EVALUATION OF STRENGTH AND FATIGUE IN SUBJECTS WITH AND WITHOUT LOWER BACK PAIN USING SORENSEN PRONE ISOMETRIC BACK EXTENSION

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Evaluation of strength and fatigue in subjects with and without lower back pain using Sørensen prone isometric back extension

By

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ABSTRACT

The Sørensen test of low back endurance is reported to be useful in evaluating potential for developing low back pain (LBP). Sørensen test postures are also used in eliciting maximal voluntary isometric activations (MVIA), which are used as normalization values. The purpose of this study was to investigate spinal musculature fatigue during a Sørensen test and to assess reliability of force and electromyography (EMG) during maximal and sub-maximal conditions in subjects with and without LBP. Repeated measures were taken on twenty male volunteers divided into LBP and control groups. EMG activity was recorded bilaterally from lower abdominal stabilizers (LAS), upper lumbar erector spinae (ULES), lower lumbar erector spinae (LLES), and biceps femoris (BF). Force and EMG during maximal and submaximal (100 – 170% of head, arms and trunk mass {HAT}) efforts were collected. Spectral contents were calculated from EMG and time to failure was recorded, testing fatigue. MVIA forces were lower ($p \leq 0.05$) in LBP vs. controls. Intraclass correlation coefficients for MVIA force, ULES and LLES EMG were excellent in controls ($R > 0.90$), but significantly less in LBP ($R = 0.36$-0.80). BF EMG demonstrated excellent reliability for both groups ($R > 0.90$). Control group EMG was more reliable with maximal efforts, whereas LBP EMG was more reliable with submaximal exertions. Endurance times were not significantly different at any level of HAT. Significant differences in EMG median frequency between groups primarily occurred in the BF. The data indicates that BF contributes to extension, but does not contribute to group differences. Whereas MVIA during a modified Sørensen test is reliable in healthy populations, submaximal efforts using %HAT may be more reliable

for clinical populations; further, the modified Sørensen test did not demonstrate significant differences between groups.

Key Words: Sørensen test, EMG, reliability, between days, fatigue, isometric prone extension, low back pain, trunk muscles

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Michael Holmes, who at the time was an undergraduate student, helped me prepare many of the subjects, participating in data collection. Without Mike's dedication to the project, I would have spent even more time in the laboratory and would have had to search for another model for our photographs.

Dr. Scott MacKinnon introduced me to some statistics, which many mathematicians that I have talked to since didn't understand. Despite his many responsibilities and his busy schedule, he always found time for my questions and concerns. Scott's insights on biomechanics, like the idea to use predicted head arms and trunk mass in conjunction with my ideas for eliciting activations with prone isometric back extension were pivotal to the development of the methodologies used in these papers.

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LIST OF ABBREVIATIONS

aEMG	Average Amplitude of the EMG
BF	Biceps Femoris
CPAFLA	Canadian Physical Activity Fitness & Lifestyle Approach
CSEP	Canadian Society for Exercise Physiology
EMG	Electromyography
HAT	Head arms and Trunk Mass
ICC	Intraclass correlation coefficient
LAS	Lower Abdominal Stabilizers
LBP	Low Back Pain
LLES	Lower Lumbar Erector Spinae
MF	Median Frequency
MVIA	Maximal Voluntary Isometric Activations
SD	Standard Deviation
ULES	Upper Lumbar Erector Spinae

CO-AUTHORSHIP STATEMENT

There have been substantial contributions to this manuscript by a number of key individuals.

i) The concept of using percentages of head arms and trunk mass as a possible means for normalization is attributable to Dr. Scott MacKinnon.

ii) Many of the statistical tools including intraclass correlation coefficients were suggested by Scott MacKinnon. His help in constructing the required tables to calculate these statistics were instrumental for these research papers.

iii) With the input and guidance of Dr. David Behm I prepared the following thesis.

As Chapters 3 and 4 are stand alone papers, it is important to note the authors in the order of their contribution to each paper.

i) Pitcher, M., Behm, D., MacKinnon, S. **Reliability of Electromyographic and Force Measures During Prone Isometric Back Extension in Subjects With and Without Low Back Pain.**

ii) Pitcher M., MacKinnon S., Behm, D. **Neuromuscular fatigue associated with Variations of a Modified Sørensen Test in Subjects With and Without Low Back Pain"**

Chapter 1
Introduction

1.1 INTRODUCTION

Low back pain (LBP) is extremely prevalent in Western society. Although acute low back pain often appears to be self limiting with recovery within a few weeks, (Coste et al. 1994) the recurrence rate of LBP is high (Smedley et al. 1998) with 10-20% of cases becoming chronic (Waddell 2004).

