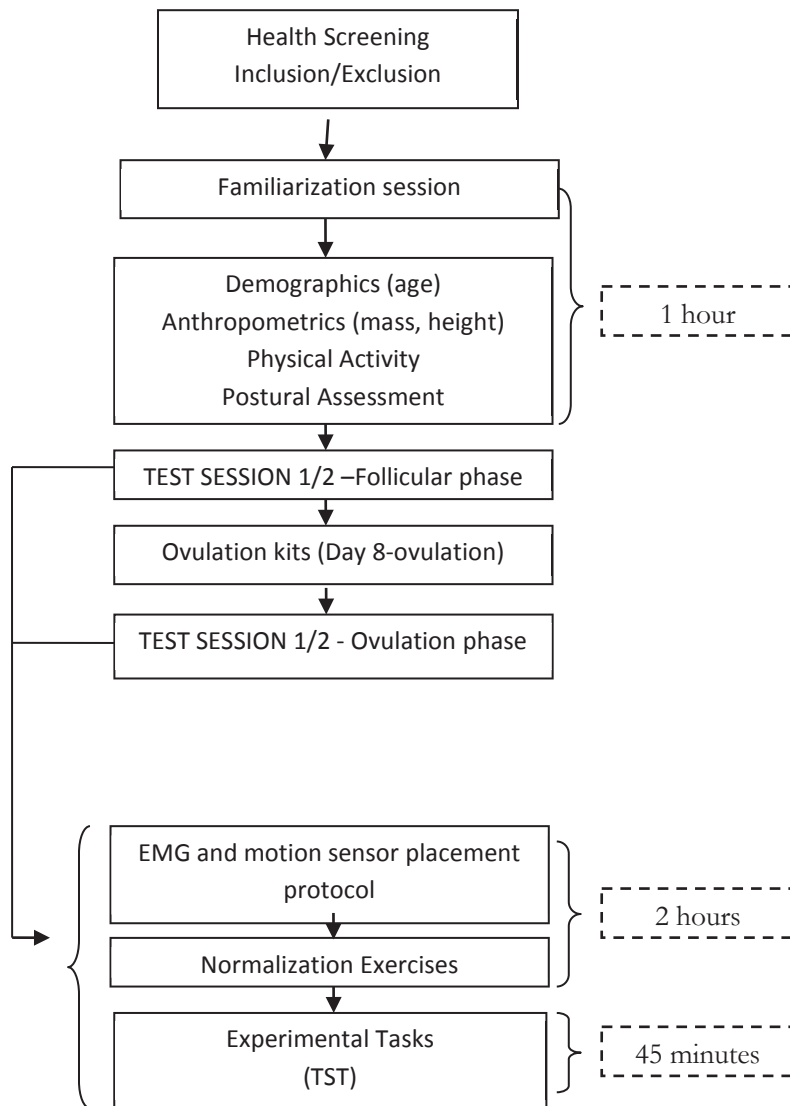
Figure 5.1: Flow chart of the experimental procedures and an overview of research variables that were measured and associated length of time. Participation time was approximately three hours per session.

5.2.3 Hormone analysis

Subjects were tested at two distinct points in their menstrual cycles when their estrogen levels were the most varied; (1) follicular phase, defined as 2-7 days after the beginning of menses and (2) ovulation phase, defined as 24-48 hrs after the pre-ovulation luteinizing hormone (LH) surge. Approximately half of the subjects had their follicular phase test session first (n=5) whereas the others had their ovulation phase test session first (n=4). Test days were selected based on self-reported average menstrual cycle lengths and ovulation prediction kit results. A woman with a 28 day cycle would have had her follicular phase test session on day 2-7 of their cycle. If a woman reported a significantly shorter or longer cycle ($\leq$ 24 days or $\geq$ 32 days), this test day would have been adjusted accordingly to fall between 7% and 25% of the length of their cycle. However no adjustments were needed as none of the subjects fell outside of the 24-32 day cycle window. The ovulation phase test day was determined independent of the subject's normal cycle length. Ovulation prediction kits, which detect the surge in LH that occurs just prior to ovulation, were used by the subject from day 8 of the cycle to ovulation. The subject came in for testing within 24 and 48 hrs of detection of the LH surge. Commercially available ovulation prediction kits are widely used by research groups in this area (Abt et al., 2007; Bryant et al., 2008; Shultz et al., 2006) and have been shown to be highly accurate and easy to use (Gudgeon et al., 1990; Leader et al., 1991).

Sleep, stress, nutritional intake and physical activity were monitored with a pre-testing questionnaire asking about sleep, stress, nutritional intake and physical activity in the past 24 hours for descriptive purposes. To evaluate stress levels the Daily Stress Index (Brantley, Waggoner, Jones, & Rappaport, 1987) was used. This index is a 58 item self-report measure designed to quantify an individual's daily stress based on events that they have experienced in

the past 24 hrs. It has been proven to have high validity and reliability (Brantley et al., 1987) and

has been shown to have convergent validity with biochemical measures of daily stress (Brantley,

Dietz, McKnight, Jones, & Tulley, 1988).

5.2.4 Data acquisition

Excess hair was shaved and the skin abraded with aqueous alcohol swabs prior to

electrode placement. Twenty four pairs of Ag/AgCL surface electrodes ($.79cm^2$) (Red Dot,

Graphics Control Canada Ltd.) were placed in a bipolar configuration (30 mm inter-electrode

distance) over the 12-bilateral trunk muscle sites. Briefly electrodes are placed over the upper

and lower rectus abdominis (LRA & URA) (Gilleard & Brown, 1994), external oblique over

anterior (EO1), lateral (EO2) and posterior fibers (EO3) (Ng et al., 1998; McGill, 1991; Nouwen

et al., 1987), internal oblique (IO) (Ng et al., 1998), lumbar levels L1 and L3 at approximately 3

and 6 cm from the midline (L13, L16, L33, L36) which correspond to the longissimus and

iliocostalis muscle sites (Vink et al., 1988), quadratus lumborum at L4 approximately 8 cm from

the midline (L48) and multifidus at L5 approximately 1-2cm from the midline (L52) (Kavcic et

al., 2004). The electrode placement positions were standardized but individualized modifications

could still be made to account for individual differences in body anthropometrics.

5.2.5 Motion capture

3-dimensional motion and rotation of the pelvis was monitored during the task to ensure

that a neutral posture was maintained using the Flock of Birds[TM] (FOB) motion capture system

(Ascension Technology Inc., Burlington, Vermont). A magnetic sensor was placed just inferior

to the mid-point of the left iliac crest (See Figure 3.1). This sensor recorded 6º of freedom (x,y

and z displacements; and roll, pitch and yaw rotations) with respect to a global coordinate axis

system located in the FOB source. The measurements were not directly related to the anatomical

reference frame however, yaw most closely represents anterior/posterior tilt of the pelvis, pitch most closely represents horizontal rotation (side to side) of the pelvis and roll most closely represents up and down tilt of the pelvis in the frontal plane.

5.2.6 Normalization exercises

Prior to the experimental task eight different maximum voluntary isometric contraction (MVIC) exercises were performed, which are each targeted at eliciting a maximal contraction from one or more of the muscles of interest (Butler, Hubley-Kozey, & Kozey, 2009b). They included a supine sit-up, seated axial rotation to the right and left, side-lying lateral flexion to the right and left, prone back extension and prone back extension coupled with axial rotation to the right and to the left. The range of exercises was selected to ensure that each muscle had the opportunity to be recruited maximally. Each exercise was repeated twice with a two-minute rest period given between trials to reduce the risk of fatigue developing. Standardized verbal feedback was provided for all MVIC exercises.

5.2.7 Experimental task

Five trials of the trunk stability test (TST) level 3 were performed (Figure 2.2). Prior to beginning the task subjects were instructed to perform an abdominal hollowing maneuver. Time points in the task were defined by event markers generated by pressure sensitive switches located in a foot pad on the surface of the bed as well as on the crossbeam where the subject's thighs made contact.

5.2.8 Data acquisition and processing

The EMG signals were pre-amplified (500x) and then amplified differentially (Bandpass 10-1000Hz; input impedance > 10GΩ; CMRR 115dB) with three AMT-8 EMG systems

(BortecInc. Calgary, Alberta). The analog signal was sampled at 2000Hz using a 16-bit analogue-to-digital (A/D) converter (National instruments, CA-1000) using LABVIEW. FOB data was simultaneously collected using LABVIEW^{TM} on a separate computer and sampled at 50Hz using a 12-bit analogue-to-digital converter (National Instruments, CA-1000). EMG and FOB data is synchronized using the event markers.

Custom programs in Matlab® (MathWorks, Inc., Natick, MA, Version 7.3) were used for processing. The details of processing are published elsewhere (Butler, Hubley-Kozey, & Kozey, 2009b). Briefly, the EMG data was filtered with a recursive fifth order Butterworth high pass filter at 30Hz to remove ECG artifact (Butler et al., 2009). The power spectrum was calculated for each EMG signal and if low level noise from the FOB system was detected it was removed with an inverse Fast Fourier filter. All EMG data were corrected for bias and gain, full wave-rectified and low pass filtered at 6Hz with a second order recursive Butterworth low pass filter to create a linear envelope profile. The linear envelope data was time normalized to 100% movement using a linear interpolation algorithm and then amplitude normalized to MVIC (Vezina & Hubley-Kozey, 2000). FOB motion data was filtered at 1Hz with a recursive second order Butterworth low pass filter and the maximal angular displacements for yaw, pitch and roll during the motion relative to the global coordinate system were calculated.

5.2.9 EMG data analyses

Principal component analysis (PCA) (Hubley-Kozey & Vezina, 2002; Jackson, 1991) was used to reduce the multi-dimensional nature of the EMG data and capture the amplitude and temporal variations within the EMG data. Details of the PCA process are described in Appendix A as well as elsewhere (Hubley-Kozey et al., 2009; Hubley-Kozey & Vezina, 2002) but briefly a covariance matrix, S (101x101) was calculated from the variances and covariances of time-

normalized EMG waveforms upon which an eigenvector decomposition is performed. Data from the reliability study (5 subjects tested twice) in Appendix B was added to increase the strength of the PCA model. Increasing the number of curves included in the model increases the ability of the model to describe overall trends seen in the data as opposed to features seen in single subjects. Therefore the EMG waveform data matrix had the dimensions of (336 x 101, 14 subjects x 2 sessions x 12 muscles = 336). The eigenvector decomposition results in the formation of an orthogonal transformation matrix (101,101) composed of a series of eigenvectors, also called principal components (PCs), which describe the principal patterns of variation seen in the EMG waveforms. Each EMG waveform is given a Z score, also called PC scores, based on how closely it matches each of the PCs. PC scores can then be used in inferential statistics. This allows quantitative statistical analysis of the patterns seen in the EMG data. Two separate PCA analyses were performed, one for the abdominal muscle EMG waveforms, and one for the back muscle EMG waveforms because the temporal characteristics of the abdominal and back muscles were different and the abdominal muscles higher amplitude waveforms would have dominated the variance in the data, resulting in the back muscles being described poorly.

5.2.10 Statistical analysis

A repeated measure two factor (menstrual cycle phase, muscle) general linear ANOVA model ($\alpha = 0.1$) was used to test for differences in PC scores for the back extensors and abdominal muscles separately. Assumptions of normality and homogeneity of variance were examined with data transformations performed where necessary. For this preliminary study post-hoc analysis was performed using non-parametric sign tests (alpha level set at 0.1) due to the small sample size restricting the use of parametric tests. Statistical analysis was performed using

a combination of SPSS (PASW Statistics v. 17.0.3, SPSS: An IBM Company) and Minitab[TM] (Minitab Inc., State College, PA, Version 15).

5.3 Results

Subject demographics are depicted in Table 5.1. There were no significant differences between sleep and stress measures between sessions (See Table 5.2). High and low stress DSI Sum scores were defined by Brantley in 1988 as 37.94 ± 20.34 and 9.81 ± 6.46 respectively (Brantley et al., 1988). Therefore, subjects had high stress levels overall but no difference in stress levels between test days. Subjects had on average 7.5 ± 1.75 hours of sleep the night prior to testing.

Table 5.1: Subject demographics

Age (yrs)	Mass (Kg)	Height (cm)	BMI
22.8 ± 2.6	63.1 ± 9.9	167.1 ± 4.0	22.6 ± 3.2

Table 5.2: Comparison of sleep and stress measures between test sessions. No significant differences ($p< 0.1$) found between sessions for hours of sleep or stress measures.

Measure	Follicular Phase	Ovulation Phase	P-value
Hours of Sleep	7.7 ± 1.6	7.3 ± 1.9	0.55
DSI Sum	37.0 ± 22.7	35.6 ± 16.3	0.88
DSI Frequency	13.8 ± 6.4	13.9 ± 6.0	0.96

The pelvic motion sensor captured less than 5° of motion in all three directions over the entire task (See Table 5.3). Differences in movement between the two phases were less than 1.5 degrees for all three directions (yaw (Z), pitch (Y) and roll (X)) and none of these differences were significant. The measurements were not directly related to the anatomical reference frame however, yaw most closely captures anterior/posterior tilt of the pelvis, pitch most closely captures horizontal (side to side) rotation of the pelvis and roll most closely captures up and

down tilt of the pelvis in the frontal plane. When the motion data was divided into movement phases (leg lift, leg extension and leg lower) there were still no significant differences between the two menstrual phases. This data confirms that minimal pelvic motion occurred. There was no statistically significant difference in time it took to complete the task during the follicular phase (7.9 secs ± 0.3 secs) compared to during the ovulation phase (7.9 secs ± 0.2 secs).

Table 5.3: Pelvic motion data

	Follicular phase	**Ovulation phase**	**P-value**
Ant/Post Tilt (Yaw)	3.57 (± 0.86)	3.85 (± 1.46)	0.727
Hor Rotation (Pitch)	4.65 (± 2.82)	3.30 (± 1.51)	0.389
Side Flexion (Roll)	3.41 (± 0.98)	2.81 (± 1.32)	0.401

Note: Ant/Post = Anterior/Poster and Hor = Horizontal.

5.3.1 Abdominal EMG waveform analysis

EMG waveforms for all abdominal muscles during the follicular phase and the ovulation phase are depicted in Figure 5.2. Minimal variation in muscle activation throughout the movement in the trunk sites was confirmed by PCA since 90.2% of the variance in the data was explained by 3 PCs (See Figure 5.3). PC1 captured the overall amplitude and shape characteristics of the waveforms (See Figure 5.3A) and PC1 *score* is highly correlated to RMS amplitude (Hubley-Kozey et al., 2009). PC2 captured difference in amplitude between the beginning of the movement and the middle of the movement when the right leg is being extended. Muscles with high PC2 scores increased their activity in response to the leg lifting task whereas muscles with low PC2 scores pre-activated and maintained a consistent level of activity throughout the entire task (See Figure 5.3B). PC3 captured a spike in activity at about 15% movement time corresponding to when the left leg was lifted off the bed (See Figure 5.3C). Representative high and low scoring curves for each of the 3 PCs are also depicted in Figure 5.3D, 5.3E and 5.3F.
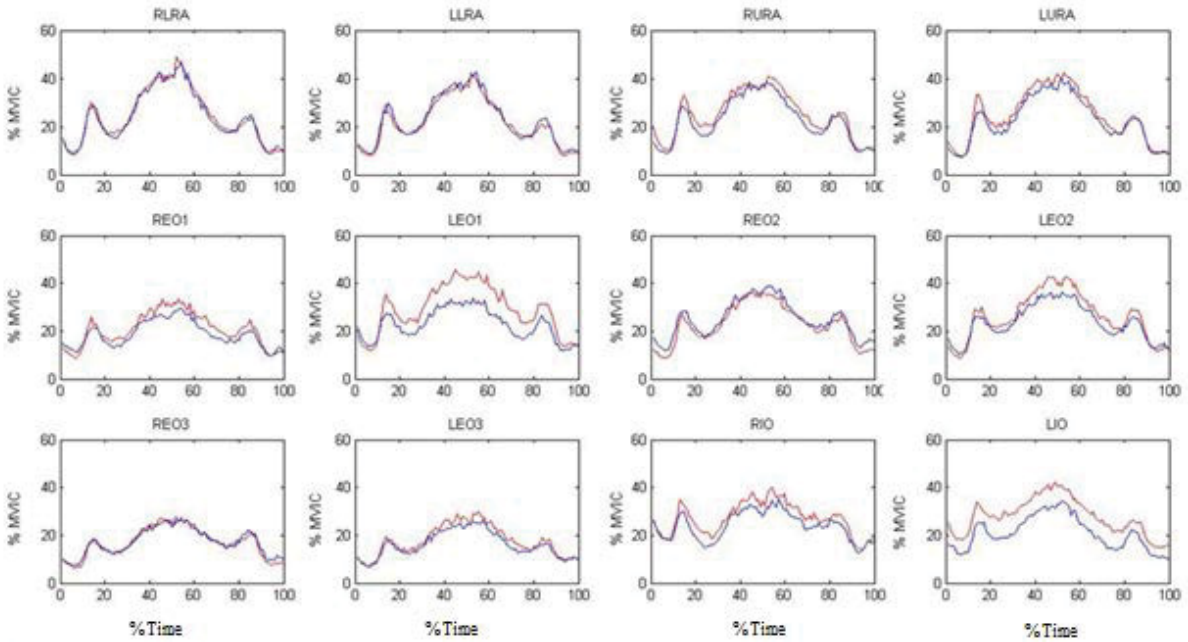
Figure 5.2: Ensemble average EMG waveforms for each of the 12 abdominal muscle sites averaged across subjects for the follicular phase (red) and the ovulation phase (blue). RLRA = right lower rectus abdominus; LLRA = left lower rectus abdominus; RURA = right upper rectus abdominus; LURA = left upper rectus abdominus; REO1= right external oblique site 1; LEO1 = left external oblique site 1; REO2= right external oblique site 2; LEO2 = left external oblique site 2; REO3= right external oblique site 3; LEO3 = left external oblique site 3; RIO = right internal oblique; LIO = left internal oblique.
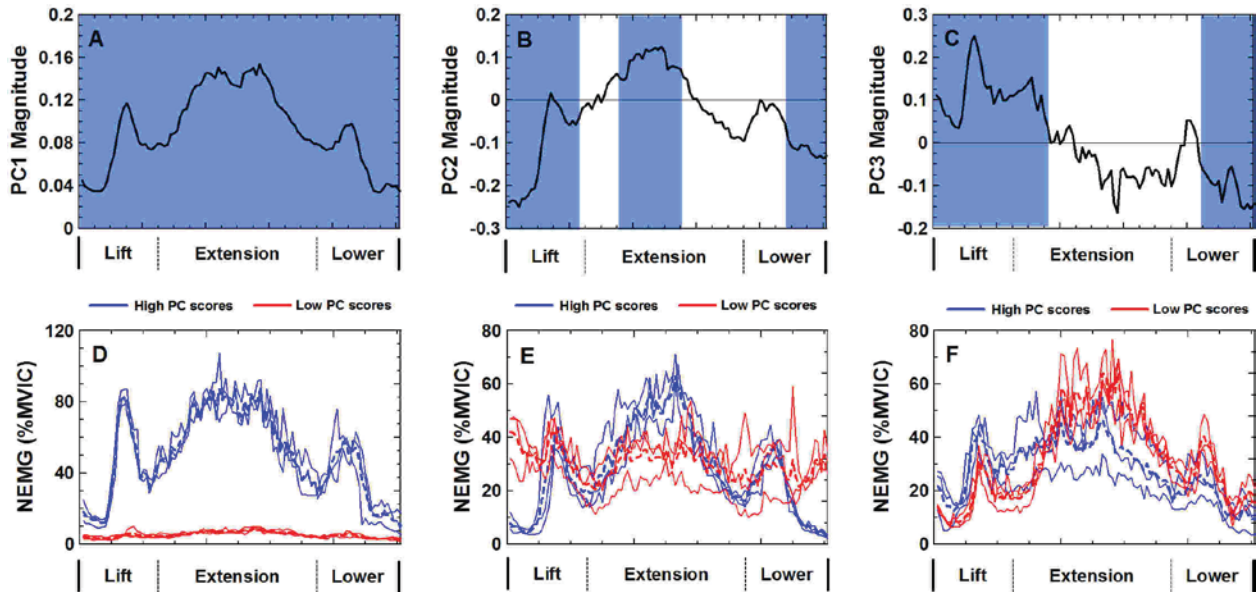
Figure 5.3: First three principal components for the abdominal muscles. Blue shading on curves A, B and C indicate portions of the curve where the most variance in the data is described by the PC. A) PC1 which explains 83.1% of the waveform variance; B) PC2 which explains 4.8% of the variance; C) PC3 which explains 2.3% of the variance; D-F) High and low scoring curves for PC1-3 respectively with high scores shown in blue and low scores shown in red and the average high and average low scores bolded.