While the direct costs of LBP on the health care system is considerable, it is overshadowed by the extraordinary indirect costs of lost work and decreased productivity. It is estimated that LBP is responsible for the 149 million lost workdays annually in the United States with 102 million of those lost because of work related back pain (Guo et al. 1999). LBP results in significant time lost from work and production loss. Back pain accounts for 25% of all work related injury, but a disproportionate 40% of lost time claims.

With such economic and social impact, it is not surprising that LBP has received much attention clinically and academically to understand, treat and prevent LBP. Identification of factors that would predispose a person to LBP would be helpful, not only in pre-employment workplace screening, but also to offer strategies to prevent and rehabilitate back pain. Methods to assess physical parameters involved in LBP such as strength, flexibility, motor control, range of motion and endurance have all been studied. Weak trunk musculature is considered an important risk factor for development of low back pain (Cady et al. 1979). More definitively, improved endurance of the lumbar

paraspinal muscles seems to confer protection from LBP (Biering-Sørensen 1984; Hultman et al. 1993; Alaranta et al. 1995; Mayer et al. 1985; Nelson et al. 1995; Smidt et al. 1983).

Biering-Sørensen (1984) introduced a prone isometric back extension test that was used clinically to elicit lumbar extensor fatigue. The findings suggested that not only did subjects with LBP have less endurance in the lumbar extensors, but that the results of the test could predict who were likely to develop LBP over the next year. The test developed by Sørensen has been modified, and used in many forms since 1984, not only in its traditional use to clinically evaluate lumbar extensor fatigue, but also in the laboratory setting for normalizing electromyography (EMG) signals.

A modification of the Sørensen test to evaluate back extension endurance is currently used by the Canadian Society for Exercise Physiology (CSEP) in its Canadian Physical Activity Fitness & Lifestyle Approach (CPAFLA) test. Scores are compared to norms and determined to fall into categories ranging from "Needs improvement" to "Excellent" (CSEP 2004). The thoracolumbar spine is a multi-segmental and complex structure and though it has been shown that the Modified Sørensen Tests elicit high muscle activity and fatigue in local lumbar extensors, additional factors may also contribute to the variations in endurance times. Of particular note is the length of time needed to elicit fatigue. In performing a back extension task from 90-180 seconds, there is considerable discomfort which can result in greater opportunity for both physiological and motivational factors to adversely affect the final score. It is currently unclear if testing of higher output leading to shorter testing times is less susceptible to psychological or motivational factors or whether it may yield more accurate delineation between LBP and control subjects.

Furthermore, in the laboratory setting, a Modified Sørensen Test is frequently used to elicit maximal voluntary isometric activations for the purposes of normalizing EMG. It is established that increased pain can negatively affect maximal muscle output and that factors such as muscle inhibition, fear avoidance or decreased motivation are all proposed to play a role (Zedka et al. 1999; Sohn et al. 2000; Babenko et al. 2000; Hirsch et al. 1991; Vlayen et al. 2005). In investigating the lower back with measures of EMG, or force production, it is common to compare groups with and without LBP over a period of time while some type of intervention is performed. However since pain affects maximal output, variations in pain as subjects improve or deteriorate may affect the ability to elicit maximal voluntary isomeric activations (MVIA) and may impact the reliability over time.

There are no studies to our knowledge that have examined both the reliability and validity (ability to differentiate between controls and low back pain sufferers) with a modified Sørensen test using submaximal and maximal loads.

1.2 PURPOSE AND HYPOTHESES:

The purpose of this study was to better understand the physiology associated with the Sørensen posture both in its traditional use as an endurance test, and in its application for eliciting values for normalizing EMG signals for biomechanical modeling. The results of this investigation may help to develop more reliable and effective methods of testing for the lower back in both the clinical and laboratory settings.

Four experimental hypotheses are proposed:

H1: EMG and force values elicited with MVIA will be less reliable for the LBP group than control subjects across four testing sessions.

H2: EMG values will be more reliable in the LBP group when the test is performed at set submaximal values with visual feedback.

H3: Endurance time will be higher in the control group as compared to the LBP group.

H4: Differences in endurance time between LBP and control groups will be larger with higher muscle output demands

1.4 ASSUMPTIONS

The following assumptions were made in this study:

1. Subjects' mass was recorded on the initial day and based on anthropometric data the head, arms and trunk weight was calculated (HAT). This value and pre-set percentages of this value were used across the four testing sessions. It is assumed that the subjects mass did not significantly change over the course of the testing.
2. All subjects were prepared for electrode placement in the same manner on each testing session.
3. All subjects were given systematic and consistent motivational cues during MVIA and endurance testing.
4. Testing procedures did not have a physiological effect to train the lumbar extensors and improve performance over the course of four sessions.

1.5 LIMITATIONS

The following limitations are recognized in this study:

1. The study was relatively small with a control group (n=10) and a LBP group (n=10).
2. Although rest time were designed for full muscle recovery, testing of reliability and fatigue, and was consistent between groups, it is possible it may result in lower overall endurance times in each group.
3. With repeated tasks across days, there is potential for effects of motor learning and improved performance across sessions.

4. For reasons of comfort and convenience we used less invasive surface electrodes, while ground electrodes helped minimize cross talk, the nature of the architecture of the spinal muscles make elimination impossible

1.6 LAYOUT OF THE THESIS

This thesis is laid out in a non-traditional manner with presentation of two individual research papers investigating two difference aspects of lumbar extension that together help address the overall purpose of the thesis. The literature review of Chapter 2 will delve into background that encompasses the topics of both papers as it relates to the overall theme. In Chapter 3, the first paper entitled *"Reliability of Electromyographic and Force Measures During Prone Isometric Back Extension in Subjects With and Without Low Back Pain"* investigates the reliability of the modified Sørensen posture as a tool in eliciting MVIA in both low back pain and controls groups and proposes a novel technique. In Chapter 4, the second paper entitled *"Neuromuscular fatigue associated with Variations of a Modified Sørensen Test in Subjects With and Without Low Back Pain"* addresses the questions associated with fatigue of the lumbar paraspinal muscles. It investigates if a simple Modified Sørensen test is sufficient to result in different endurance times between control and LBP groups. Chapter 5 presents overall conclusions and discussion based on the findings of the individual papers.

Chapter 2
Review of Literature

2.1 INTRODUCTION

Low back pain (LBP) is a pervasive problem in modern societies and is reported to affect 60-85% of the population at some point in their lifetime. Approximately 8% of the working age population experience disabling LBP in a given year (Straus 2002). These problems are reported to become chronic 10-20% of the time (Waddell 2004). Although there is a reported high rate of recovery during the first few weeks (Coste et al. 1994) the recurrence rate is high (Smedley et al. 1998). Musculoskeletal disorders impose significant direct cost that burden the health care system in North America. The estimated total cost of musculoskeletal disorders in Canada in 1994 was 25.6 billion dollars. Twenty nine percent (29%) of that were direct medical costs, and the remaining 71% incurred through indirect losses. The largest proportion of the overall cost (8.1 billion) was a result of lower back and spine disorders. The vast majority (92%) of the overall financial burden of lower back and spine disorders were associated with the indirect costs (Coyte et al. 1998). In the United States 24.3 billion was spent in 1990 in the United States on direct costs of LBP alone (Frymoyer and Cats-Baril 1991). In industrialized countries direct and indirect costs can amount to 0.8% - 2.1% of the gross domestic product (Hemmila 2002). It is estimated that LBP is responsible for the loss of 149 million workdays annually in the United States with 102 million of those workdays lost because of work-related back pain (Guo et al. 1999).

The costs of medical treatment for all work-related and non work-related back pain was at least $13 billion in 1990 with an estimated increase of 7% per year (Straus 2002). At the workplace, back problems are the single most costly injury in terms of contribution to the total overall cost. The National Safety Council (1993) reported back cases comprised 24% of the US workers compensation claims and 31 % of the costs.

LBP results in significant time lost from work and loss of production. Back pain accounts for 25% of all work-related injury, but a disproportionate 40% of lost time claims. Further, workers with back pain are less likely to return to work than with other injuries (Johnson 1998). Frank et al. (1996) describes a timeline of work related back pain and reports the likelihood of returning to work during each phase. There are three phases: acute, subacute and chronic. The acute phase spans from onset to 3-4 weeks and 50% of cases return to work during this period. During the subacute phase, described as 4-12 weeks after onset, an additional 30% return to work. Cases that become chronic, beyond 12 weeks, have less likelihood to return to work. These cases are responsible for the majority of indirect costs. Costs distributed over time as reported by Williams et al. (1998) is consistent with the natural timeline of return to work previously described by Frank et al. (1996) and reveals a disproportionate distribution of costs with chronic cases responsible for 42% of direct medical cost, 54% of indirect indemnity cost and 52% of the cost overall. Further, 25% of workers that lost time over 3 months accounted for over 75% of the costs. Data from the Quebec Workers Compensation Board reported in the Quebec Task Force reveals an even higher disproportionate distribution of cost with 7-10% of back cases accounting for more than 70% of total health care and indemnity costs (Spitzer et al. 1987).