PC1 results indicate significant main effects for phase (p=0.092) and muscles (p=0.005) only. The abdominal muscles showed a significantly higher average PC1 score during the follicular phase (-1.9 ± 112) compared to during the ovulation phase (-21.1 ± 119). The significant between muscle site differences are shown in Figure 5.4.
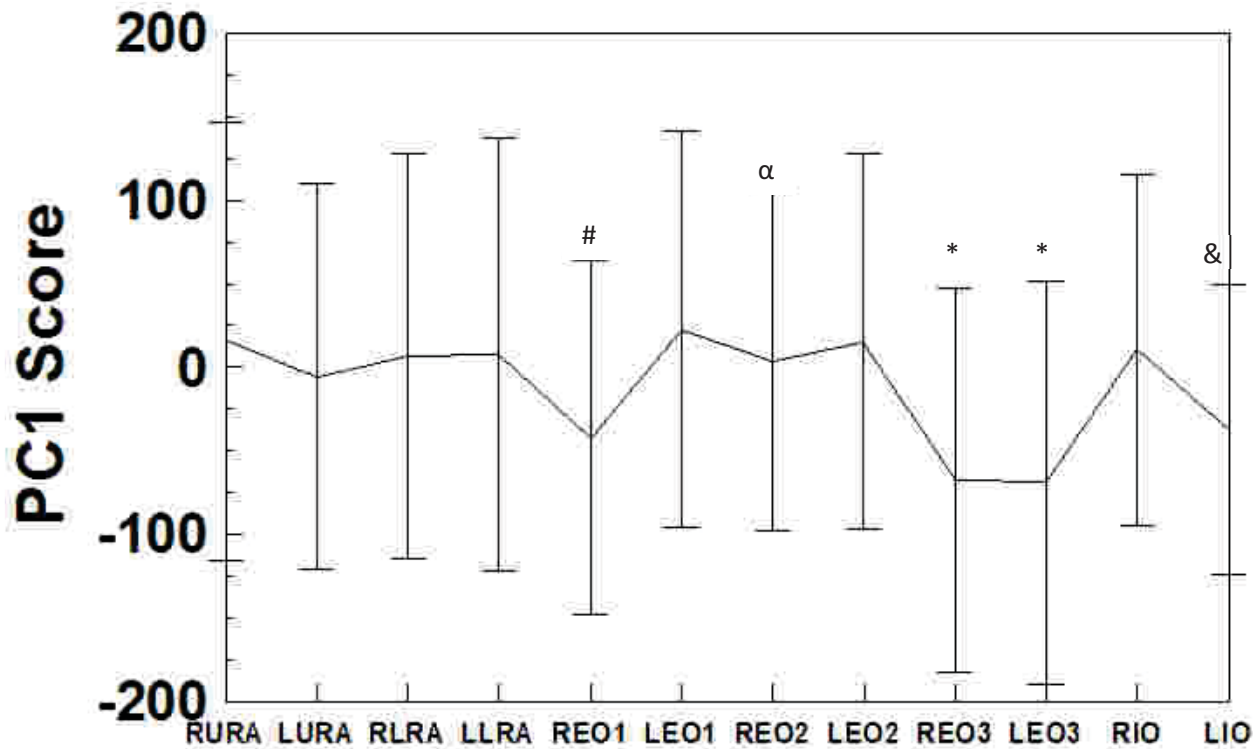
Figure 5.4: Mean PC1 scores and standard deviations for each of the 12 abdominal muscle sites. *REO3 and LEO3 are less than RURA, LURA, RLRA, LLRA, LEO1, REO2, LEO2 and RIO; #REO1 is less than RURA, LURA, LEO1 and REO2; &LIO is less than LLRA and RIO; αREO2 is less than LEO1.

PC2 results indicate a significant main effect for muscles (p=0.000) and a significant phase and muscles interaction effect (p=0.070). Mean PC2 scores for each of the 12 abdominal sites during the two different menstrual cycle phases are shown in Figure 5.5. Non-parametric post hoc tests revealed that REO2 had significantly lower PC2 scores during the ovulation phase compared to during the follicular phase. The between muscles differences for each phase are shown in Figure 5.5.
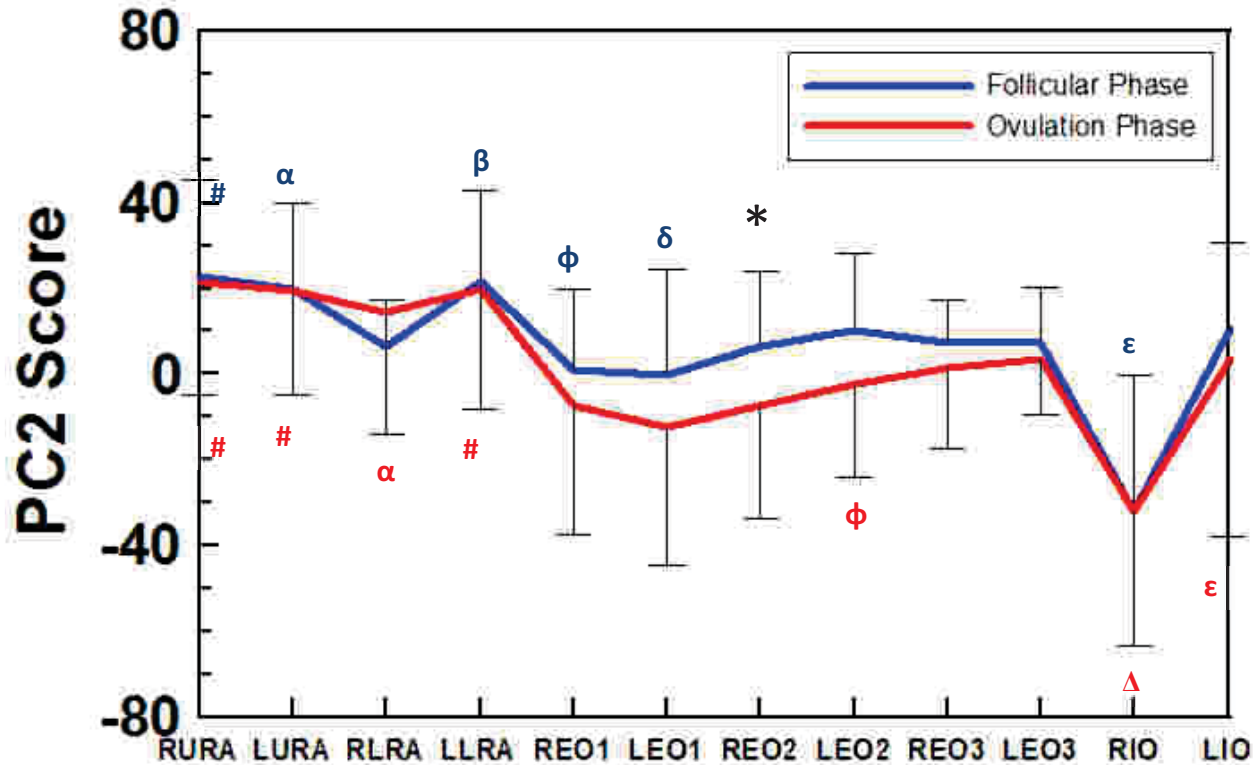
Figure 5.5: Mean PC2 scores and standard deviations for each of the 12 abdominal muscle sites during the follicular phase and the ovulation phase. * indicates a muscle site with a significant difference between phases.

For the Follicular Phase: #RURA is significantly different from RLRA, REO1, REO2 and LEO2. αLURA is significantly different from RLRA, REO1, LEO1, REO2 and LEO2. βLLRA is significantly different from RLRA, REO1, LEO1, REO2 and LEO2. φREO1 is significantly different from REO2 and LEO2. δLEO1 is significantly different from LIO. εRIO is significantly different from RURA, LURA, RLRA, LLRA, REO2, LEO3 and LIO.

For the Ovulation Phase: : #RURA, LURA and LLRA are significantly different from REO1, LEO1, REO2, LEO2, REO3 and RIO. αRLRA is significantly different from REO1, REO2, LEO2, REO3 and RIO. φLEO2 is significantly different from LEO1, REO3, RIO and LIO. ΔRIO is significantly different from REO2, REO3 and LEO3. εLIO is significantly different from LURA, REO1 and RIO.

PC3 results indicate a significant main effects for muscles (p=0.008) only. The significant between muscle site differences are displayed in Figure 5.6.
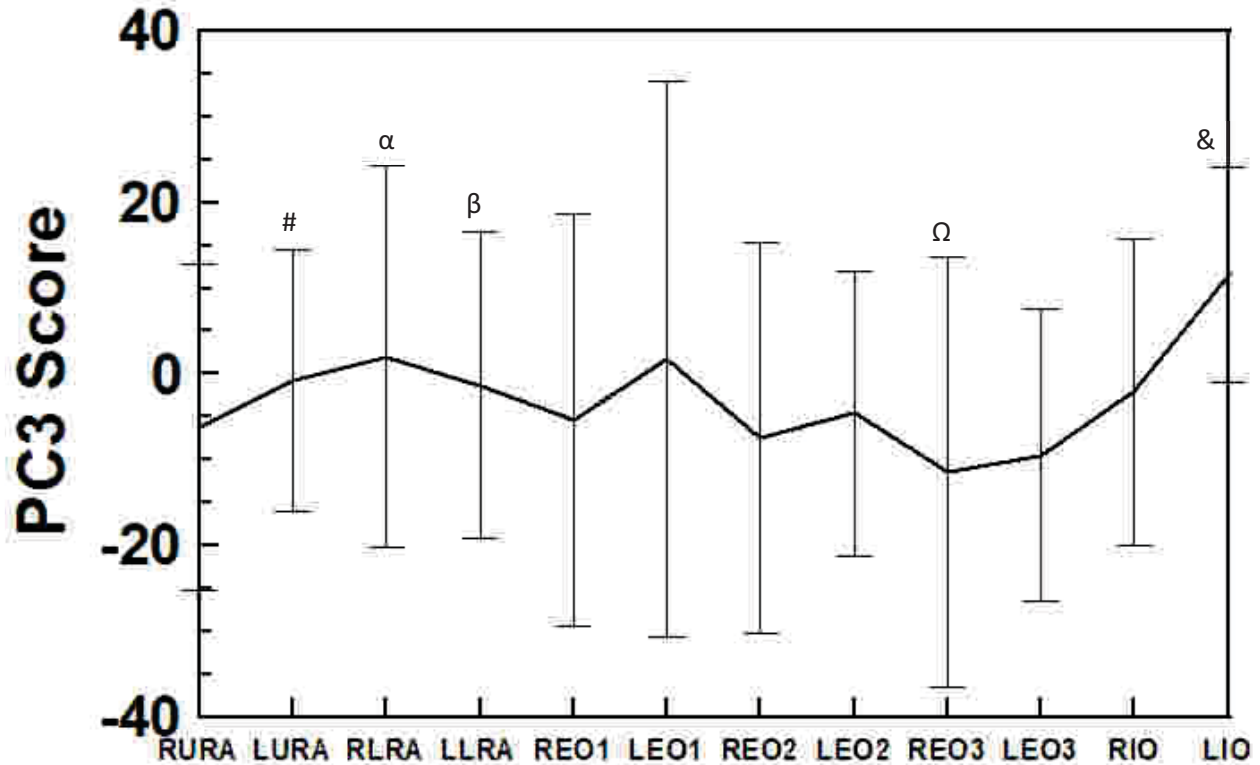
Figure 5.6: Mean PC3 scores and standard deviations for each of the 12 abdominal muscle sites. *LURA is higher than RURA, REO3 and RIO; αRLRA is higher than RIO; βLLRA is higher than RURA, LEO3 and RIO; ΩREO3 is lower than LEO1 and REO2; &LIO is higher than RURA, LURA, LLRA REO2, REO3, LEO3 and RIO.

5.3.2 Back extensor EMG waveform analysis

Mean EMG waveforms for all back muscles during the follicular phase and the ovulation phase are depicted in Figure 5.7. Subtle variation in muscle activation throughout the movement in the back extensor sites was confirmed by PCA, in that 96% of the variance was explained by two PCs (See Figure 5.8). Similar to the abdominals, PC1 captures the overall amplitude and shape characteristics of the waveforms (See Figure 5.8A). PC2 captured a pattern of varied amplitude in the signal over the course of the task with reduced amplitude at the beginning and end of the task compared to the middle (See Figure 5.8B). Representative high and low scoring curves for both PCs are also depicted in Figure 5.8C and 5.8D.
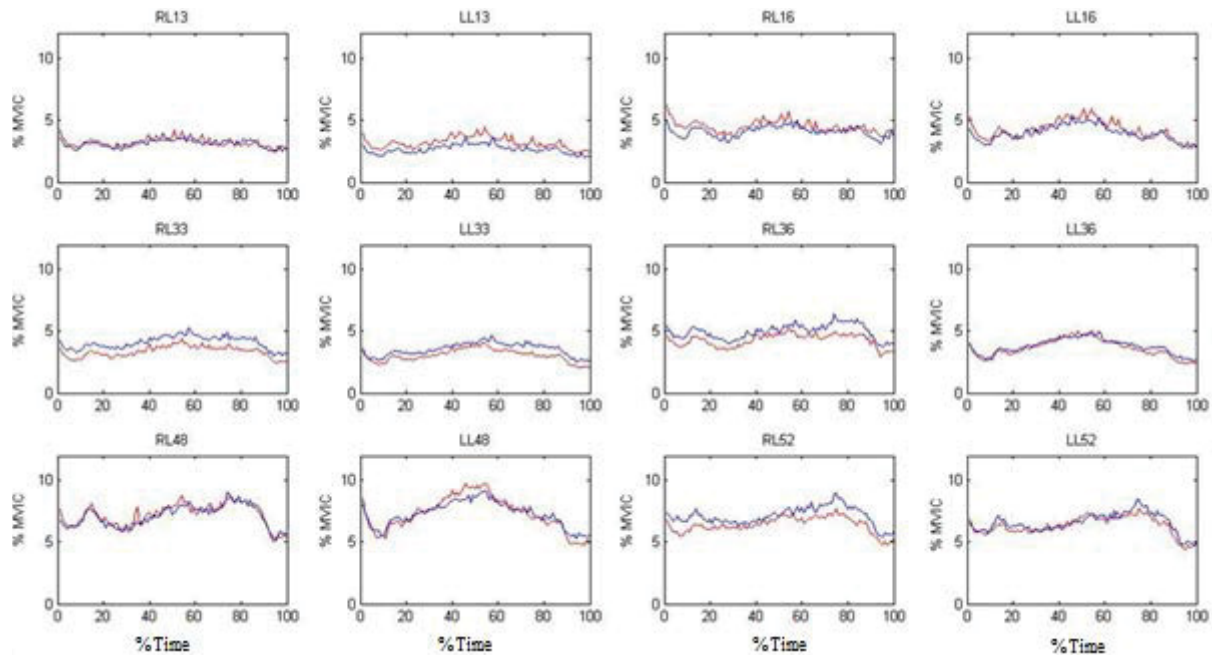
Figure 5.7: Ensemble average EMG waveforms for each of the 12 back muscle sites averaged across subjects for the follicular phase (red) and the ovulation phase (blue). RL13= right medial erector spinae (level L1); LL13= left medial erector spinae (level L1); RL16= right lateral erector spinae (level L1); LL16= left lateral erector spinae (level L1); RL33= right medial erector spinae (level L3); LL33= left medial erector spinae (level L3); RL36= right lateral erector spinae (level L3); LL36= left lateral erector spinae (level L3); RL48 = right quadratus lumborum; LL48 = left quadratus lumborum; RL52 = right multifidus; LL52 = left multifidus.
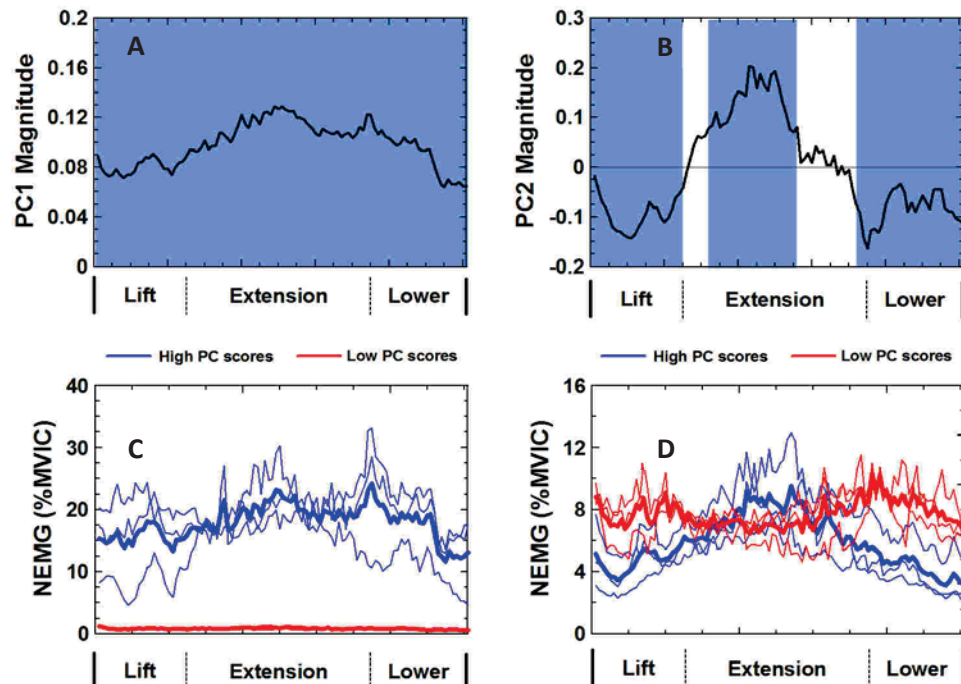
Figure 5.8: First two principal components for the back muscles. Blue shading on curves A and B indicate portions of the curve where the most variance in the data is described by the PC. A) PC1 which explains 91.4% of the waveform variance; B) PC2 which explains 5.0% of the variance; C-D) High and low scoring curves for PC1 and PC2 respectively with high scores shown in blue and low scores shown in red and the average high and average low scores bolded

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PC1 results indicate a significant main effect for muscles (p=0.000) and a phase and muscles interaction effect (p=0.085). Non-parametric post hoc tests revealed that LL13 has significantly higher PC1 scores during the follicular phase and RL33 has significantly higher PC1 scores during the ovulation phase. The significant between muscle site differences for each menstrual phase are shown in Figure 5.9.
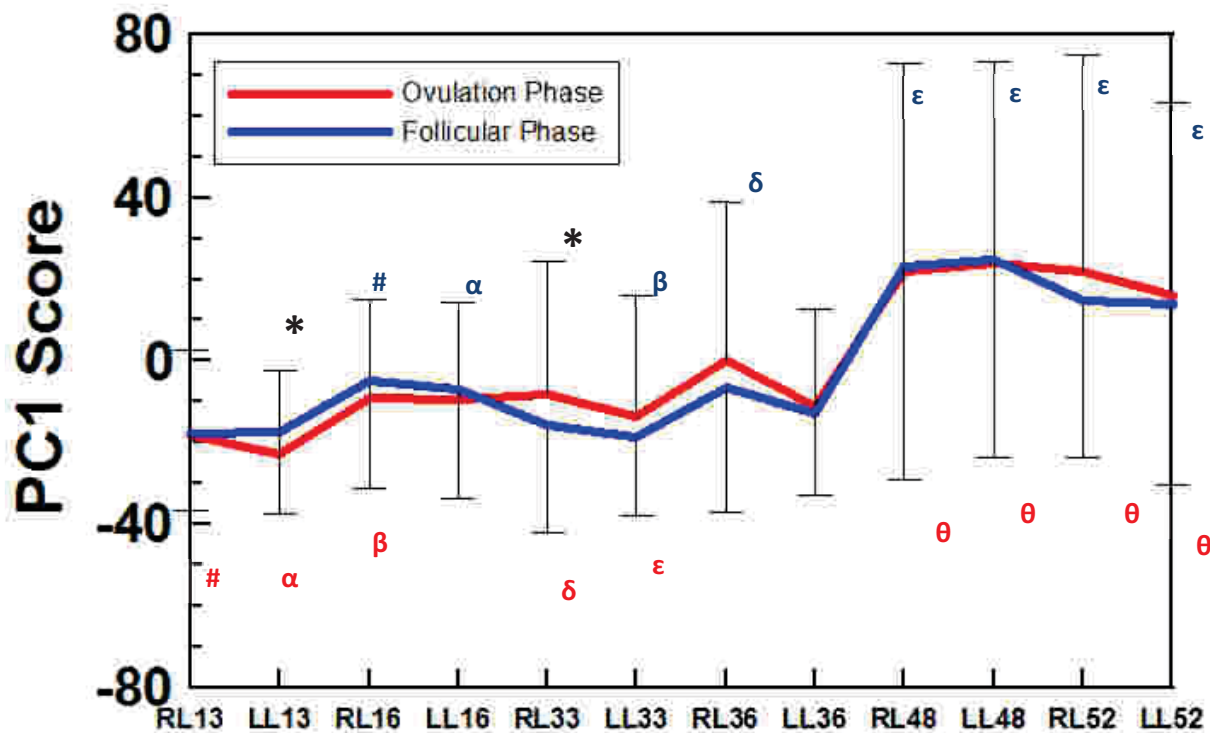
Figure 5.9: Mean PC1 scores and standard deviations for the 12 back muscle sites during the follicular phase and during the ovulation phase. * indicates a muscle site with a significant difference between phases.