In summary, LBP accounts for the majority of musculoskeletal pain and is responsible for a disproportionate amount of the societal cost. The longer the problem persists and the more chronic it becomes, the less likely the individual will return to work, and the higher the economic burden becomes. Armed with such statistics, it is clear that one of the key roles of the primary care physician that deals with musculoskeletal disorders, especially back pain, should be to identify risk factors for developing chronicity and to prevent transition of the problem into a chronic stage.

2.2 CAUSES OF TISSUE INJURY AND THE ROLE OF SPINAL STABILITY:

It has been reported that over 90% of LBP cases resolve within 6 weeks (Carey et al. 1995). However, Croft et al. (1998) has argued that this number is not accurate as patients with back pain simply stop seeking care from their primary physician. Of the 463 patients who consulted their primary care physician with a new episode of LBP, 59% had only a single consultation with only 32% additional consultations within 3 months. After a three month follow up, only 21% had completely recovered and at the 12 month follow up only 25% had completely recovered. The authors concluded that while 90% of patients with LBP will have stopped consulting after three months most continue to be symptomatic after a year (Croft et al. 1998).

Much effort is directed towards establishing specific organic diagnoses such as a herniated nucleus pulposus, annular tear or facet irritation. The general failure of such efforts has led to the assumption that 85% of back pain has no apparent cause (White and Gordon 1982). Indeed pathologic causes cannot be found for many instances of LBP (Papageorgiou et al. 1996) and only 15% of patients have a definitive diagnosis (Kelsey

1982) Such ideas are supported by studies that found minor (45-50%) or absent (35-41%) degenerative changes in patients that were referred for radiology studies to evaluate LBP. (Hollingsworth 2002). Although, it is well established that there is poor correlation between pain and degenerative changed observed on radiographs or magnetic resonance imaging (Lawrence 1977; Jensen et al. 1994).

2.2.1 Mechanism of Tissue Injury:

Injury to tissue occurs when loads placed upon the tissue exceed the failure tolerance. There are essentially three mechanisms of tissue injury: 1) Single load failure, 2) Repetitive or cumulative load failure and 3) Continuous load failure (McGill 2002)

Typically tissues that are loaded, are done so, well below failure tolerance and thus exhibit a margin of safety. Single load failure is probably the most familiar mechanism, but not necessarily the most common. Such injury occurs when a load exceeding the failure tolerance is applied in a singular event resulting in failure of the tissue (McGill 2002). Perhaps more commonly is failure that is a result of cumulative effects of tissue loading, where multiple sub-failure loads result in tissue hysteresis, negatively affecting the integrity of the tissue and decreasing the failure tolerance. The result, is a decreasing margin of safety until at some point in the loading cycle, a sub-failure load exceeds the failure tolerance and tissue injury occurs (McGill 2002). Although the loads that are performed in repetitive tasks are well below what is considered "safe", the cumulative effect is to weaken the involved tissue bringing the tissue closer to failure. Repetitive activities requiring constant loading of a joint is a known risk factor for development of cumulative trauma disorders (Norman et al. 1998; Punnett et al. 1991). Tissue can also be

damaged with application of sustained sub-failure loads. Constant and repetitive loading of viscoelastic materials such as human tissue causes hysteresis resulting in deformation and creep in the tissue. If sustained, the loading can cause sufficient energy loss to the tissue reducing its margin of safety, to where it reaches the point of breaking strain and exceeds the failure tolerance of the tissue (McGill 2002). Solomonow et al. (2003) discovered that periods of static flexion in the lumbar spine resulted in creep of the viscoelastic tissues which interestingly, did not return to pre-existing levels after an equal period of rest. When loaded repetitively there was additional creep created in the tissues. (Solomonow et al. 2003) Creep of viscoelastic tissues can also result in ligament microdamage (Woo 1999). Solomonow et al. (2003) demonstrated that acute inflammation was evident in ligaments 2 hours after a 20 minute period of static spinal flexion. It is likely that if creep does not recover then tissue tolerance is reduced and there is less resistance to injury (McGill 2002). McGill suggests that the intersection of load and risk creates a U-shaped function and argues that moderate loading is beneficial to tissue health and results in adaptive changes and increased tolerance to failure, and either too little or too much load will result in weakening of the tissue and increased risk to injury (McGill 2002).