For the Follicular phase: #RL16 is significantly different from RL13, LL13, RL33, LL33, RL48 and LL48. αLL16 is significantly different from RL13, RL33 and LL33. βLL33 is significantly different from LL36. δRL36 is significantly different from RL13 and RL33. εRL48, LL48, RL52 and LL52 are significantly different from RL13, LL13, LL16, RL33, LL33, RL36 and LL36.

For the Ovulation phase: : #RL13 is significantly different from RL16, LL16, RL33 and RL36. αLL13 is significantly different from RL16, LL16, RL33, LL33, RL36 and LL36. βRL16 is significantly different from RL52 and LL52. δRL33 is significantly different from LL33 and RL36. εLL33 is significantly different from RL36. θRL48, LL48, RL52 and LL52 are significantly different from RL13, LL13, LL16, RL33, LL33, RL36 and LL36

PC2 results indicate a significant main effect for phase (p=0.096) and muscles (p=0.000) and a phase and muscles interaction effect (p=0.044). Non-parametric post hoc tests revealed that RL13, RL33, RL36, LL36 and LL48 all have significantly higher PC2 scores during the follicular phase compared to during the ovulation phase. The significant between muscle site differences for each menstrual cycle phase are depicted in Figure 5.10.
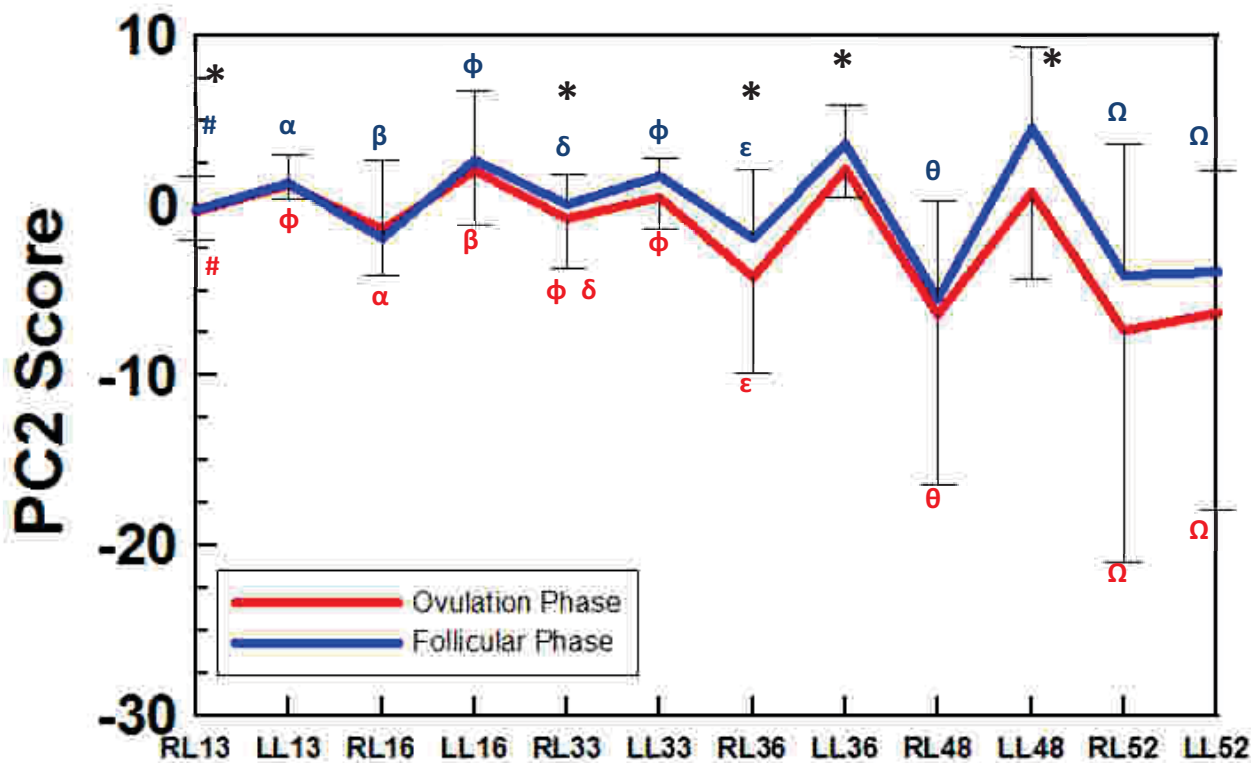
Figure 5.10: Mean PC2 scores and standard deviations for the 12 back muscle sites during the follicular phase and ovulation phase. * indicates a muscle site with a significant difference between phases.

For the Follicular phase: # RL13 is significantly different from LL13, LL16, LL33 and LL36. αLL13 is significantly different from RL16, LL33, RL36, LL36, RL48 and LL48. βRL16 is significantly different from LL16, RL33, LL33, LL36, RL48 and LL48. φLL16 and LL33 are significantly different from RL36, LL36 and RL48. δRL33 is significantly different from LL33, LL36 and RL48. εRL36 is significantly different from LL36, RL48 and LL48. θRL48 is significantly different from LL36 and LL48. ΩRL52 and LL52 are significantly different from LL13, RL16, LL16, RL33, LL33, LL36 and LL48.

For the Ovulation phase: #RL13 is significantly different from LL13, LL16, RL33, LL36 and RL52. αRL16 is significantly different from LL13, LL16, LL33 and LL36. βLL16 is significantly different from RL36 and RL52. φLL13, RL33 and LL33 are significantly different from RL36, LL36 and RL48. δRL33 is significantly different from LL33. εRL36 is significantly different from LL36 and LL48. θRL48 is significantly different from LL36 and LL48. ΩRL52 and LL52 are significantly different from LL13, RL33, LL33, LL36 and LL48.

5.4 Discussion

The purpose of this study was to evaluate the amplitude and temporal characteristics of EMG waveforms generated by the trunk musculature during the TST during the follicular and ovulation phases of the menstrual cycles. The results of this study generally found trends that were opposite to the study hypotheses. It was hypothesized that during the ovulation phase of the menstrual cycle women would show muscular activation patterns indicative of reduced stability; increased co-activation and more temporal asynchronies. However these patterns were seen more so in the follicular phase. The short term spike in estrogen levels seen during the ovulation phase was likely not enough to impose clear changes in neuromuscular control, as was also seen with changes in passive tissue properties (Zazulak et al., 2006).

The dominant amplitude and temporal features in the EMG waveforms were captured by three principal patterns in the abdominal PCA, which describe 90.2% of the variance in the waveforms (See Figure 5.4), and two principal patterns in the back extensor PCA, which describe 96% of the variance in the waveforms (See Figure 5.6). PC1 scores from both models captured amplitude differences (Hubley-Kozey et al., 2009) between waveforms, representing overall level of recruitment of various muscles. PC2 and PC3 from both models identified unique temporal characteristics within the data set.

5.4.1 Abdominal amplitude characteristics and co-activation

In the abdominal muscles PC1 scores were significantly higher during the follicular phase than during the ovulation phase. This represents a higher level of agonist co-activation during the follicular phase. In the trunk, co-activation has been shown to be beneficial as it increases joint stability and protects against damaging movements following perturbations (Gardner-Morse & Stokes, 1998). However, co-activation of trunk musculature can increase lumbar spine loads (Granata & Marras, 1995; Thelen et al., 1995) which could weaken spinal tissues, like ligaments,

discs and vertebrae, through repetitive loading that can eventually lead to injury. In Chapter 4 it was reported that women respond to the TST with higher levels of agonist co-activation than men. This was thought to be due to their reduced passive stiffness, which increases their need for active stiffness through co-activation to maintain lumbopelvic stability throughout the TST. The higher co-activation seen during the follicular phase could represent decreased passive stiffness during this phase of the menstrual cycle, since during the ovulation phase women are able to perform the task as well without relying on excess co-activation. Performance of the task was deemed to be equivalent during the two phases in that no differences existed in pelvic motion or time to complete the task (See Table 5.3). Though a decrease in passive stiffness during the follicular phase may be a possible explanation for the increased co-activation present, previous research has suggested that short term hormone fluctuations over the menstrual cycle are not enough to clearly affect passive tissue properties (Zazulak et al., 2006). Another possible explanation is a disconnect somewhere within the neuromuscular control system that results in a corrupted signal being sent that increases the demand for co-activation. Estrogen levels could still play a role in a possible neurophysiological effect due to the presence of alpha and beta estrogen receptors on skeletal muscles (Lemoine et al., 2003; Wilk et al., 2005).

These results are similar to recent findings showing that a pattern of increased agonist co-activation was characteristic of individuals who were not able to sustain lumbopelvic stability throughout the TST (Hubley-Kozey et al., 2010). Therefore increased co-activation may be a coping mechanism for a reduced level of passive stability or may reflect an altered neuromuscular control pathway in response to a lumbopelvic stability demand. Our results indicate that during the follicular phase women have increased demands for active stiffness

through co-activation of the abdominal muscles, demonstrating a difference in the neuromuscular control system at different points in the menstrual cycle.

5.4.2 Abdominal temporal characteristics and inter-muscle differences

A significantly higher PC2 score was reported for REO2 during the follicular phase compared to during the ovulation phase. This was the only abdominal muscle that showed a significant difference in PC2 score between phases. PC2 captures a pattern of responsiveness to the leg lifting task, representing an increase in activity during the leg extension portion of the task, relative to the initial portion. During the ovulation phase of their cycle, women began the task with an increased base level of activity in their REO2 site compared to during the follicular phase.

Differences in PC2 scores between muscles within a phase were also evaluated to assess temporal coordination between abdominal muscles. There was a higher number of inter-muscle differences during the ovulation phase (34) compared to the follicular phase (25). On first glance this might indicate that there is less temporal coordination of the abdominal muscles during the ovulation phase. However, when the various sites from the same muscle are grouped together (ie. rectus abdominus, external obliques and internal obliques) there are more differences within these groups during the follicular phase than during the ovulation phase. During the follicular phase, three rectus abdominus sites have significantly different PC2 scores than other rectus abdominus sites (ie. RURA has significantly higher scores than RLRA, LURA has significantly higher scores than RLRA, and RLRA has significantly lower scores than LLRA). No rectus abdominus sites have significantly different scores than other rectus abdominus sites during the ovulation phase. This may indicate that rectus abdominus recruitment is less coordinated during the follicular phase compared to during the ovulation phase. During the ovulation phase most of

the inter-muscle differences occur between different larger muscles rather than within them, which may represent a more selective response to the task. For example, during the ovulation phase women had lower PC2 scores in the external obliques than during the follicular phase, though only the REO2 differences reached significance. Not only were these scores relatively lower than those during the follicular phase, they were also lower relative to the rectus abdominus sites. During the follicular phase women had more similar PC2 scores between their external oblique sites and their rectus abdominus sites compared to during the ovulation phase. PC2 captures the relative increase in muscle activity in response to right leg extension. This manoeuver imposes a moment that acts to anteriorly tilt the pelvis making the vertical fiber orientation of the rectus abdominus muscle better suited to counteract it. The oblique muscle sites likely play more of a stabilizing role due to their more diagonal muscle fiber orientations. During the ovulation phase women are better able to selectively recruit their rectus abdominus sites over their external oblique sites during the leg extension portion of the motion. During the follicular phase the oblique muscle sites are recruited to similar levels during the leg extension portion of the movement as the rectus abdominus. This may indicate a less selective response, perhaps stemming from a need for more active stiffness throughout the task during the follicular phase.

The third PC generated for the abdominal EMG waveforms mainly captured increased activity at the beginning of the task, including a spike in activity at 15% of the motion corresponding to lifting the second leg off the bed. This feature in the data does not seem to be affected by menstrual cycle phase as there was no significant main phase effect or interaction effect between phase and muscle found.

5.4.3 Back extensor amplitude characteristics

In the back extensors, PC1 scores reveal an interaction between phase and muscle. When examining the upper erector spinae muscles (RL13, LL13, RL16 and LL16) a decrease in PC1 score is seen from the follicular phase to the ovulation phase, although only significantly at LL13. In the lower erector spinae (RL33, LL33, RL36 and LL36) there is an increase in PC1 scores from the follicular phase to the ovulation phase, though again this change is only significant in one muscle (RL33). However, the number of subjects included in this study was quite small (n=9). It is possible that with more subjects all eight changes might have reached significance. A power analysis, based on the means and standard deviations generated in this data set, revealed that with 24 subjects 5 of the 8 changes would have been significant at an alpha level of 0.1 (LL13, RL16, RL33, LL33 and RL36) (van Belle, 2008). Other studies have observed differential recruitment at different lumbar levels of the erector spinae (Jonsson, 1970; Vink et al., 1987). An explanation for this is that the load bearing abilities may be different at various lumbar levels. A detailed anatomical model demonstrated that the moment capability of the erector spinae at L1 is less than the moment at L3 (Bogduk, Macintosh, & Pearcy, 1992). As the back extensors are the antagonists during this exercise increased moments created by the back extensors would further challenge lumbopelvic stability during the task. During the ovulation phase, individuals shift towards a recruitment pattern that more preferentially recruits the lower segment of the erector spinae muscle, compared to the follicular phase. When the inter-muscle differences are examined, a similar trend is visible with upper erector spinae sites showing higher PC1 scores than lower erector spinae sites during the follicular phase (RL16 and LL16 sites are significantly higher than RL33 and LL33 sites) and lower erector spinae sites showing higher PC1 scores than upper erector spinae sites during the ovulation phase (RL33, LL33, RL36 and LL36 sites are significantly higher than LL13 and additionally RL33 and RL36

sites are significantly higher than RL13). Not enough information has been gathered to date to elucidate the mechanism for these differences in erector spinae muscle recruitment at different times in the menstrual cycle. These differences do further support the findings from the abdominal data, that there are differences in how the neuromuscular control system in women responds to the same task at the two measured points in the menstrual cycle.

5.4.4 Back extensor temporal characteristics

The second PC in the back extensors captured a pattern of varied amplitude in the signal over the course of the task with reduced amplitude at the beginning and end of the task compared to the middle. This pattern is a similar pattern to the PC2 pattern seen in the abdominal waveforms, representing a response to the perturbation coming from the extension of the right leg during the middle of the task. Five back extensor muscles (RL13, RL33, RL36, LL36 and LL48) show a significantly higher PC2 score during the follicular phase compared to during the ovulation phase. This indicates that during the follicular phase there is more co-activation between specific antagonists and agonists as the back extensor muscles display patterns that are more similar to those seen in the abdominal muscles. Specifically there is an increased response in the follicular phase during the leg extension portion of the task. Increased antagonist co-activation can increase the active stiffness of the spine, similarly to agonist co-activation. Antagonist co-activation, however, increases the load on the spine more than agonist co-activation does; not only is it imposing its own muscle forces, but the additional antagonist moments will create the necessity for increased agonist moments to balance the desired resultant force. The reason for this increase is not clear. Similar results were found in chapter 4 of this thesis, with women responding to the TST with a pattern referred to as 'bracing', demonstrating more active stiffness through agonist and antagonist co-activation than men. The results from this chapter expand on the results from chapter 4, indicating that this increased active stiffness

through co-activation is not consistent in women over the course of the menstrual cycle, and is greatest during the follicular phase.

5.4.4 Conclusion

In conclusion, the differences in PC scores seen between the two phases support the concept that muscle activation patterns are not consistent in response to a similar task demand across the menstrual cycle. The higher levels of both agonist and antagonist co-activation seen during the follicular phase indicate that there may be a deficit in stability or a neurophysiological effect on the neuromuscular control system during this phase of the menstrual cycle that leads to the greater demand for active stiffness through co-activation. There also seems to be more temporal coordination in muscle activation patterns during the ovulation phase compared to during the follicular phase. These results are opposite from the original study hypothesis which leads to the conclusion that fluctuating hormones over the course of the menstrual cycle impact neuromuscular control.

Chapter 6: Conclusion

6.1 Summary of study objectives

The purpose of this study was to investigate differences in neuromuscular control of the trunk during a dynamic stability exercise between men and women as well as within women at different time points in their menstrual cycles. The overall aim of the study was to gain insight to help clarify why women sustain musculoskeletal injuries at higher rates than men as well as why women seem to be more likely to sustain higher rates of musculoskeletal injuries at specific times in their menstrual cycles. The two main research objectives were:

1. To compare the relative amplitudes of EMG activity as well as the temporal EMG characteristics during a dynamic stability exercise, the TST, between men and women.

2. To examine the relative amplitudes of EMG activity as well as the temporal EMG characteristics during a dynamic stability exercise, the TST, performed at two different times in the menstrual cycle, once during the follicular phase, when estrogen levels are at their lowest point, and once during the ovulation phase, when estrogen levels are at their highest.

6.1.1 Summary of Chapter 4: Sex Differences in Neuromuscular Control Patterns during the Trunk Stability Test

The purpose of Chapter 4 was to examine sex differences in muscle activation patterns in response to the TST. It was hypothesized that women would show higher overall EMG amplitudes and more co-activation than men during the TST. It was also hypothesized that women would perform the task in a less coordinated fashion than men and would show this by having more temporal asynchronies between activation profiles of different muscles during the task.

Key findings from this chapter were:

- Women showed more agonist and antagonist co-activation than men. Because the higher co-activation in women was not accompanied by an altered task performance, as determined by motion data, it may be resulting as a consequence of the deficiency in the passive component of the female stabilizing system. Increased co-activation in women could potentially result in higher relative loads on the spine.

- Women require a greater relative increase in activity to maintain lumbopelvic stability when their leg is fully extended resulting in women recruiting their abdominals to higher levels preferentially at this time point.

- Women recruit more motor units as a percentage of maximum than men to complete the second leg lift during the TST, in a pattern similar to that seen in individuals who could not maintain lumbopelvic stability during the TST.

- Men respond to the task with more selective recruitment of back extensor muscle activation patterns while women respond with a less selective 'bracing' pattern.

- Women showed more inter-subject variability in their PC scores than men.

With respect to the overall study hypotheses the first hypothesis was confirmed. Women did respond to the TST task with more co-activation than men did. The second hypothesis was not confirmed as there were no significant interaction effects between sex and muscle for any of the PC scores that captured temporal characteristics. The results from this study indicate that women perform the TST with an overall higher relative activation and a 'bracing' strategy whereas men did not. The bracing strategy seen in women may be protective in the short term, but over time the extra loading on the spine from excess muscle forces may overload the tissues in the female spine. However, without actually calculating the spinal loads it can only be

concluded that the neuromuscular activation patterns were different between the sexes which suggests differences in spinal loading. More detailed biomechanical analysis would be needed to make concrete conclusions. These suggested loading differences may help explain why women experience more low back injuries compared to men. Results also indicate that women as a population respond to the task with more varied activation patterns than men. It has been suggested that this might be due variability caused within women by their menstrual cycles.

6.1.2 Summary of Chapter 5: Effect of the Female Menstrual Cycle on Neuromuscular Control of Trunk Musculature.

The purpose of Chapter 5 was to examine if changes occurred in how women controlled their trunk musculature during a dynamic stability exercise over the course of their menstrual cycle. It was hypothesized that during the ovulation phase, while performing the TST task, women would have increased co-activation and increased temporal asynchronies between activation patterns of different muscles consistent with patterns reported for a group that was unable to maintain lumbopelvic stability (Hubley-Kozey et al., 2010).

Key findings from this chapter were:

- Women demonstrated higher levels of agonist co-activation during the follicular phase. This may result as a consequence of a deficiency in another component of the stabilizing system or an altered signal generated somewhere along the neuromuscular control pathway. Increased co-activation may result in higher loads on the spine during the follicular phase of the menstrual cycle.
- There is also more antagonist co-activation, particularly during the leg extension phase of the motion, during the follicular phase compared to the ovulation phase.

- More temporal asynchronies between muscle sites within the same abdominal muscle were identified during the follicular phase. This indicates less temporal coordination between muscles during the follicular phase.

- During the ovulation phase, compared to the follicular phase, individuals selectively recruit the rectus abdominus muscles relative to the obliques during the leg extension portion of the task.

- Women selectively recruit lumbar levels of their erector spinae differently during the two different phases of the menstrual cycle.

The results of the menstrual cycle phase on neuromuscular control of the trunk are opposite from what was hypothesized. It was hypothesized that women would have more co-activation and more inter-muscle asynchronies during the ovulation phase. Instead the results of this study demonstrate that during the ovulation phase women respond to the demands of the TST with a more coordinated response and with less agonist and antagonist co-activation compared to during the follicular phase. Other neuromuscular control differences, such as erector spinae recruitment and temporal pattern asynchronies, were seen between the two different phases of the menstrual cycle. Overall these findings indicate fundamental differences in the neuromuscular control system over the course of the menstrual cycle. Potential reasons for these differences which were suggested include altered passive stability in the spine at specific times in the cycle or altered signals generated along the neuromuscular control pathway. However, as this work is relatively preliminary, reasons for these differences can only be speculated on.

6.2 Clinical implications

It was found that women perform the TST task with a less coordinated and less selective response than men, resorting to more muscular co-activation to generate the necessary stiffness

to perform the task with minimal pelvis motion. This increased active stiffness due to co-activation in women is likely necessary for them to maintain the lumbar spine in a neutral posture, reducing the risk of injury due to movement outside of the physiological range of motion of the tissues (Panjabi, 1992). It is not just during the TST task that women respond with increased trunk muscular co-activation around the spine compared to men. Other authors have demonstrated similar findings during static extension efforts (Granata & Orishimo, 2001; Granata et al., 2001), whole body tilt (Anders et al., 2007) and lifting efforts (Hubley-Kozey et al., 2011; Marras et al., 2003). Women are therefore imposing higher relative levels of loading on their spines compared to men during a wide range of tasks, which could accumulate over time. According to McGill, low back injuries are frequently the result of long term loading, which gradually, but progressively, reduces the tolerance of the tissues to withstand perturbations (McGill, 1998). It is therefore possible that women sustain low back injuries at rates higher than men due to the long term exposure to increased relative loading on the female spine due to muscular activation patterns of increased co-activation that women adopt for most tasks.

Women also seem to have altered neuromuscular control of the trunk musculature during different phases of the menstrual cycle. During the follicular phase women responded to the task with a less coordinated muscular response as well as with more agonist and antagonist co-activation. Though some studies found the rate of ACL injuries to be highest at ovulation (Adachi et al., 2008; Wojtys et al., 2002; Wojtys et al., 1998), others demonstrated an increased risk during the whole pre-ovulatory phase (Hewett et al., 2007; Shultz et al., 2010), which includes the follicular phase. Therefore, our study results are in accordance with some of the ACL injury susceptibility literature. However, no research has been done looking at low back injury susceptibility over the course of the menstrual cycle and it cannot be assumed that the low

back injury susceptibility results would mirror those for ACL injury susceptibility over the menstrual cycle.

6.3 Future research

One of the main limitations of the menstrual cycle project work was a small sample size. Due to strict inclusion criteria, our menstrual cycle sample only included nine women who were each tested at two different occasions and no control group. Future research should include larger sample sizes, in the anticipation of obtaining statistical significance.

A gap in this area of research is an investigation into the injury susceptibility of the back relative to different phases of the menstrual cycle. ACL injuries are most common during the pre-ovulatory phase but it cannot be assumed the same is true for low back injuries.

This was one of the first studies to look at the effect of the menstrual cycle on neuromuscular activation patterns of the trunk musculature, thus providing a foundation for further investigation in this area. This study revealed that neuromuscular response patterns are altered over the course of the menstrual cycle. One possible link to reduced neuromuscular control during the follicular phase could be reports of premenstrual syndrome (PMS). PMS is characterized by cyclical physical and mood disturbances that occur at the end of the luteal phase during the menstrual cycle and disappear following the first few days of the follicular phase (Sundström et al., 1998). Low back pain is one of the most frequently reported PMS symptoms (Pinar, Colak, & Oksuz, 2011). The causes of PMS are unclear with hypothesized causes ranging from hormonal changes and neurotransmitter concentrations to dietary and lifestyle factors (Clare, 1985; Schellenberg, 2001). Friden et al., found that women who experienced PMS symptoms had greater amounts of postural sway than those who did not (Friden et al., 2005). Therefore, it is possible that there may be a link between PMS and deficiencies in neuromuscular

control that could potentially lead to low back pain. Future studies should include an evaluation of severity of PMS symptoms in each participant to investigate this possibility. Future studies could also plan to test individuals at a time in their cycle when PMS symptoms would be the most severe.

6.4 Conclusions

The purpose of this study was to evaluate the effects of sex and menstrual cycle phase on neuromuscular control of the trunk during a dynamic stability exercise. The overall aim of the study was to gain insight into why women sustain musculoskeletal injuries at higher rates than men as well as at higher rates at specific times in their menstrual cycles.

It was found that women perform the TST with an overall 'bracing' strategy, including more muscular co-activation around the spine, whereas men had more differences between muscles, perhaps indicative of an ability to more selectively recruit muscles as needed. The muscle recruitment strategy seen in women would tend to increase spine loading due to the increased muscle forces. Women's neuromuscular activation patterns were found to be altered over the course of the menstrual cycle, with women responding to the TST task with more temporal asynchronies and more co-activation during the follicular phase relative to during the ovulation phase. Results of this study may assist in understanding why women injure their backs at rates higher than men and suggests that women may be at higher risk of low back injury during the follicular phase of their menstrual cycle.

References

Abt, J. P., Sell, T. C., Laudner, K. G., McCrory, J. L., Loucks, T. L., Berga, S. L., et al. (2007). Neuromuscular and biomechanical characteristics do not vary across the menstrual cycle. *Knee Surgery, Sports Traumatology, Arthroscopy, 15*(7), 901-907.

Adachi, N., Nawata, K., Maeta, M., & Kurozawa, Y. (2008). Relationship of the menstrual cycle phase to anterior cruciate ligament injuries in teenaged female athletes. *Archives of Orthopaedic and Trauma Surgery, 128*(5), 473-478.

Akuthota, V., & Nadler, S. F. (2004). Core strengthening. *Archives of Physical Medicine and Rehabilitation, 85*(3 Suppl 1), S86-92.

Allison, G. T., Godfrey, P., & Robinson, G. (1998). EMG signal amplitude assessment during abdominal bracing and hollowing. *Journal of Electromyography and Kinesiology, 8*(1), 51-57.

Anders, C., Brose, G., Hofmann, G. O., & Scholle, H. C. (2007). Gender specific activation patterns of trunk muscles during whole body tilt. *European Journal of Applied Physiology, 101*, 195-205.

Arendt, E., Agel, J., & Dick, R. (1999). Anterior cruciate ligament injury patterns amoung collegiate men and women. *Journal of Athletic Training, 34*, 86-92.

Arendt, E., & Dick, R. (1995). Knee injury patterns among men and women in collegiate basketball and soccer: NCAA data and review of literature. *The American Journal of Sports Medicine, 23*, 694-701.

Arnold, C., Van Bell, C., Rogers, V., & Cooney, T. (2002). The relationship between serum relaxin and knee joint laxity in female athletes. *Orthopedics, 25*, 669-673.

Belanger, M. J., Moore, D. C., Crisco, J. J., Fadale, P. D., Hulstyn, M. J., & Ehrlich, M. G. (2004). Knee laxity does not vary with the menstrual cycle, before or after exercise. *The American Journal of Sports Medicine, 32*(5), 1150-1157.

Bell, D. R., Blackburn, J. T., Norcorss, M. F., Ondrak, K. S., Hudson, J. D., Hackney, A. C., et al. (2011). Estrogen and muscle stiffness have a negative relationship in females. *Knee Surgery, Sports Traumatology, Arthroscopy, 20*(2), 361-367.

Bell, D. G., & Jacobs, I. (1986). Electro-mechanical response times and rate of force development in males and females. *Medicine and Science in Sports and Exercise, 18*(1), 31-36.

Bergmark, A. (1989). Stability of the lumbar spine. A study in mechanical engineering. *Acta Orthopaedica Scandinavica.Supplementum, 230*, 1-54.

Beynnon, B. D., Bernstein, I. M., Belisle, A., Brattbakk, B., Devanny, P., Risinger, R., et al. (2005). The effect of estradiol and progesterone on knee and ankle laxity. *The American Journal of Sports Medicine, 33*, 1298-1304.

Bogduk, N., Macintosh, J. E., & Pearcy, M. J. (1992). A universal model of the lumbar back muscles in the upright position. *Spine, 17*, 897-913.

Bouisset, S., & Goubel, F. (1971). Interdependence of relations between integrated EMG and diverse biomechanical quantities in normal voluntary movements. *Activitas Nervosa Superior, 13*(1), 23-31.

Bowen, V., & Cassidy, J. D. (1981). Macroscopic and microscopic anatomy of the sacroiliac joint from embryonic life until the eighth decade. *Spine, 6*(6), 620-628.

Brantley, P. J., Dietz, L. S., McKnight, G. T., Jones, G. N., & Tulley, R. (1988). Convergence between the daily stress inventory and endocrine measures of stress. *Journal of Consulting and Clinical Psychology, 56*(4), 549-551.

Brantley, P. J., Waggoner, C. D., Jones, G. N., & Rappaport, N. B. (1987). A daily stress inventory: Development, reliability, and validity. *Journal of Behavioral Medicine, 10*(1), 61-74.

Brown, S. H., Vera-Garcia, F. J., & McGill, S. M. (2006). Effects of abdominal muscle coactivation on the externally preloaded trunk: Variations in motor control and its effect on spine stability. *Spine, 31*(13), E387-93.

Bryant, A. L., Clark, R. A., Bartold, S., Murphy, A., Bennell, K. L., Hohmann, E., et al. (2008). Effects of estrogen on mechanical behaviour of the achilles tendon in vivo. *Journal of Applied Physiology, 105*, 1035-1043.

Buchanan, T. S., Lloyd, D. G., Manal, K., & Besier, T. F. (2004). Neuromusculoskeletal modeling: Estimation of muscle forces and joint moments and movements from measurements of neural command. *Journal of Applied Biomechanics, 20*(4), 367-395.

Buchanan, T. S., Lloyd, D. G., Manal, K., & Besier, T. F. (2005). Estimation of muscle forces and joint moments using a forward-inverse dynamics model. *Medicine and Science in Sports and Exercise, 37*(11), 1911-1916.

Burden, A., & Bartlett, R. (1999). Normalisation of EMG amplitude: An evaluation and comparison of old and new methods. *Medical Engineering & Physics, 21*(4), 247-257.

Butler, D. L., Noyes, F. R., & Grood, E. S. (1980). Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study. *The Journal of Bone and Joint Surgery, 62*(2), 259-270.

Butler, H. L. (2007). *Electromyographic analysis of trunk muscle patterns during functional lifting in a healthy population.* Unpublished PhD, Dalhousie University, Halifax, NS.

Butler, H. L., Hubley-Kozey, C. L., & Kozey, J. W. (2009a). Activation amplitude patterns do not change for back muscles but are altered for abdominal muscles between dominant and non-dominant hands during one-handed lifts. *European Journal of Applied Physiology, 106*(1), 95-104.

Butler, H. L., Hubley-Kozey, C. L., & Kozey, J. W. (2009b). Electromyographic assessment of trunk muscle activation amplitudes during a simulated lifting task using pattern recognition techniques. *Journal of Electromyography and Kinesiology, 19*(6), 505-512.

Butler, H. L., Hubley-Kozey, C. L., & Kozey, J. W. (2010). Characterisation of trunk muscle activation amplitude patterns during a simulated checkstand operation with continuously changing flexor and lateral moment demands. *Ergonomics, 53*(5), 685-695.

Butler, H. L., Newell, R., Hubley-Kozey, C. L., & Kozey, J. W. (2009). The interpretation of abdominal wall muscle recruitment strategies change when the electrocardiogram (ECG) is removed from the electromyogram (EMG). *Journal of Electromyography and Kinesiology, 19*(2), e102-13.

Cason, C. L., & Grissom, N. L. (1997). Ameliorating adults' acute pain during phlebotomy with a distraction intervention. *Applied Nursing Research, 10*(4), 168-173.

Chiazze, L., Brayer, F. T., Macisco, J. J., Parker, M. P., & Duffy, B. J. (1968). The length and variability of the human menstrual cycle. *The Journal of the American Medical Association, 203*, 377.

Cholewicki, J., Juluru, K., Radebold, A., Panjabi, M. M., & McGill, S. M. (1999). Lumbar spine stability can be augmented with an abdominal belt and/or increased intra-abdominal pressure. *European Spine Journal, 8*, 388-395.

Cholewicki, J., & McGill, S. M. (1996). Mechanical stability of the in vivo lumbar spine: Implications for injury and chronic low back pain. *Clinical Biomechanics, 11*(1), 1-15.

Cholewicki, J., Panjabi, M., & Khachatryan, A. (1998). Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. *Spine, 22*(19), 2207-2212.

Cholewicki, J., Panjabi, M. M., & Khachatryan, A. (1997). Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. *Spine, 22*(19), 2207-2212.

Cholewicki, J., & VanVliet, J. J.,4th. (2002). Relative contribution of trunk muscles to the stability of the lumbar spine during isometric exertions. *Clinical Biomechanics, 17*(2), 99-105.

Clare, A. W. (1985). Premenstrual syndrome: Single or multiple causes? *Canadian Journal of Psychiatry, 30*(7), 474-482.

Clarke-Davidson, K. L., & Hubley-Kozey, C. L. (2005). Trunk muscle responses to demands of an exercise progression to improve dynamic spinal stability. *Archives of Physical Medicine and Rehabilitation, 86*(2), 216-223.

Cohen, S. P., Argoff, C. E., & Carragee, E. J. (2008). Management of low back pain. *British Medical Journal, 337*, a2718.

Cole, L. A., Ladner, D. G., & Byrn, F. W. (2009). The normal variabilities of the menstrual cycle. *Fertility and Sterility, 91*(2), 522-527.

Collett, M. E., Wertenberger, G. E., & Fiske, V. M. (1954). The effect of age upon the pattern of the menstrual cycle. *Fertility and Sterility, 5*, 437.

Crisco, J. J.,3rd, & Panjabi, M. M. (1991). The intersegmental and multisegmental muscles of the lumbar spine. A biomechanical model comparing lateral stabilizing potential. *Spine, 16*(7), 793-799.

Da Silva, R. A., Lariviere, C., Arsenault, A. B., Nadeau, S., & Plamondon, A. (2009). Pelvic stabilization and semisitting position increase the specificity of back exercises. *Medicine and Science in Sports and Exercise, 41*(2), 435-443.

Daffertshofer, A., Lamoth, C. J., Meijer, O. G., & Beek, P. J. (2004). PCA in studying coordination and variability: A tutorial. *Clinical Biomechanics, 19*(4), 415-428.

Dankaerts, W., O'Sullivan, P. B., Burnett, A. F., Straker, L. M., & Danneels, L. A. (2004). Reliability of EMG measurements for trunk muscles during maximal and sub-maximal voluntary isometric contractions in healthy controls and CLBP patients. *Journal of Electromyography and Kinesiology, 14*(3), 333-342.

Danneels, L. A., Coorevits, P. L., Cools, A. M., Vanderstraeten, G. G., Cambier, D. C., Witvrouw, E. E., et al. (2002). Differences in electromyographic activity in the multifidus muscle and the iliocostalis lumborum between healthy subjects and patients with sub-acute and chronic low back pain. *European Spine Journal, 11*(1), 13-19.

Danneels, L. A., Cagnie, B. J., Cools, A. M., Vanderstraeten, G. G., Cambier, D. C., Witvrouw, E. E., et al. (2001). Intra-operator and inter-operator reliability of surface electromyography in the clinical evaluation of back muscles. *Manual Therapy, 6*(3), 145-153.

De Luca, C. J. (1997). The use of surface electromyography in biomechanics. *Journal of Applied Biomechanics, 13*, 135-163.

De Luca, C. J., & Merletti, R. (1988). Surface myoelectric signal cross-talk among muscles of the leg. *Electroencephalography and Clinical Neurophysiology, 69*(6), 568-575.

Dedrick, G. S., Sizer, P. S., Merkle, J. N., Hounshell, T. R., Robert-McComb, J. J., Sawyer, S. F., et al. (2006). Effect of sex hormones on neuromuscular control patterns during landing. *Journal of Electromyography and Kinesiology, 18*(1), 68-78.

Deie, M., Sakamaki, Y., Sumen, Y., Urabe, Y., & Ikuta, Y. (2002). Anterior knee laxity in young women varies with their menstrual cycle. *International Orthopaedics, 26*, 154-156.

Dyer, R. F., Sodek, J., & Heersche, J. N. W. (1980). The effect of 17beta-estradiol on collagen and non-collagenous protein synthesis in the uterus and some periodontal *Endocrinology, 107*, 1014-1021.

Dzugan, S., & Scipione, A. (2006). Progesterone misconceptions. *Life Extension.*

Edman, K. A. (1978). Maximum velocity of shortening in relation to sarcomere length and degree of activation of frog muscle fibres proceedings. *The Journal of Physiology, 278*, 9P-10P.

Edman, K. A. (1979). The velocity of unloaded shortening and its relation to sarcomere length and isometric force in vertebrate muscle fibres. *The Journal of Physiology, 291*, 143-159.