2.2.2 Spinal Stability and Lower Back Pain:

Functional instability has received much attention for its role in low back dysfunction (O'Sullivan et al. 1997; Cholewicki and McGill 1996). The osseoligamentous spine is well known to be unstable at loads significantly less than body weight and must

rely on external forces to confer stability even under minimal loading (Cholewicki et al. 1997).

Adequate biomechanical and structural integrity coupled with proper neuromuscular control are needed for spinal stability. Panjabi et al. (1991) have presented a model of spinal stability with interaction of active, passive and control subsystems. Stability of the spine is maintained by the integrity of the passive viscoelastic structures (McGill and Norman 1986) and through increase in spinal stiffness by adequate activation of muscles (Granata and Marras 1995; Lavender et al. 1992; McGill 1991; McGill and Norman 1986). There is also evidence that supports the integral role of the motor control system (Cholewicki and McGill 1992; O'Sullivan et al. 2003) and that this control is largely involuntary (Granata et al. 2001).

A joint with a given range of motion will have more deformation of potentially pain sensitive structures near its end range of motion. Further, the relationship between the amount of deformation of joint structures and load is non linear, with significantly less deformation near the mid-range of a joint. A given joint has potential to move within its active range of motion and when loaded by external forces into its passive range of motion. Spinal stability as conceptualized by Panjabi may be therefore interpreted as relative maintenance of a joint within its mid range or "neutral zone", or in dynamic movements as preventing excessive motion into areas of higher deformation and thus potentially reducing the likelihood of injury. Neutral zone maintenance and spinal stability are achieved through activity of three interactive systems: the passive, active and neural control subsystems (Panjabi 1992).

A deficit in one or more of these subsystems may contribute to reduced spinal stability and possible segmental injury (Granata and Orishimo 2001; Oxland and Panjabi 1992). To maintain adequate levels of stability when a subsystem is compromised, trunk muscles compensate by altering their normal activation pattern (Radebold et al. 2000; Panjabi 1992). Alterations in normal activation patterns interfere with ideal muscle anticipatory contraction and proper co-contraction patterns (Hodges 2001; Hodges et al. 2001).

Ligaments are not engaged until end range of motion; after some injury has potentially already occurred (Panjabi 1992). Muscles however may be engaged earlier preventing excursion beyond the neutral zone and therefore alterations in muscle activity in instances of LBP is of particular importance. As part of the active stability subsystem, muscles can compensate for instability from deficits in other areas by increasing muscle stiffness through muscle activation, ultimately increasing spinal stiffness (Panjabi 1992; Cholewicki and McGill 1996; Gardner-Morse, Stokes et al. 1995; Granata and Marras 2000). Cholewicki and McGill (1996) have also suggested that inherent muscle stiffness may contribute to spinal stability prior to engaging neural control mechanisms. Decreases in muscle stiffness that result from fatigue, degenerative changes and injury may contribute to instability (Gardner-Morse et al. 1995).

Muscles can act locally or globally to offer spinal stability. Differing roles have been proposed for the deeper segmental muscles and the more superficial multisegmental muscles (Panjabi et al. 1989; Bergmark 1989). The smaller muscles such as the intertransversarii and interspinalis have high concentration of muscle spindles and have small moment arms, (Bogduk and Bogduk 1997; Peck et al. 1984) being located so close

to the centre of rotation and therefore are theorized to act as force transducers providing proprioceptive input into the control subsystem and may play a role in adaptive protection (Crisco and Panjabi 1991). Larger intersegmental muscles such as the multifidus has been shown to be able to increase spinal stiffness (Cholewicki and McGill 1992) necessary for spinal stability (Panjabi et al. 1989; Wilke et al. 1995; Crisco and Panjabi 1991). Others suggest that the larger global muscles also play a significant role by increasing stiffness across the larger areas of the spinal column and not just segmentally over a few levels (Panjabi et al. 1989).

Unlike Crisco and Panjabi's work that highlighted the importance of the smaller local muscles for their role in stability (Crisco and Panjabi 1991), McGill (2004) found that global muscles offer better mechanical advantage and are better able to stabilize the spine than the smaller local stabilizers. Differences are possibly due to divergent modeling techniques. The previous study used a straight elastic column with motion limited to the frontal plane whereas the model created by McGill (2004) used a lordotic curve that allowed 18 degrees of freedom and the intersegmental muscles to follow the curvature of the spine (McGill 2004).