Eiling, E., Bryant, A. L., Peterson, W., Murphy, A., & Hohmann, E. (2007). Effects of menstrual-cycle hormone fluctuations on musculotendinous stiffness and knee joint laxity. *Knee Surgery, Sports Traumatology, Arthroscopy, 15*, 126-132.

Farina, D., Merletti, R., & Stegeman, D. F. (2004). Biophysics of the generation of EMG signals. In R. Merletti, & P. Parker (Eds.), *Electromyography - physiology, engineering, and noninvasive applications*, John Wiley & Sons.

Farina, D., Merletti, R., Indino, B., & Graven-Nielsen, T. (2004). Surface EMG crosstalk evaluated from experimental recordings and simulated signals. reflections on crosstalk interpretation, quantification and reduction. *Methods of Information in Medicine, 43*(1), 30-35.

Fehring, R. J., Schneider, M., & Raviele, K. (2006). Variability in the phases of the menstrual cycle. *Journal of Obstetric, Gynecologic & Neonatal Nursing, 36*(3), 376-384.

Ferreira, P. H., Ferreira, M. L., & Hodges, P. W. (2004). Changes in recruitment of the abdominal muscles in people with low back pain: Ultrasound measurement of muscle activity. *Spine, 29*(22), 2560-2566.

Feuerstein, M., Berkowitz, S. M., & Peck, C. A. (1997). Musculoskeletal-related disability in the U.S. army personnel: Prevalence, gender and military occupational specialties. *Journal of Occupational and Environmental Medicine, 39*, 68-78.

Fischer, G. M. (1973). Comparison of collagen dynamics in different tissues under the influence of estradiol. *Endocrinology, 93*(5), 1216-1218.

Ford, K. R., Myer, G. D., & Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Medicine and Science in Sports and Exercise, 35*(10), 1745-1750.

Ford, K. R., Myer, G. D., Toms, H. E., & Hewett, T. E. (2005). Gender differences in the kinematics of unanticipated cutting in young athletes. *Medicine and Science in Sports and Exercise, 37*(1), 124-129.

Freburger, J. K., Holmes, G. M., Agans, R. P., Jackman, A. M., Darter, J. D., Wallace, A. S., et al. (2009). The rising prevalence of chronic low back pain. *Archives of Internal Medicine, 169*(3), 251-258.

Friden, C., Hirschberg, A. L., Saartok, T., Backstrom, T., Leanderson, J., & Renstrom, P. (2003). The influence of premenstrual symptoms on postural balance and kinesthesia during the menstrual cycle. *Gynecological Endocrinology, 17*(6), 433-439.

Friden, C., Ramsey, D. K., Backstrom, T., Benoit, D. L., Saartok, T., & Hirschberg, A. L. (2005). Altered postural control during the luteal phase in women with premenstrual symptoms. *Neuroendocrinology, 81*(3), 150-157.

Fuglevand, A. J., Winter, D. A., Patla, A. E., & Stashuk, D. (1992). Detection of motor unit action potentials with surface electrodes: Influence of electrode size and spacing. *Biological Cybernetics, 67*(2), 143-153.

Gardner-Morse, M. G., & Stokes, I. A. (1998). The effects of abdominal muscle coactivation on lumbar spine stability. *Spine, 23*(1), 86-91.

Gilleard, W. L., & Brown, J. M. (1994). An electromyographic validation of an abdominal muscle test. *Archives of Physical Medicine and Rehabilitation, 75*(9), 1002-1007.

Goldfuss, A. J., Morehouse, C. A., & LeVeau, B. F. (1973). Effect of muscular tension on knee stability. *Medicine and Science in Sports, 5*(4), 267-271.

Gordon, A. M., Huxley, A. F., & Julian, F. J. (1966). The variation in isometric tension with sarcomere length in vertebrate muscle fibres. *The Journal of Physiology, 184*(1), 170-192.

Grabiner, M. D., Donovan, S., Bareither, M. L., Marone, J. R., Hamstra-Wright, K., Gatts, S., et al. (2008). Trunk kinematics and fall risk of older adults: Translating biomechanical results to the clinic. *Journal of Electromyography and Kinesiology, 18*(2), 197-204.

Granata, K. P., & Marras, W. S. (1995). The influence of trunk muscle coactivity on dynamic spinal loads. *Spine, 20*(8), 913-919.

Granata, K. P., & Orishimo, K. F. (2001). Response of trunk muscle coactivation to changes in spinal stability. *Journal of Biomechanics, 34*(9), 1117-1123.

Granata, K. P., Orishimo, K. F., & Sanford, A. H. (2001). Trunk muscle coactivation in preparation for sudden load. *Journal of Electromyography and Kinesiology, 11*(4), 247-254.

Granata, K. P., Padua, D. A., & Wilson, S. E. (2002). Gender differences in active musculoskeletal stiffness. Part II. Quantification of leg stiffness during functional hopping tasks. *Journal of Electromyography and Kinesiology, 12*(2), 127-135.

Granata, K. P., Wilson, S. E., & Padua, D. A. (2002). Gender differences in active musculoskeletal stiffness. Part I. Quantification in controlled measurements of knee joint dynamics. *Journal of Electromyography and Kinesiology, 12*(2), 119-126.

Gudgeon, K., Leader, L., & Howard, B. (1990). Evaluation of the accuracy of the home ovulation detection kit, clearplan, at predicting ovulation. *The Medical Journal of Australia, 152*(7), 344-349.

Hackney, A. C., & Viru, A. (2008). Research methodology: Endocrinologic measurements in exercise science and sports medicine. *Journal of Athletic Training, 43*(6), 631-639.

Hagood, S., Solomonow, M., Baratta, R., Zhou, B. H., & D'Ambrosia, R. (1990). The effect of joint velocity on the contribution of the antagonist musculature to knee stiffness and laxity. *The American Journal of Sports Medicine, 18*(2), 182-187.

Hakkinen, K. (1991). Force production characteristics of leg extensor, trunk flexor and extensor muscles in male and female basketball players. *The Journal of Sports Medicine and Physical Fitness, 31*(3), 325-331.

Harmon, K. G., & Ireland, M. L. (2000). Gender differences in noncontact anterior cruciate ligament injuries. *Clinics in Sports Medicine, 19*(2), 287-302.

Heckathorne, C. W., & Childress, D. S. (1981). Relationships of the surface electromyogram to the force, length, velocity, and contraction rate of the cineplastic human biceps. *American Journal of Physical Medicine, 60*(1), 1-19.

Heitz, N. A., Eisenman, P. A., Beck, C. L., & Walker, J. A. (1999). Hormonal changes throughout the menstrual cycle and increased anterior cruciate ligament laxity in females. *Journal of Athletic Training, 34*(2), 144-149.

Hemingway, M. A., Biedermann, H. J., & Inglis, J. (1995). Electromyographic recordings of paraspinal muscles: Variations related to subcutaneous tissue thickness. *Biofeedback and Self-Regulation, 20*(1), 39-49.

Hewett, T. E., & Myer, G. D. (2011). The mechanistic connection between the trunk, hip, knee, and anterior cruciate ligament injury. *Exercise and Sport Sciences Reviews, 39*(4), 161-166.

Hewett, T. E., Myer, G. D., & Ford, K. R. (2004). Decrease in neuromuscular control about the knee with maturation in female athletes. *The Journal of Bone and Joint Surgery.American Volume, 86-A*(8), 1601-1608.

Hewett, T. E., Myer, G. D., & Ford, K. R. (2005). Reducing knee and anterior cruciate ligament injuries among female athletes: A systematic review of neuromuscular training interventions. *The Journal of Knee Surgery, 18*(1), 82-88.

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Jr, Colosimo, A. J., McLean, S. G., et al. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: A prospective study. *The American Journal of Sports Medicine, 33*(4), 492-501.

Hewett, T. E., Zazulak, B. T., & Myer, G. D. (2007). Effects of the menstrual cycle on anterior cruciate ligament injury risk: A systematic review. *The American Journal of Sports Medicine, 35*(4), 659-668.

Hirokawa, S., Solomonow, M., Luo, Z., Lu, Y., & D'Ambrosia, R. (1991). Muscular co-contraction and control of knee stability. *Journal of Electromyography and Kinesiology, 1*(3), 199-208.

Hodges, P. W. (1999). Is there a role for transversus abdominis in lumbo-pelvic stability? *Manual Therapy, 4*(2), 74-86.

Hodges, P. W., & Richardson, C. A. (1996). Inefficient muscular stabilization of the lumbar spine associated with low back pain. A motor control evaluation of transversus abdominis. *Spine, 21*(22), 2640-2650.

Hodges, P. W., & Richardson, C. A. (1997). Contraction of the abdominal muscles associated with movement of the lower limb. *Physical Therapy, 77*(2), 132-42; discussion 142-4.

Hodges, P. W., & Richardson, C. A. (1998). Delayed postural contraction of transversus abdominis in low back pain associated with movement of the lower limb. *Journal of Spinal Disorders, 11*(1), 46-56.

Hodges, P. W., & Richardson, C. A. (1999). Altered trunk muscle recruitment in people with low back pain with upper limb movement at different speeds. *Archives of Physical Medicine and Rehabilitation, 80*(9), 1005-1012.

Hopkins, W. G. (2000). Measures of reliability in sports medicine and science. *Sports Medicine, 30*(1), 1-15.

Hsu, W. H., Fisk, J. A., Yamamoto, Y., Debski, R. E., & Woo, S. L. (2006). Differences in torsional joint stiffness of the knee between genders: A human cadaveric study. *The American Journal of Sports Medicine, 34*(5), 765-770.

Hubley-Kozey, C. L., Butler, H. L., & Kozey, J. W. (2011). Activation amplitude and temporal synchrony among back extensor and abdominal muscles during a controlled transfer task: Comparison of men and women. *The Journal of Human Movement Science, In Press*

Hubley-Kozey, C. L., Deluzio, K. J., Landry, S. C., McNutt, J. S., & Stanish, W. D. (2006). Neuromuscular alterations during walking in persons with moderate knee osteoarthritis. *Journal of Electromyography and Kinesiology, 16*(4), 365-378.

Hubley-Kozey, C. L., Hanada, E. Y., Gordon, S., Kozey, J., & McKeon, M. (2009). Differences in abdominal muscle activation patterns of younger and older adults performing an asymmetric leg-loading task. *PM & R: The Journal of Injury, Function, and Rehabilitation, 1*(11), 1004-1013.

Hubley-Kozey, C. L., Hatfield, G. L., & Davidson, K. C. (2010). Temporal coactivation of abdominal muscles during dynamic stability exercises. *Journal of Strength and Conditioning Research, 24*(5), 1246-1255.

Hubley-Kozey, C. L., & Vezina, M. J. (2002). Differentiating temporal electromyographic waveforms between those with chronic low back pain and healthy controls. *Clinical Biomechanics, 17*(9-10), 621-629.

Hurt, C. P., Rosenblatt, N., Crenshaw, J. R., & Grabiner, M. D. (2010). Variation in trunk kinematics influences variation in step width during treadmill walking by older and younger adults. *Gait & Posture, 31*(4), 461-464.

Huston, L. J., & Wojtys, E. M. (1996). Neuromuscular performance characteristics in elite female athletes. *The American Journal of Sports Medicine, 24*(4), 427-436.

Ivanenko, Y. P., Poppele, R. E., & Lacquaniti, F. (2004). Five basic muscle activation patterns account for muscle activity during human locomotion. *The Journal of Physiology, 556*, 267-282.

Jackson, J. E. (1991). *User's guide to principal components*. New York: John Wiley & Sons.

Janse de Jonge, X. A., Boot, C. R., Thom, J. M., Ruell, P. A., & Thompson, M. W. (2001). The influence of menstrual cycle phase on skeletal muscle contractile characteristics in humans. *The Journal of Physiology, 530*, 161-166.

Jarvik, J. G., Hollingworth, W., Heagerty, P. J., Haynor, D. R., Boyko, E. J., & Deyo, R. A. (2005). Three-year incidence of low back pain in an initially asymptomatic cohort: Clinical and imaging risk factors. *Spine, 30*(13), 1541-1549.

Jonsson, B. (1970). The functions of individual muscles in the lumbar part of the spinae muscle. *Electromyography, 10*(1), 5-21.

Karageanes, S. J., Blackburn, K., & Vangelos, Z. A. (2000). The association of the menstrual cycle with the laxity of the anterior cruciate ligament in adolescent female athletes. *Clinical Journal of Sport Medicine, 10*(3), 162-168.

Kavcic, N., Grenier, S., & McGill, S. M. (2004). Determining the stabilizing role of individual torso muscles during rehabilitation exercises. *Spine, 29*(11), 1254-1265.

Kellis, E., Arabatzi, F., & Papadopoulos, C. (2003). Muscle co-activation around the knee in drop jumping using the co-contraction index. *Journal of Electromyography and Kinesiology, 13*(3), 229-238.

Kendall, F. P., & McCreary, E. K. (1983). *Muscles: Testing and function* (3rd ed.). New Zealand: Williams & Wilkins.

Krause, N., Ragland, D. R., Greiner, B. A., Fisher, J. M., Holman, B. L., & Selvin, S. (1997). Physical workload and ergonomic factors associated with prevalence of back and neck pain in urban transit operators. *Spine, 22*(18), 2117-2127.

Kutz, M. (2009). Biomechanics of the musculoskeletal system. *Biomedical engineering and design handbook*. New York: McGraw-Hill.

Lamoth, C. J., Meijer, O. G., Daffertshofer, A., Wuisman, P. I., & Beek, P. J. (2006). Effects of chronic low back pain on trunk coordination and back muscle activity during walking: Changes in motor control. *European Spine Journal, 15*(1), 23-40.

Landry, S. C., McKean, K. A., Hubley-Kozey, C. L., Stanish, W. D., & Deluzio, K. J. (2009). Gender differences exist in neuromuscular control patterns during the pre-contact and early stance phase of an unanticipated side-cut and cross-cut maneuver in 15-18 years old adolescent soccer players. *Journal of Electromyography and Kinesiology, 19*(5), e370-9.

Lariviere, C., Arsenault, A. B., Gravel, D., Gagnon, D., & Loisel, P. (2002). Evaluation of measurement strategies to increase the reliability of EMG indices to assess back muscle fatigue and recovery. *Journal of Electromyography and Kinesiology, 12*(2), 91-102.

Lariviere, C., Gagnon, D., & Loisel, P. (2000). An application of pattern recognition for the comparison of trunk muscles EMG waveforms between subjects with and without chronic low back pain during flexion-extension and lateral bending tasks. *Journal of Electromyography and Kinesiology, 10*(4), 261-273.

Leader, L. R., Russell, T., Clifford, K., & Stenning, B. (1991). The clinical value of clearplan home ovulation detection kits in infertility practice. *The Australian & New Zealand Journal of Obstetrics & Gynaecology, 31*(2), 142-144.

Lebrun, C. M., McKenzie, D. C., Prior, J. C., & Taunton, J. E. (1995). Effects of menstrual cycle phase on athletic performance. *Medicine and Science in Sports and Exercise, 27*, 437-444.

Lee, C. Y., Liu, X., Smith, C. L., Zhang, X., Hsu, H. C., Wang, D. Y., et al. (2004). The combined regulation of estrogen and cyclic tension on fibroblast biosynthesis derived from anterior cruciate ligament. *Matrix Biology, 23*(5), 323-329.

Lemoine, S., Granier, P., Tiffoche, C., Rannou-Bekono, F., Thieulant, M. L., & Delamarche, P. (2003). Estrogen receptor alpha mRNA in human skeletal muscles. *Medicine and Science in Sports and Exercise, 35*(3), 439-443.

Liebenson, C. (2000). The quadratus lumborum and spinal stability. *Journal of Bodywork and Movement Therapies, 4*(1), 49-55.

Linde, R.S., Metter, E.J., Lynch, N.A., Fleg, J.L., Fozard, J.L., Tobin, J., Roy, T.A. & Hurley. B.F. (1997). Age and gender comparisons of muscle strength in 654 women and men aged 20-93 yr. *Journal of Applied Physiology, 83*, 1581-1587.

Liu, S. H., al-Shaikh, R., Panossian, V., Yang, R. S., Nelson, S. D., Soleiman, N., et al. (1996). Primary immunolocalization of estrogen and progesterone target cells in the human anterior cruciate ligament. *Journal of Orthopaedic Research, 14*(4), 526-533.

Liu, S. H., Al-Shaikh, R. A., Panossian, V., Finerman, G. A., & Lane, J. M. (1997). Estrogen affects the cellular metabolism of the anterior cruciate ligament. A potential explanation for female athletic injury. *The American Journal of Sports Medicine, 25*(5), 704-709.

Loeb, G. E., & Gans, C. (1986). In Univ. of Chicago Press (Ed.), *Electromyography for experimentalists*. Chicago, IL: Univ. of Chicago Press.

Lovering, R. M., & Romani, W. A. (2005). Effect of testosterone on the female anterior cruciate ligament. *American Journal of Physiology, 289*(1), R15-22.

Lowery, M. M., Stoykov, N. S., & Kuiken, T. A. (2003). A simulation study to examine the use of cross-correlation as an estimate of surface EMG cross talk. *Journal of Applied Physiology, 94*(4), 1324-1334.

MacDonald, D., Moseley, G. L., & Hodges, P. W. (2010). People with recurrent low back pain respond differently to trunk loading despite remission from symptoms. *Spine, 35*(7), 818-824.

Macfarlane, G. J., Beasley, M., Jones, E. A., Prescott, G. J., Docking, R., Keeley, P., et al. (2011). The prevalence and management of low back pain across adulthood: Results from a population-based cross-sectional study (the MUSICIAN study). *Pain, 153*(1), 27-32.

Mannion, A. F., Dumas, G. A., Cooper, R. G., Espinosa, F. J., Faris, M. W., & Stevenson, J. M. (1997). Muscle fibre size and type distribution in thoracic and lumbar regions of erector spinae in healthy subjects without low back pain: Normal values and sex differences. *Journal of Anatomy, 190*, 505-513.

Markolf, K. L., Graff-Radford, A., & Amstutz, H. C. (1978). In vivo knee stability. A quantitative assessment using an instrumented clinical testing apparatus. *The Journal of Bone and Joint Surgery. American Volume, 60*(5), 664-674.

Markolf, K. L., Mensch, J. S., & Amstutz, H. C. (1976). Stiffness and laxity of the knee--the contributions of the supporting structures. A quantitative in vitro study. *The Journal of Bone and Joint Surgery. American Volume, 58*(5), 583-594.

Marras, W. S., & Davis, K. G. (1998). Spine loading during asymmetric lifting using one versus two hands. *Ergonomics, 41*(6), 817-834.

Marras, W. S., Davis, K. G., & Jorgensen, M. (2003). Gender influences on spine loads during complex lifting. *The Spine Journal, 3*(2), 93-99.

Marras, W. S., Jorgensen, M. J., Granata, K. P., & Wiand, B. (2001). Female and male trunk geometry: Size and prediction of the spine loading trunk muscles derived from MRI. *Clinical Biomechanics, 16*(1), 38-46.

Mathur, S., Eng, J. J., & MacIntyre, D. L. (2005). Reliability of surface EMG during sustained contractions of the quadriceps. *Journal of Electromyography and Kinesiology, 15*(1), 102-110.

McGill, S., Seguin, J., & Bennett, G. (1994). Passive stiffness of the lumbar torso in flexion, extension, lateral bending, and axial rotation. effect of belt wearing and breath holding. *Spine, 19*(6), 696-704.

McGill, S. M. (1991). Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: Implications for lumbar mechanics. *Journal of Orthopaedic Research, 9*(1), 91-103.

McGill, S. M. (1998). Low back exercises: Prescription for the healthy back and when recovering from injury. *Resources manual for guidelines for exercise testing and prescription* (3rd Edition). Baltimore: Williams and Wilkins.

McGill, S. M. (1999). Stability: From biomechanical concept to chiropractic practice. *Journal of the Canadian Chiropractic Association, 43*(2), 79-92.

McGill, S. M., Grenier, S., Kavcic, N., & Cholewicki, J. (2003). Coordination of muscle activity to assure stability of the lumbar spine. *Journal of Electromyography and Kinesiology, 13*(4), 353-359.

McGill, S. M., Sharratt, M. T., & Seguin, J. P. (1995). Loads on spinal tissues during simultaneous lifting and ventilatory challenge. *Ergonomics, 38*(9), 1772-1792.

McKean, K. (2003). Principal Component Analysis - User's Manual. *Unpublished Report.*

Menz, H. B., Lord, S. R., & Fitzpatrick, R. C. (2003). Age-related differences in walking stability. *Age and Ageing, 32*(2), 137-142.

Miller, A. E., MacDougall, J. D., Tarnopolsky, M. A., & Sale, D. G. (1993). Gender differences in strength and muscle fiber characteristics. *European Journal of Applied Physiology and Occupational Physiology, 66*(3), 254-262.

Miller, B. F., Hansen, M., Olesen, J. L., Schwarz, P., Babraj, J. A., Smith, K., et al. (2007). Tendon collagen synthesis at rest and after exercise in women. *Journal of Applied Physiology, 102*(2), 541-546.

Mirka, G., Kelaher, D., Baker, A., Harrison, A., & Davis, J. (1997). Selective activation of the external oblique musculature during axial torque production. *Clinical Biomechanics, 12*(3), 172-180.

Mori, A. (2004). Electromyographic activity of selected trunk muscles during stabilization exercises using a gym ball. *Electromyography and Clinical Neurophysiology, 44*(1), 57-64.

Moritani, T., & deVries, H. A. (1978). Reexamination of the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction. *American Journal of Physical Medicine, 57*(6), 263-277.

Nachemson, A., & Morris, J. M. (1964). In vivo measurements of intradiscal pressure. discometry, a method for the determination of pressure in the lower lumbar discs. *The Journal of Bone and Joint Surgery, 46*, 1077-1092.

Nadler, S. F., Malanga, G. A., Bartoli, L. A., Feinberg, J. H., Prybicien, M., & Deprince, M. (2002). Hip muscle imbalance and low back pain in athletes: Influence of core strengthening. *Medicine and Science in Sports and Exercise, 34*(1), 9-16.

Nadler, S. F., Malanga, G. A., DePrince, M., Stitik, T. P., & Feinberg, J. H. (2000). The relationship between lower extremity injury, low back pain, and hip muscle strength in male and female collegiate athletes. *Clinical Journal of Sport Medicine, 10*(2), 89-97.

Nadler, S. F., Malanga, G. A., Feinberg, J. H., Prybicien, M., Stitik, T. P., & DePrince, M. (2001). Relationship between hip muscle imbalance and occurrence of low back pain in collegiate athletes: A prospective study. *American Journal of Physical Medicine & Rehabilitation, 80*(8), 572-577.

Nelson-Wong, E., & Callaghan, J. P. (2010). Is muscle co-activation a predisposing factor for low back pain development during standing? A multifactorial approach for early identification of at-risk individuals. *Journal of Electromyography and Kinesiology, 20*(2), 256-263.

Ng, J. K., Kippers, V., & Richardson, C. A. (1998). Muscle fibre orientation of abdominal muscles and suggested surface EMG electrode positions. *Electromyography and Clinical Neurophysiology, 38*(1), 51-58.

Nouwen, A., Van Akkerveeken, P. F., & Versloot, J. M. (1987). Patterns of muscular activity during movement in patients with chronic low-back pain. *Spine, 12*(8), 777-782.

Noyes, F. R., Mooar, P. A., Matthews, D. S., & Butler, D. L. (1983). The symptomatic anterior cruciate-deficient knee. part I: The long-term functional disability in athletically active individuals. *The Journal of Bone and Joint Surgery, American Volume, 65*(2), 154-162.

Olsen, O. E., Myklebust, G., Engebretsen, L., & Bahr, R. (2004). Injury mechanisms for anterior cruciate ligament injuries in team handball: A systematic video analysis. *The American Journal of Sports Medicine, 32*(4), 1002-1012.

O'Sullivan, P. B., Twomey, L., & Allison, G. T. (1998). Altered abdominal muscle recruitment in patients with chronic back pain following a specific exercise intervention. *The Journal of Orthopaedic and Sports Physical Therapy, 27*(2), 114-124.

Panjabi, M. M. (1992). The stabilizing system of the spine. part I. function, dysfunction, adaptation, and enhancement. *Journal of Spinal Disorders, 5*(4), 383-389.

Panjabi, M. M. (2006). A hypothesis of chronic back pain: Ligament subfailure injuries lead to muscle control dysfunction. *European Spine Journal, 15*(5), 668-676.

Park, S. K., Stefanyshyn, D. J., Loitz-Ramage, B., Hart, D. A., & Ronsky, J. L. (2009). Changing hormone levels during the menstrual cycle affect knee laxity and stiffness in healthy female subjects. *The American Journal of Sports Medicine, 37*(3), 588-598.

Pichler, W., Tesch, N. P., Schwantzer, G., Fronhofer, G., Boldin, C., Hausleitner, L., et al. (2008). Differences in length and cross-section of semitendinosus and gracilis tendons and their effect on anterior cruciate ligament reconstruction: A cadaver study. *The Journal of Bone and Joint Surgery, 90*(4), 516-519.

Pinar, G., Colak, M., & Oksuz, E. (2011). Premenstrual syndrome in turkish college students and its effects on life quality. *Sexual & Reproductive Healthcare, 2*(1), 21-27.

Pool-Goudzwaard, A. L., Vleeming, A., Stoeckart, R., Snijders, C. J., & Mens, J. M. (1998). Insufficient lumbopelvic stability: A clinical, anatomical and biomechanical approach to 'a-specific' low back pain. *Manual Therapy, 3*(1), 12-20.

Posthuma, B. W., Bass, M. J., Bull, S. B., & Nisker, J. A. (1987). Detecting changes in functional ability in women with premenstrual syndrome. *American Journal of Obstetrics and Gynecology, 156*(2), 275-278.

Renstrom, P., Ljungqvist, A., Arendt, E., Beynnon, B., Fukubayashi, T., Garrett, W., et al. (2008). Non-contact ACL injuries in female athletes: An international olympic committee current concepts statement. *British Journal of Sports Medicine, 42*(6), 394-412.

Richardson, C. A., Jull, G. A., Hodges, P. W., & Hides, J. A. (Eds.). (1999). *Therapeutic exercise for lumbopelvic stabilization: A motor control approach for the treatment and prevention of low back pain* (2nd Edition ed.). Edinburgh, New York: Churchill Livingstone.

Romani, W., Curl, L. A., & Lovering, R. (2001). The effect of endogenous estradiol levels at three phases of the menstrual cycle on anterior cruciate ligament stiffness in active females. *Journal of Athletic Training, 2*, S62.

Romani, W., Patrie, J., Curl, L. A., & Flaws, J. A. (2003). The correlations between estradiol, estrone, estriol, progesterone, and sex hormone-binding globulin and anterior cruciate ligament stiffness in healthy, active females. *Journal of Women's Health (2002), 12*(3), 287-298.

Sale, D.G., MacDougall, J.D., Always, S.E. & Sutton, J.R. (1987). Voluntary strength and muscle characteristics in untrained men and women and male bodybuilders. *Journal of Applied Physiology, 62*, 1786-1793.

Schellenberg, R. (2001). Treatment for the premenstrual syndrome with agnus castus fruit extract: Prospective, randomised, placebo controlled study. *British Medical Journal, 322*(7279), 134-137.

Schneider, S., Randoll, D., & Buchner, M. (2006). Why do women have back pain more than men? A representative prevalence study in the federal republic of Germany. *The Clinical Journal of Pain, 22*(8), 738-747.

Schneider, S., Schmitt, H., Zoller, S., & Schiltenwolf, M. (2005). Workplace stress, lifestyle and social factors as correlates of back pain: A representative study of the German working population. *International Archives of Occupational and Environmental Health, 78*(4), 253-269.

Seemann, S., & Reinhardt, A. (2000). Blood sample collection from a peripheral catheter system compared with phlebotomy. *Journal of Intravenous Nursing, 23*(5), 290-297.

Shultz, S. J., Gansneder, B. M., Sander, T. C., Kirk, S. E., & Perrin, D. H. (2006). Absolute serum hormone levels predict the magnitude of change in anterior knee laxity across the menstrual cycle. *Journal of Orthopaedic Research, 24*(2), 124-131.

Shultz, S. J., Kirk, S. E., Johnson, M. L., Sander, T. C., & Perrin, D. H. (2004). Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Medicine and Science in Sports and Exercise, 36*(7), 1165-1174.

Shultz, S. J., Schmitz, R. J., Nguyen, A. D., Chaudhari, A. M., Padua, D. A., McLean, S. G., et al. (2010). ACL research retreat V: An update on ACL injury risk and prevention. *Journal of Athletic Training, 45*(5), 499-508.

Shultz, S. J., Shimokochi, Y., Nguyen, A. D., Schmitz, R. J., Beynnon, B. D., & Perrin, D. H. (2007). Measurement of varus-valgus and internal-external rotational knee laxities in vivo--part II: Relationship with anterior-posterior and general joint laxity in males and females. *Journal of Orthopaedic Research, 25*(8), 989-996.

Silfies, S. P., Bhattacharya, A., Biely, S., Smith, S. S., & Giszter, S. (2009). Trunk control during standing reach: A dynamical system analysis of movement strategies in patients with mechanical low back pain. *Gait & Posture, 29*(3), 370-376.

Sirin, A. V., & Patla, A. E. (1987). Myoelectric changes in the triceps surae muscles under sustained contractions. evidence for synergism. *European Journal of Applied Physiology and Occupational Physiology, 56*(2), 238-244.

Sizonenko, P. C. (1978). Endocrinology in preadolescents and adolescents. I. Hormonal changes during normal puberty. *American Journal of Diseases of Children (1960), 132*(7), 704-712.

Smith, B. A., Livesay, G. A., & Woo, S. L. (1993). Biology and biomechanics of the anterior cruciate ligament. *Clinics in Sports Medicine, 12*(4), 637-670.

Snijders, C. J., Vleeming, A., & Stoeckart, R. (1993). Transfer of lumbosacral load to iliac bones and legs: Part 1: Biomechanics of self-bracing of the sacroiliac joints and its significance for treatment and exercise. *Clinical Biomechanics, 8*(6), 285-294.

Soderberg, G. L. (Ed.). (1992). *Selected topics in surface electromyography for use in the occupational setting: Expert perspectives* (first ed. ed.). USA: DHHS: (NIOSH) Publication No. 91-100.

Solomonow, M., Baratta, R., Bernardi, M., Zhou, B., Lu, Y., Zhu, M., et al. (1994). Surface and wire EMG crosstalk in neighbouring muscles. *Journal of Electromyography and Kinesiology, 4*(3), 131-142.

Stanczyk, F. Z., Cho, M. M., Endres, D. B., Morrison, J. L., Patel, S., & Paulson, R. J. (2003). Limitations of direct estradiol and testosterone immunoassay kits. *Steroids, 68*(14), 1173-1178.

Stocker, J. M. (1974). Motor performance and state anxiety at selected stages of the menstrual cycle. *Dissertation Abstracts International, 34*, 3971-3972.

Sundström, I., Andersson, A., Nyberg, S., Ashbrook, D., Purdy, R. H., & Bäckström, T. (1998). Patients with premenstrual syndrome have a different sensitivity to a neuroactive steroid during the menstrual cycle compared to control subjects. *Neuroendocrinology, 67*, 126-138.

Thelen, D. G., Schultz, A. B., & Ashton-Miller, J. A. (1995). Co-contraction of lumbar muscles during the development of time-varying triaxial moments. *Journal of Orthopaedic Research, 13*(3), 390-398.

Thorstensson, A., & Carlson, H. (1987). Fibre types in human lumbar back muscles. *Acta Physiologica Scandinavica, 131*(2), 195-202.

Tortora, G. J., & Anagnostakos, N. P. (Eds.). (1987). *Principles of anatomy and physiology* (5th ed. ed.). New York, NY: Harper & Row Publishers.

Tremblay, M. S., Chu, S. Y., & Mureika, R. (1995). Methodological and statistical considerations for exercise-related hormone evaluations. *Sports Medicine, 20*(2), 90-108.

Urquhart, D. M., Hodges, P. W., Allen, T. J., & Story, I. H. (2005). Abdominal muscle recruitment during a range of voluntary exercises. *Manual Therapy, 10*(2), 144-153.

van Belle, G. (2008). In Balding D. J., Cressie N. A., Fitzmaurice G. M., Johnstone I. M., Molenberghs G. and Weisberg S. (Eds.), *Statistical rules of thumb* (2nd Edition ed.). Hoboken, NJ: John Wiley & Sons.

van Dieen, J. H., Cholewicki, J., & Radebold, A. (2003). Trunk muscle recruitment patterns in patients with low back pain enhance the stability of the lumbar spine. *Spine, 28*(8), 834-841.

Van Lunen, B. L., Roberts, J., Branch, J. D., & Dowling, E. A. (2003). Association of menstrual-cycle hormone changes with anterior cruciate ligament laxity measurements. *Journal of Athletic Training, 38*(4), 298-303.

Vescovi, J. D. (2011). The menstrual cycle and anterior cruciate ligament injury risk: Implications of menstrual cycle variability. *Sports Medicine, 41*(2), 91-101.

Vezina, M. J., & Hubley-Kozey, C. L. (2000). Muscle activation in therapeutic exercises to improve trunk stability. *Archives of Physical Medicine and Rehabilitation, 81*(10), 1370-1379.

Vink, P., van der Velde, E. A., & Verbout, A. J. (1987). A functional subdivision of the lumbar extensor musculature. recruitment patterns and force-RA-EMG relationships under isometric conditions. *Electromyography and Clinical Neurophysiology, 27*(8), 517-525.

Vleeming, A., Stoeckart, R., Volkers, A. C., & Snijders, C. J. (1990). Relation between form and function in the sacroiliac joint. part I: Clinical anatomical aspects. *Spine, 15*(2), 130-132.

Vleeming, A., Volkers, A. C., Snijders, C. J., & Stoeckart, R. (1990). Relation between form and function in the sacroiliac joint. part II: Biomechanical aspects. *Spine, 15*(2), 133-136.

Warden, S. J., Saxon, L. K., Castillo, A. B., & Turner, C. H. (2006). Knee ligament mechanical properties are not influenced by estrogen or its receptors. *American Journal of Physiology, Endocrinology and Metabolism, 290*(5), E1034-40.

Weir, J. P. (2005). Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *Journal of Strength and Conditioning Research, 19*(1), 231-240.

Wilk, A., Gustafsson, T., Esbjornsson, M., Johansson, O., Ekman, M., Sundberg, C. J., et al. (2005). Expression of oestrogen receptor alpha and beta is higher in skeletal muscle of highly endurance-trained than of moderately active men. *Acta Physiologica Scandinavica, 184*(2), 105-112.

Wilke, H. J., Wolf, S., Claes, L. E., Arand, M., & Wiesend, A. (1995). Stability increase of the lumbar spine with different muscle groups. A biomechanical in vitro study. *Spine, 20*(2), 192-198.

Winter, D. A., Fuglevand, A. J., & Archer, S. E. (1994). Crosstalk in surface electromyography: Theoretical and practical estimates. *Journal of Electromyography and Kinesiology, 4*(1), 15-26.

Winter, E. M., & Brookes, F. B. (1991). Electromechanical response times and muscle elasticity in men and women. *European Journal of Applied Physiology and Occupational Physiology, 63*(2), 124-128.

Wojtys, E. M., Huston, L. J., Boynton, M. D., Spindler, K. P., & Lindenfeld, T. N. (2002). The effect of the menstrual cycle on anterior cruciate ligament injuries in women as determined by hormone levels. *The American Journal of Sports Medicine, 30*(2), 182-188.

Wojtys, E. M., Huston, L. J., Lindenfeld, T. N., Hewett, T. E., & Greenfield, M. L. (1998). Association between the menstrual cycle and anterior cruciate ligament injuries in female athletes. *The American Journal of Sports Medicine, 26*(5), 614-619.

Woods, J. J., & Bigland-Ritchie, B. (1983). Linear and non-linear surface EMG/force relationships in human muscles. An anatomical/functional argument for the existence of both. *American Journal of Physical Medicine, 62*(6), 287-299.

Wrigley, A. T., Albert, W. J., Deluzio, K. J., & Stevenson, J. M. (2005). Differentiating lifting technique between those who develop low back pain and those who do not. *Clinical Biomechanics, 20*(3), 254-263.

Wust, R. C., Morse, C. I., de Haan, A., Jones, D. A., & Degens, H. (2008). Sex differences in contractile properties and fatigue resistance of human skeletal muscle. *Experimental Physiology, 93*(7), 843-850.

Yang, J. F., & Winter, D. A. (1983). Electromyography reliability in maximal and submaximal isometric contractions. *Archives of Physical Medicine and Rehabilitation, 64*(9), 417-420.

Zazulak, B. T., Paterno, M., Myer, G. D., Romani, W. A., & Hewett, T. E. (2006). The effects of the menstrual cycle on anterior knee laxity: A systematic review. *Sports Medicine, 36*(10), 847-862.

Principal component analysis (PCA) is a multivariate statistical tool that describes the

variability within a group of related variables by transforming the original variables into new

uncorrelated variables (Jackson, 1991). The following is a summary of PCA based on a report by

McKean, 2003. How it is applied to the specific data sets in the study is illustrated. PCA is

applied to a data set in n x p format (n are observations and p are variables). When PCA is

applied to the EMG data each muscle waveform from each subject represents an observation (n),

so for example, with 9 subjects performing the task at two different points in their menstrual

cycle with 24 muscles sites measured n would be 9*2*24 = 432. Each data point in the time

normalized EMG waveform represents a variable (p=101) (See Figure 1). Means and variances

are calculated on each variable across observations; these values will be discussed throughout the

analysis.

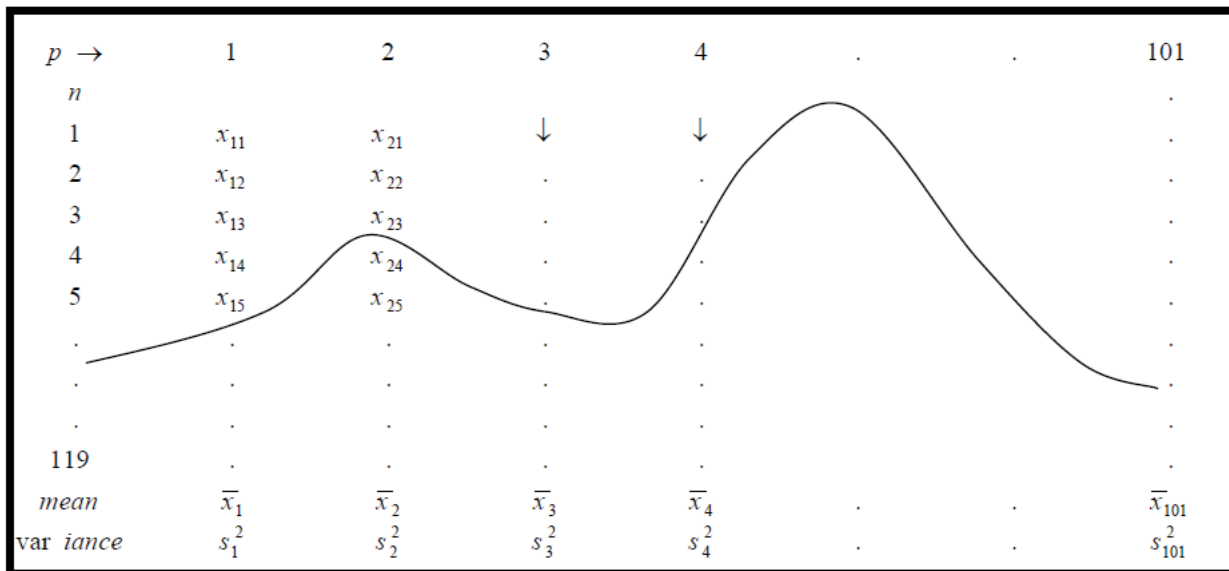| $p \rightarrow$	1	2	3	4	.	.	101
n							.
1	x_{11}	x_{21}	$\downarrow$	$\downarrow$			.
2	x_{12}	x_{22}	.	.			.
3	x_{13}	x_{23}	.	.			.
4	x_{14}	x_{24}	.	.			.
5	x_{15}	x_{25}	.	.			.
.	.	.	.	.			.
.	.	.	.	.			.
119	.	.	.	.			.
mean	$\bar{x}_1$	$\bar{x}_2$	$\bar{x}_3$	$\bar{x}_4$	.	.	$\bar{x}_{101}$
variance	s_1^2	s_2^2	s_3^2	s_4^2	.	.	s_{101}^2

Figure A1: How the data should be structured when conducting a n x p PCA (figure from unpublished report by McKean, 2003).

This data is then used to construct a covariance matrix (S) which is composed of the variances in the diagonal entries of S. The remaining entries in S are filled with covariances (See Equation 3). Variance (S_i^2) is the average square of the difference between the value of a variable for one observation and the mean for that given variable for all observations (all subjects) (See Equation 1). Therefore variance indicates how much a variable strays from its mean (Jackson, 1991). Covariance (S_{ij}) gives a measure of how much one variable depends on another. It is calculated by multiplying the variable (i) for each subject (k) by the next variable (j) for the same subject (k). The second part of the covariance equation involves multiplying the sum of the values of the first variable for all subjects by the sum of the values of the second variable for all subjects (Jackson, 1991) (See Equation 2).

$$s_i^2 = \frac{\sum_{i=1}^{n}[x_i - \overline{x}]}{(n-1)}$$
(Equation 1)

$$s_{ij} = \frac{n\sum x_{ik}x_{jk} - \sum x_{ik}\sum x_{jk}}{[n(n-1)]}$$
(Equation 2)

$$S = \begin{bmatrix} S_1^2 & S_{21} & S_{31} & S_{41} & \cdot & S_{1011} \\ S_{12} & S_2^2 & S_{23} & \cdot & \cdot & \cdot \\ S_{13} & S_{23} & S_3^2 & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ S_{1101} & S_{2101} & \cdot & \cdot & \cdot & S_{101}^2 \end{bmatrix}$$
(Equation 3)

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Extracting the eigenvectors from the covariance matrix (S) yields an orthogonal matrix (U). The diagonal elements of U are the eigenvalues (L) which are attained by pre and post multiplying S by U' (See Equation 4). Eigenvalues indicate how much variance is described by each eigenvector (Hubley-Kozey et al., 2009).

$$U'SU = L \qquad \text{(Equation 4)}$$

$$[pxp]'[pxp][pxp] \ = \ [pxp]$$

The eigenvectors (U) are transformed variables that each describe a feature of variation in the data (Jackson, 1991). Eigenvectors are also called principal components. The maximum number of principal components that can be extracted from a data set is equal to the number of variables within the data. However, since each principal component describes less and less variation in the data, it is only usually relevant to examine the first few principal components. The eigenvalues, which are the diagonal elements of the U matrix (referred to as L), indicate how much of the total variation in the data set is described by each particular principal component. The transformation matrix (U) is then used to rotate the original data observations into new, uncorrelated observations (Z) (See equation 5).

$$Z \quad = [x_i - x_{mean}]\,U \qquad \text{(Equation 5)}$$

$$[nxp] \ = [nxp]\,[pxp]$$

The newly transformed variables are the selected principal components and the newly transformed observations are referred to as Z scores (Jackson, 1991). Z scores are also referred to as PC scores. Each subject has one Z score for how closely their EMG waveform reflects each important principal component (Hubley-Kozey et al., 2002). If a subject has a high Z score for a particular principal component it means that their waveform exhibits a large amount of variation for the particular feature picked up by that principal component.

Appendix B: The Test-retest Reliability of Trunk EMG – Focusing on the Amplitude and Temporal EMG Characteristics Analyzed using Principal Component Analysis.

Electromyography (EMG) is an important measurement technique that provides an objective measure of the electrophysiological process associated with the generation of force in muscles (De Luca, 1997). EMG has been extensively used in the study of low back disorders as it has been hypothesized that abnormal muscular activation patterns are linked to dysfunction in the biomechanical system of the spine (Panjabi, 1992). The EMG signals are most commonly analyzed by evaluating amplitude, temporal and frequency variables. However, there is a large amount of inherent variability in EMG signals due to several factors such as electrode placements, signal crosstalk and spatial filtering (De Luca, 1997), which can mask true differences. To make day to day comparisons, establishing reliability of an EMG measurement protocol is necessary.

Reliability refers to the consistency of a test or measurement (Weir, 2005). It is easy for the seemingly clear concept of reliability to become unclear as it can be assessed in several different contexts. Inter-rater reliability tests the consistency of a measurement performed by several different investigators and is quite different than test-retest reliability, which tests the consistency of a test performed by the same investigator at different time points. For this study we were interested in test-retest reliability, sometimes referred to as repeatability. This was necessary to establish since study two (Chapter 5) in this thesis examined EMG from tests performed on different days of a subject's menstrual cycle. It is important to establish that any differences that appear in the data between days is due to the independent variable being examined (time point in menstrual cycle) and not due to some inherent variability within the test protocol. To improve reliability all tests were performed by the same investigator.

Good within-day test-retest reliability of trunk muscle EMG has been reported in healthy

subjects (Allison et al., 1998; Dankaerts, O'Sullivan, Burnett, Straker, & Danneels, 2004;

Danneels et al., 2002; Danneels et al., 2001; O'Sullivan, Twomey, & Allison, 1998). However,

more pertinent to study two, between-day reliability of trunk muscle surface EMG has been

evaluated with varying results. Danneels et al., 2001 reported that amplitude measures of spinal

stabilizer muscle EMG during a wide range of exercise demonstrated good between-day

reliability (ICC > 0.75), although only when the tests were performed by the same investigator

(Danneels et al., 2001). Dankaerts et al., 2004 reported reliability of trunk muscle surface EMG

for both healthy subjects and subjects with low back pain. They demonstrated between-day

reliability that was excellent during sub-maximal exercises (ICC mean = 0.88, range 0.75 - 0.98)

and acceptable during maximal voluntary contraction (MVC) exercises (ICC mean = 0.7, range

0.19 – 0.99).

A confusing aspect of reliability studies are the multitude of reliability calculations

available to be used. Weir discusses the different calculations such as Pearson r coefficients,

coefficients of variation, limits of agreement from Bland-Altman plots, and intra-class

correlation coefficients to evaluate reliability (Weir, 2005). The Pearson *r* coeffiecient is

probably the most commonly used index of reliability, however its use is discouraged for

assessing test-retest reliability as it does not detect systematic differences (Weir, 2005). Bland-

Altman plots are useful for visually inspecting the overall reliability of the data but using the

limits of agreement produced by the plots as an index of reliability has been criticized (Hopkins,

2000; Weir, 2005). Intra-class correlations (ICCs) are commonly used in the test-retest literature.

An ICC is a relative measure of reliability derived using a ratio of variances from ANOVA.

However limitations exist in the use of ICCs on small homogeneous populations or those with

large variability that are tested on only two occasions. Paired t-tests can also be used when assessing the test-retest reliability of a measure from two occasions using the same subjects to see if there is a consistent and significant difference between the two measures.

The purpose of this study was to assess the test-retest reliability of amplitude and temporal characteristics of EMG signals collected during a dynamic leg-lifting task, the TST. Amplitude and temporal characteristics of the data were extracted using a data analysis technique called Principal Component Analysis (PCA) which allows EMG waveforms to be given Principal Component (PC) scores based on how closely they represent Principal Components generated that capture overall trends in the data (See Appendix A for details). Reliability of PC scores generated representing amplitude and temporal characteristics of the EMG waveform data were assessed using paired t-tests.

Methodology

Subjects

Five young healthy women participated in the reliability test. Participants were between 20 and 30 years of age (mean age of 24.8 ± 2.7 years, mean BMI of 21.9 ± 2 kg/m^2). Subjects were excluded if they had experienced any low back pain episodes within the past 12 months or if they had ever experienced a low back injury severe enough to require medical assistance. The presence of any neurological, musculoskeletal, vestibular or cardiorespiratory conditions also excluded a subject from the study. If subjects were not taking oral contraceptives they were tested at times that were not near ovulation time (not tested on cycle days 12-16).

Procedure

Prior to testing participants were given a complete description of the experimental protocol and risks and signed an informed consent form in accordance with the Health Sciences Research Ethics Board, Dalhousie University. Subjects came into the lab a total of two times on back-to-back days. During the first test session the subject's demographics (age and sex) were recorded as well as structural anthropometric parameters: body mass (kg) and height (cm). A brief assessment was performed by the principal investigator to check for normal posture and evidence of abnormalities such as scoliosis.

Experimental Task

The experimental task evaluated was the trunk stability test (TST) (Clarke-Davidson & Hubley-Kozey, 2005) (See Figure 2.2). Prior to beginning the task subjects were instructed to perform an abdominal hollowing maneuver. Proper timing of the task was ensured using event markers generated by pressure sensitive switches located on the right foot, the top of the right thigh and the crossbeam. Prior to data collection during the first session the subjects practiced the TST at least 5 times until they were able to perform it correctly with proper timing and technique. During each test session the subject performed five trials of the task with a 1 minute rest period between each trial.

Data acquisition

A surface EMG system (3-8 channel, Bortec Inc., Calgary, Alberta) was used for recording the myoelectric signals from 24 muscle sites on the back and abdominals. The EMG signals were pre-amplified (500x) and then amplified differentially (Bandpass 10-1000Hz; input impedance $> 10G\Omega$; CMRR 115dB) with three AMT-8 EMG systems (BortecInc. Calgary,

Alberta). The analog signal were sampled at 2000Hz using a 16-bit analogue-to-digital (A/D) converter (National instruments, CA-1000) using LABVIEW. Triggering signals from various sources during the tests were collected so that the EMG signals could be analysed in terms of timing phases.

Twenty-four pairs of silver/silver chloride electrodes (Red Dot, 10mm) were placed over the back and abdominal muscle sites (30 mm inter-electrode distance). Standardized landmarking procedures were used for placing the electrodes properly over six bilateral abdominal sites and six bilateral back sites. The abdominal muscles that were monitored include the rectus abdominus, the external obliques and the internal obliques. The back muscles that were monitored included the erector spinae (longissimus and iliocostalis) at different lumbar levels, the quadraus lumborum and the multifidus (Butler, Hubley-Kozey, & Kozey, 2009b). Though the electrode placements were standardized, minor location adjustments were made based on individual anthropometrics and palpations. Three ground electrodes were placed over the iliac crest as a reference for each amplifier. All electrodes were placed along the muscle fibers of the underlying muscles. Prior to electrode placement to reduce skin resistance standard skin preparation methods were used, which include shaving hair if necessary and abrading the skin with alcohol swabs (Vezina & Hubley-Kozey, 2000). Skin impedance was checked with a multimeter (Fluke 77) to ensure it was below 200KΩ before beginning testing. This is well below the acceptable skin/electrode to amplifier impedance ratio of 1% (Soderberg, 1992).

Validation exercises were used to isolate specific muscles to ensure proper electrode function and placement for each muscle site consistent with published protocol (Butler, Hubley-Kozey, & Kozey, 2009a; Butler, Hubley-Kozey, & Kozey, 2009b; Butler et al., 2010). During the validation exercises the signals were checked for quality to ensure low noise and good signal

with the gains on each channel adjusted to ensure maximum signal without the signal being clipped.

Normalization Procedures

A series of normalization exercises were performed to produce a maximal voluntary isometric contraction (MVIC) to which all data was normalized. A total of eight different isometric exercises previously used for normalization (Butler, Hubley-Kozey, & Kozey, 2009b)were performed to recruit MVICs from all muscle sites. During the isometric maximum contractions subjects were kept in a static position with the help of several straps. Manual resistance was provided as well, though more for proprioceptive purposes as the straps resisted the force produced by the subject. All normalization exercises were held for three seconds with a two minute rest interval between each contraction.

Motion Measurement

The Flock of Birds ^{TM} (FOB) motion system (Ascension Technology Inc., Burlington, Vermont) was used to monitor motion of the pelvis during the TST task. One electromagnetic sensor was placed on the subject's left iliac crest (See Figure 3.1). The sensor provided 6 degrees of freedom (x,y,z displacement, yaw, pitch and roll rotations) with respect to a global coordinate axis system located in the FOB source. Minimal pelvic motion was desired and thus any trials with excess motion were excluded from analysis. The analog signals were sampled at 50 Hz by a 16-bit analogue-to-digital (A/D) converter (National instruments, CA-1000) using LABVIEW. The measurements were not directly related to the anatomical reference frame however, yaw most closely represents anterior/posterior tilt of the pelvis, pitch most closely represents

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horizontal rotation (side to side) of the pelvis and roll most closely represents up and down tilt of the pelvis in the frontal plane.

Data Processing

All raw EMG signals were visually inspected for quality and noise levels or artefacts (eg. spikes, DC offsets). A custom program in Matlab® (MathWorks, Inc., Natick, MA. Version 7.3) was used to process the data. A recursive fifth order Butterworth high pass filter at 30Hz was used to remove the ECG signal from any low amplitude EMG signals in which it was present. The power spectrum was calculated for each EMG signal and if low level noise from the FOB system was detected it was removed with an inverse Fast Fourier transform. The raw EMG was corrected for the system and subject bias, adjusted for the true channel gain and full wave-rectified. The signals was then filtered with a 6Hz second order recursive Butterworth low pass filter to create a linear envelope profile. The normalization trials were used to determine the maximum amplitudes for each of the 24 muscle sites using a 500 msec moving window (Vezina & Hubley-Kozey, 2000). The maximum amplitudes were used to normalize the EMG signals for all trials resulting in EMG patterns that are reported in percent maximal voluntary isometric contraction (%MVIC). All EMG signals were time normalized to 100% of the movement from right foot off to right foot on using a linear interpolation algorithm.

The kinematic motion data from the FOB was low-pass filtered at 1-Hz using a recursive second order Butterworth filter using a customized program in Matlab® (MathWorks, Inc., Natick, MA. Version 7.3). The angular displacement of the marker was calculated as the difference in degrees between the maximum and minimum angular positions during the task.

Data Analysis

EMG patterns were extracted from the waveforms using Principal Component Analysis (PCA). PCA was performed using a customized Matlab® (MathWorks, Inc., Natick, MA. Version 7.3) program (Hubley-Kozey et al., 2009). To increase the strength of the PCA model for this smaller study the EMG waveforms generated from the five subjects on the two separate days were also combined with the EMG waveforms collected in the second study of this thesis (Chapter 5). Increasing the number of curves included in the model increases the ability of the model to describe overall trends seen in the data as opposed to features seen in single subjects. Two separate PCA models were run, one for the 12 abdominal muscle sites and one for the 12 back muscle sites. The PCA process is described in more detail in Appendix A. Briefly, an eigenvector decomposition was computed on the covariance matrix developed from all of the ensemble averaged EMG profiles from each individual trunk muscle for all participants. Data matrix dimensions were n (336) x p (101). Principal components (PCs) were extracted from the transformed matrix which represents overall trends seen in the data. Each PC captures a different feature of the EMG waveform data. Each measured waveform was given a PC score for each PC based on how similar the waveform is to the PC curve. PC scores can then be used in statistical analysis to compare temporal synchronies in the EMG between the two test days.

Statistical Analysis

Paired t-tests ($\alpha = 0.05$) were run to assess the differences in data between days for all 24 muscle sites for the first 2 (backs) or 3 (abdominals) PCs generated.

Results

Descriptive data for the subjects can be found in Table B1. The trunk FOB sensor captured angular motion of the pelvis in three directions (Yaw (Z), Pitch (Y) and Roll (X)). The greatest motion was in the direction of anterior/posterior tilt of the pelvis (Yaw) and was on average a maximum of approximately 4 degrees (2 degrees to each side of the midline). Differences between the two days were less than 0.3 degrees for all three directions and none of these differences were significant (See Table B2). These data confirm that minimal motion occurred during testing. There was no statistically significant difference in the time it took to complete the task on the first day (7.86 secs ± 0.36 secs) compared to the second day (7.83 secs ± 0.28 secs.

Table B1: Subject demographics. Average values with standard deviations in brackets. N=5.

Age	Mass (Kg)	Height (cm)	BMI (kg/m^2)
24.8 (2.7)	60.3 (5.6)	166.1 (3.4)	21.9 (1.8)

Table B2: Pelvic motion data. No significant differences (p<0.05) between days.

	Ant/Post Tilt (Yaw)	Hor Rotation (Pitch)	Side Flexion (Roll)
Day 1	4.1 (±1.1)	1.9 (±0.7)	3.0 (±0.5)
Day 2	4.1 (±1.0)	2.0 (±0.5)	3.2 (±1.1)

Note: Ant/Post = Anterior/Poster and Hor = Horizontal.

Abdominal EMG Waveform Analysis

EMG waveforms for all abdominal muscles during test day 1 and test day 2 are depicted in Figure B1. Visual inspection of the figure reveals that the day 1 and day 2 waveforms were very similar for most of the muscle sites. Subtle variation in muscle activation throughout the movement in the trunk sites was confirmed by PCA since 90.2% of the variance in the data was explained by 3 PCs (See Figure B2). PC1 captured the overall amplitude and shape

characteristics of the waveforms (See Figure B2A) and PC1 *score* is highly correlated to RMS amplitude (Hubley-Kozey et al., 2009). PC2 captured an increase in amplitude of the EMG signal during the middle of the motion when the right leg was being extended, relative to the beginning of the motion. Muscles with high PC2 scores increased their activity in response to the leg lifting task whereas muscles with low PC2 scores pre-activated and maintained a consistent level of activity throughout the entire task (See Figure B2B). PC3 captured a spike in activity at about 15% movement time corresponding to when the left leg was lifted off the bed (See Figure B2C). Representative high and low scoring curves for each of the 3 PCs are also depicted in Figure B2D, B2E and B2F.
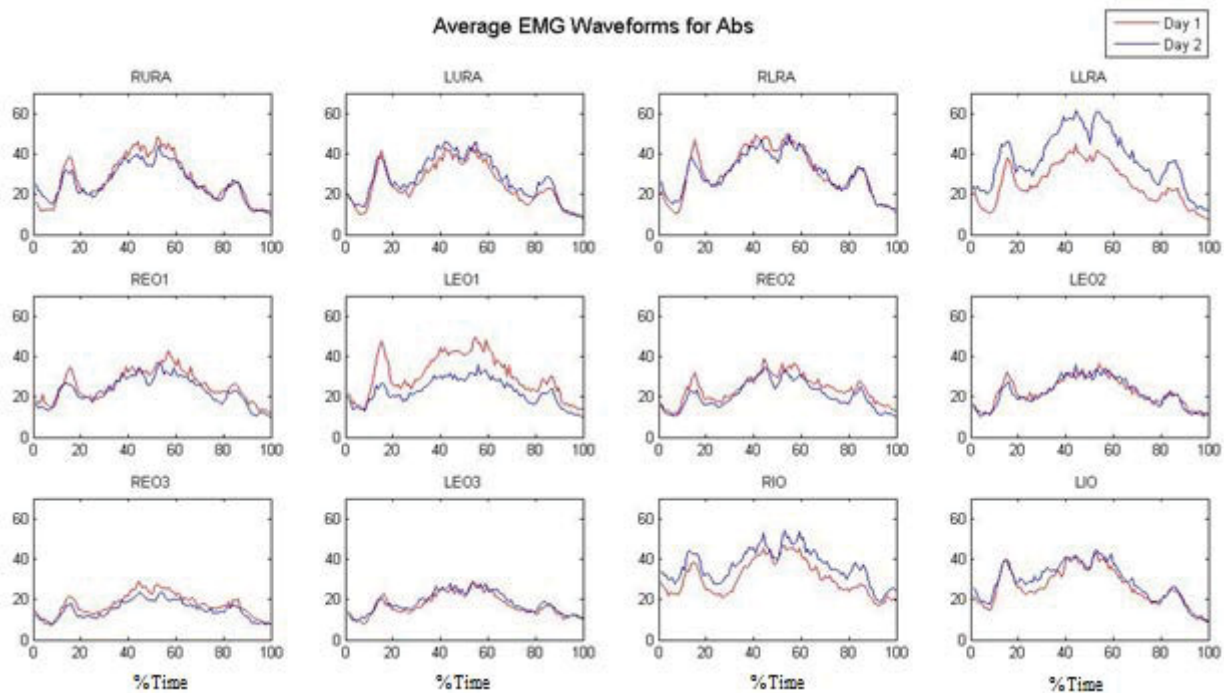

Figure B1: Ensemble average EMG waveforms for each of the 12 abdominal muscle sites averaged across subjects for day 1 and day 2. RLRA = right lower rectus abdominus; LLRA = left lower rectus abdominus; RURA = right upper rectus abdominus; LURA = left upper rectus abdominus; REO1= right external oblique site 1; LEO1 = left external oblique site 1; REO2= right external oblique site 2; LEO2 = left external oblique site 2; REO3= right external oblique site 3; LEO3 = left external oblique site 3; RIO = right internal oblique; LIO = left internal oblique.
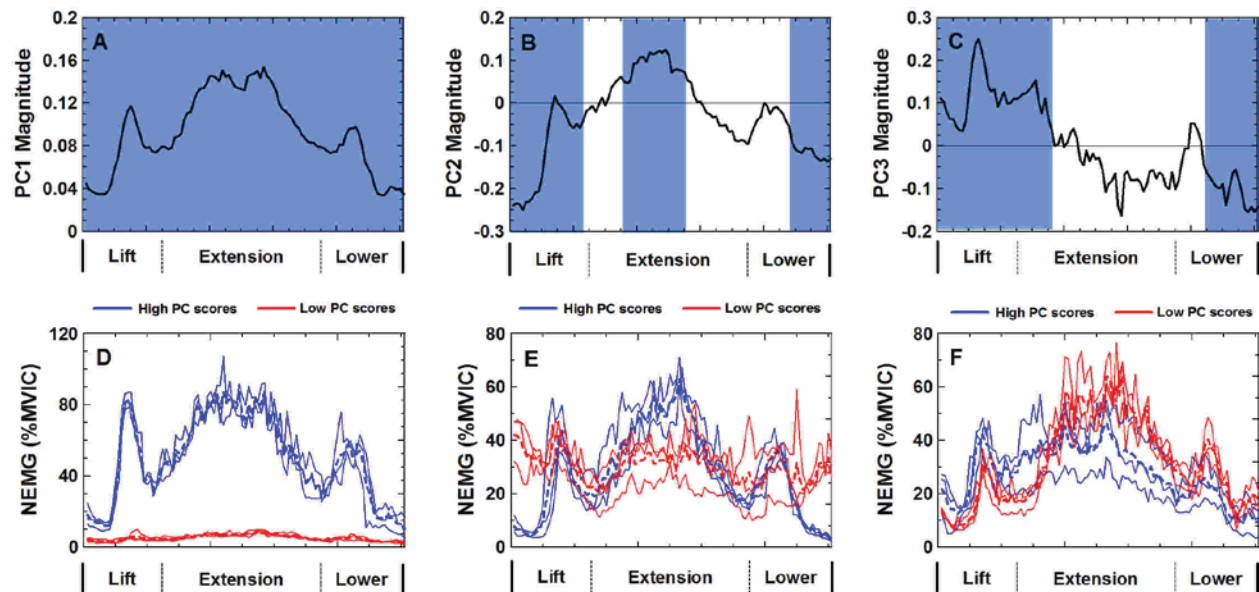
Figure B2: First three principal components for the abdominal muscles. Blue shading on curves A, B and C indicate portions of the curve where the most variance in the data is described by the PC. A) PC1 which explains 83.1% of the waveform variance; B) PC2 which explains 4.8% of the variance; C) PC3 which explains 2.3% of the variance; D-F) High and low scoring curves for PC1-3 with high scores shown in blue and low scores shown in red and the average high and average low scores bolded.

Three scores were significantly different between days. Of the 36 tests run this represents 8% of the tests which is only slightly higher than the 5% of tests that would return a significant result by chance. The scores that were significantly different were PC2 scores for RLRA (higher on day 1), PC2 scores for RIO (higher on day 2) and PC3 scores for LIO (higher on day 2) (see Table B3). Examining the averaged EMG waveforms in Figure B1 it appears that largest differences between the days occur in the LLRA and the LEO1 and therefore one might have expected those muscle sites to have shown significant differences between days. However when the EMG data for each separate subject is examined it becomes clear that the large differences seen in the averaged EMG curves are driven by one outlier subject having a very different LLRA waveforms between days and another outlier subject having very different LEO1 waveforms between days. These large differences seen in specific EMG curves in these two subjects between days could possibly be due to problems recruiting the specific muscle during normalizations on one specific day. With a small sample size of only 5 subjects these

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inconsistent differences can be seen in the averaged waveforms, however with a larger sample size they would likely be washed out.

Table B3: P-values from paired t-tests comparing PC scores on day 1 to PC scores on day 2. Significant differences between day 1 and day 2 were found in PC2 for RLRA and RIO and in PC3 for LIO. * indicates that the PC scores were statistically higher for day 1 and ** indicates that the PC scores were statistically higher for day 2.

Muscle	PC1			PC2			PC3		
	Day 1	Day 2	P-value	Day 1	Day 2	P-value	Day 1	Day 2	P-value
RURA	44.4 ± 142	22.4 ± 104	0.482	**13.6 ± 26**	**-7.9 ± 19**	**0.027***	6.5 ± 11	8.1 ± 14	0.782
LURA	20.0 ± 112	49.3 ±158	0.323	9.4 ± 15	6.8 ± 28	0.721	16.4 ± 10	18.9 ± 16	0.727
RLRA	81.7 ± 161	71.0 ± 136	0.855	9.2 ± 23	-9.4 ± 25	0.052	13.4 ± 17	9.9 ± 21	0.475
LLRA	24.3 ± 97	151 ± 120	0.183	10.4 ± 15	-2.1 ± 38	0.380	14.4 ± 12	16.6 ± 14	0.731
REO1	16.9 ± 105	-9.7 ± 119	0.627	-19.6 ± 53	-14.1 ± 45	0.813	-5.0 ± 14	2.6 ± 15	0.118
LEO1	72.9 ± 204	-14.7 ± 140	0.413	-9.5 ± 35	-14.7 ± 48	0.814	7.0 ± 22	0.8 ± 35	0.423
REO2	-1.8 ± 114	-30.0 ± 62	0.612	-14.0 ± 30	-3.8 ± 15	0.282	-8.2 ± 6.1	-7.1 ± 12	0.832
LEO2	-14.2 ±108	-18.0 ± 127	0.928	-7.8 ± 20	-2.5 ± 15	0.342	4.6 ± 10	-3.3 ± 16	0.241
REO3	-66.5 ± 60	-94.2 ± 17	0.368	-0.6 ± 18	-3.1 ± 22	0.750	-5.5 ± 8.1	-5.9 ± 8.1	0.951
LEO3	-66.9 ± 57	-57.5 ± 57	0.142	-3.0 ± 11	-5.4 ± 18	0.678	-6.3 ± 4.7	-5.7 ± 4.5	0.831
RIO	66.1 ± 55	124 ± 78	0.060	**-32.4 ± 26**	**-56.5 ± 16**	**0.017****	-2.1 ± 16	7.0 ± 11	0.367
LIO	30.5 ± 129	49.9 ± 111	0.428	-7.2 ± 30	-13.5 ± 41	0.717	**19.7 ± 8.0**	**27.8 ± 9.4**	**0.033****

Back EMG Waveform Analysis

EMG waveforms for all back muscles during test day 1 and test day 2 are depicted in Figure B3. Subtle variation in muscle activation throughout the movement in the trunk sites was confirmed by PCA since 96% of the variance in the data was explained by 2 PCs (See Figure B4). Like in the abdominals PC1 in the back extensors captured the overall amplitude and shape characteristics of the waveforms (See Figure B4A). PC2 captured a pattern of varied amplitude in the signal over the course of the task with reduced amplitude at the start and end of the task compared to the middle (See Figure B4B). Paired t-tests ($\alpha = 0.05$) between the two days for both PC scores for each of the muscle sites were not significantly different between days.
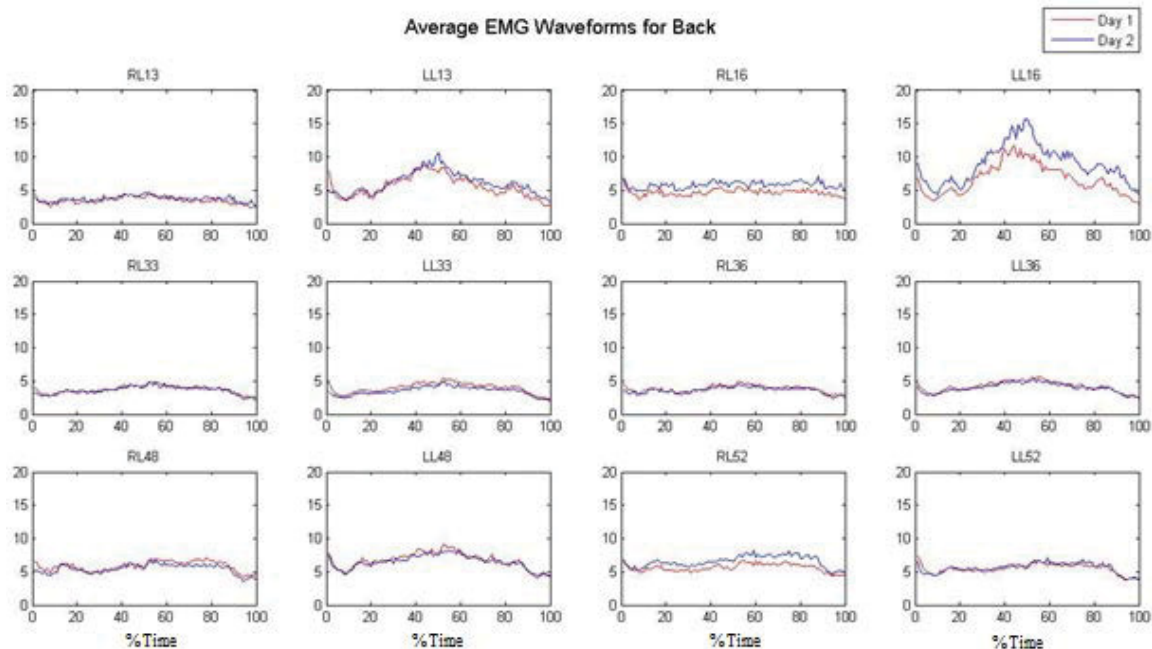
Average EMG Waveforms for Back

Figure B3: Ensemble average EMG waveforms for each of the 12 back muscle sites averaged across subjects for day 1 and day 2. RL13= right medial erector spinae (level L1); LL13= left medial erector spinae (level L1); RL16= right lateral erector spinae (level L1); LL16= left lateral erector spinae (level L1); RL33= right medial erector spinae (level L3); LL33= left medial erector spinae (level L3); RL36= right lateral erector spinae (level L3); LL36= left lateral erector spinae (level L3); RL48 = right quadratus lumborum; LL48 = left quadratus lumborum; RL52 = right multifidus; LL52 = left multifidus.
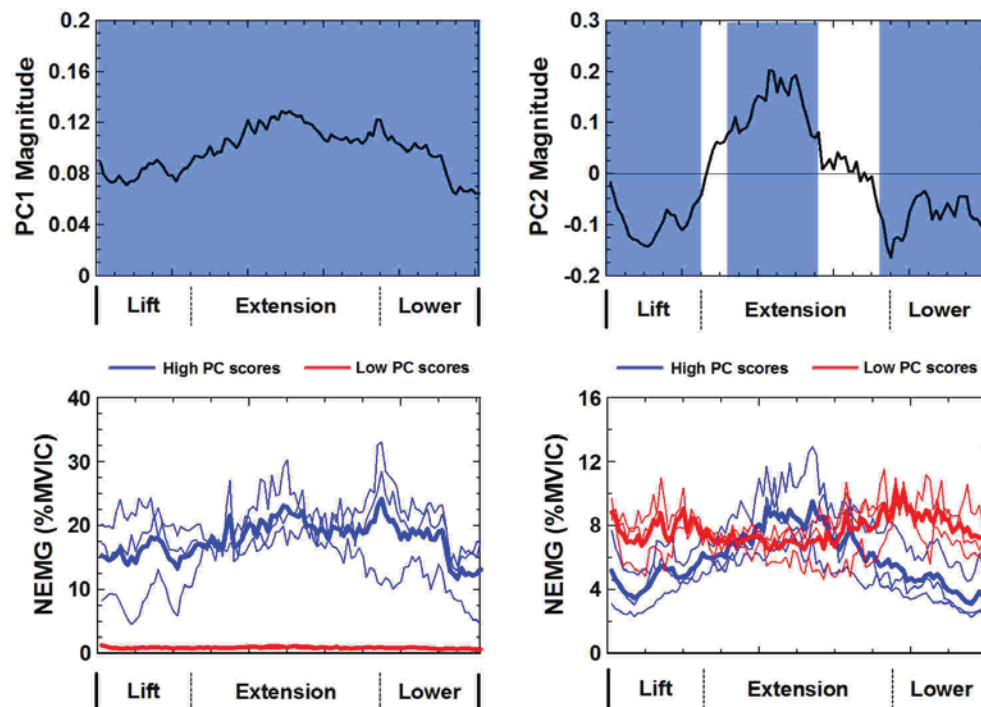

Figure B4: First two PCs for the back muscles. Blue shading on curves A and B indicate portions of the curve where the most variance in the data is described by the PC. A) PC1 which explains 91.4% of the waveform variance; B) PC2 which explains 5.0% of the variance; C-D) High and low scoring curves for PC1-2 with high scores shown in blue and low scores shown in red and the average high and average low scores bolded.

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Table B4: P-values from paired t-tests comparing PC scores on day 1 to PC scores on day 2 are shown in brackets. No significant differences were found between day 1 and day 2.

Muscle	PC1			PC2		
	Day 1	Day 2	P-value	Day 1	Day 2	P-value
RL13	-14.4 ± 12	-12.3 ± 14	0.350	1.1 ± 1.8	0.26 ± 1.2	0.223
LL13	7.8 ± 49	13.4 ± 54	0.277	9.8 ± 18	9.5 ± 18	0.614
RL16	-3.0 ± 7.0	8.2 ± 17	0.086	-2.1 ± 3.8	-3.5 ± 6.1	0.292
LL16	19.6 ± 56	43.2 ± 87	0.170	15.1 ± 26	17.1 ± 33	0.589
RL33	-12.2 ± 19	-13.0 ± 14	0.893	0.81 ± 1.5	1.1 ± 1.9	0.408
LL33	-9.4 ± 22	-13.8 ± 12	0.485	2.3 ± 2.5	2.2 ± 1.6	0.856
RL36	-11.2 ± 18	-13.2 ± 12	0.766	-0.25 ± 1.2	-0.01 ± 2.1	0.781
LL36	-7.0 ± 21	-9.2 ± 12	0.798	2.8 ± 1.9	2.6 ± 1.3	0.792
RL48	8.8 ± 31	5.4 ± 17	0.769	-4.7 ± 4.0	-2.0 ± 2.0	0.343
LL48	18.4 ± 40	16.0 ± 34	0.885	2.4 ± 3.6	1.4 ± 3.7	0.402
RL52	6.5 ± 11	16.2 ± 27	0.376	-3.9 ± 0.8	-4.4 ± 5.4	0.826
LL52	5.6 ± 22	6.6 ± 25	0.925	-2.5 ± 0.6	-2.2 ± 4.3	0.869

Discussion

Three principal patterns from each of the two PCA models run captured the dominant amplitude and temporal features in the EMG waveforms of the 12 abdominal and 12 back muscle sites. The three patterns from the first analysis explain the majority (90.2%) of the variability of the abdominal site EMG waveforms whereas the two patterns from the second analysis explain the majority (96.4%) of the variability of the back muscle site EMG waveforms (Figures B2 & B4).

The overall purpose of this study was to assess the test-retest reliability of a highly standardized surface EMG protocol for analyzing a controlled dynamic leg lifting task and to determine if any specific muscles sites measured show less reliability compared to others. Paired t-tests run on PC scores from day 1 and day 2 for all muscle sites found minimal significant differences between the two days in the 5 PCs analysed for any of the 24 muscle sites. Overall out of the 60 t-tests run only 3 indicated significant differences between the two days (see Table B2 & B3), representing 5% of the tests run. Statistically this is the same as the percentage of tests

(5%) that should return significance by chance as the alpha level of the t-tests was set at 0.05. There seems to be no systematic pattern in which PCs or which muscles showed these few differences. The muscles that showed significant differences in one of the three PC scores between test sessions are RLRA, RIO and LIO.

A possible source of variation between the two test sessions is any training effect seen within the subjects as they become more familiar with the task. Prior to beginning testing during the first session all subjects practiced the TST task at least 10 times in an attempt to eliminate any training effect. Most studies do not include any more practice than this in their protocol, with many including no practice (Dankaerts et al., 2004; Danneels et al., 2001; Lariviere et al., 2000; Lariviere et al., 2002; Mathur, Eng, & MacIntyre, 2005). It may be that novel stability exercises, such as the TST task, are the ones that need the most practice to produce repeatable EMG responses. Danneels found that out of a large subset of exercises examined the only ones that did not show good reliability in amplitude and frequency EMG parameters were exercises that involved balancing, one of which involved a leg extension similar to the TST (Danneels et al., 2001). This justifies the necessity of more practice for more demanding exercises that challenge stabilizer muscles such as the TST task. In the study in Chapter 5 of this thesis the TST task was practiced in the lab during a separate practice session scheduled before the test session and further practiced at home by the subject 3 times prior to testing.

Other studies have reported that reliability of EMG measures during maximal contractions have lower levels of reliability than during submaximal exercises (Lariviere et al., 2000; Mathur et al., 2005; Yang & Winter, 1983). In our protocol maximal contractions were collected during the normalizations and were used to normalize all other EMG data to MVIC. In this protocol the normalization exercises were not practiced however it is possible that since

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maximal contractions have more inherent variability than submaximal contractions practicing them could benefit the reliability of a protocol. Therefore in the study in Chapter 5 of this thesis, protocols for the normalization exercises were practiced prior to testing during a separate practice session. Furthermore, using a series of normalization exercises like in the studies in Chapter 4 and 5, as opposed to a single exercise has been shown to improve the reliability of MVICs (Burden & Bartlett, 1999; Kavcic et al., 2004).

Overall, the results of this study suggest that the reliability of amplitude and temporal characteristics of EMG data analyzed using PCA are acceptable for the protocol reported above. One problem with assessing reliability in the type of homogeneous population tested with this type of data is that there does not exist a well-established, commonly used tool. Paired t-tests, were used to get an overall picture of the reliability of the data and to test for any systematic differences seen in the data between test sessions. It was concluded that the adjustments in the protocol (adding more practice of the experimental task and well as the normalization exercises) could help to improve reliability for future studies.