From biomechanical analysis it is clear that no one single muscle possesses a dominant responsibility for lumbar stability (Cholewicki and McGill 1996; Cholewicki and VanVliet 2002). In fact, data from McGill (2004) suggest that each muscle plays multiple roles at once and their roles depend on the demand on the spine at a given moment. Because no single muscle confers stability to the osseo-ligamentous spine, a motor control strategy of coordination between muscles is required to prevent buckling and injury of the spine. Cholewicki and McGill (1996) argue that one of the most

important ways that neuromuscular control contributes to spinal stability in the lumbar spine is through co-contraction of agonistic muscles. Co-contraction has been shown to occur during many daily activities (Marras and Mirka 1990) and is more prominent with sudden or unexpected loading of the spine (Marras et al. 1987; Lavender et al. 1989). Even momentary lack of motor control could potentially contribute to injury. Cholewicki and McGill (1992) observed the occurrence of an injury in a power lifter as he performed a task while under videofluroscopy. They observed that there was a momentary increase in relative flexion that did not occur in any of the other unaffected levels. These observations highlight the importance not only of the integral role of motor control, but also illustrate the consequences of loading spinal structures outside of Panjabi's neutral zone.

2.2.3 Acute and chronic back pain:

For many back pain sufferers frequent episodes of LBP only results in temporary disability. Unfortunately 10-20% of acute LBP may become chronic (Waddell 2004). The most common means of categorizing acute and chronic LBP is through use of timelines from the original episode. Many authors suggest acute LBP is pain within 1-14 days on the precipitating event (Fordyce et al. 1986; Philips and Grant 1991; Malmivaara et al. 1995). Carey et al. (1995) liberally extends the definition to include cases that persist to 10 weeks. Van Tulder et al. (1997) describes subacute pain as episodes of pain that extend from 5-7 weeks but no longer than 3 months. Merskey (1994) defined chronic pain as an episode of pain that persists beyond 3 months. Skouen et al. (2002) proposes a nicely defined timeline as follows: acute <28days; subacute 4-12 weeks; and chronic

lasting more than 12 weeks. While consistent terms of reference are important in discussing the progression of LBP, framing acute, subacute and chronic LBP solely in terms of progression of time may be inappropriate. There are many possible explanations for the onset of acute and chronic LBP, however the nature of the transition that must occur between acute to chronic need is not well understood (Fransen et al. 2002). It is argued that chronic LBP is not simply the continuation of acute LBP, but is hallmarked by changes in strength, endurance, pain perception and motor control.

2.3 PHYSIOLOGICAL CHANGES ACCOMPANYING LOWER BACK PAIN:

2.3.1 Motor Control and Morphological Changes with Low Back Pain:

Afferent input from receptors in muscle, joint and skin aid in providing proprioceptive input (Gandevia et al. 1992). Injury in peripheral joints has been shown to result in kinesthetic deficit (Forwell and Carnahan 1996; Smith and Brunolli 1989). Proprioceptive deficit may delay normal reaction of the neuromuscular component of a joint's stability system, leading to inappropriate segmental motion or loading, leading to pain and tissue injury (Forwell and Carnahan 1996). There is conflicting reports in the literature on repositioning deficits in subjects with LBP with some authors reporting discrepancy between controls and subjects with LBP (Brumagne et al. 1999), while others do not (Newcomer et al. 2000; Lam, Jull et al. 1999). However subjects that are specifically diagnosed with clinical segmental instability were unable to resume a neutral lumbar posture while seated (O'Sullivan et al. 2003). Individuals with LBP have been found to use different motor control strategies than those without pain. (Grabiner et al. 1992; Taimela et al. 1999; Luoto et al. 1998; Luoto et al. 1999). Functional changes

(Hodges and Richardson 1996; Hodges et al. 1996; O'Sullivan et al. 1997) and alterations in patterning of muscle activation associated with chronic LBP have been observed in the deep abdominal muscles such as the transverse abdominis (O'Sullivan et al. 1997; Richardson and Jull 1995). In recent years much attention has also been directed towards the characteristics of the lumbar multifidus for its role in stability and its potential protective role in LBP. The multifidus is generally considered an important stabilizer of the lumbosacral spine (Danneels et al. 2000; Goel et al. 1993; Kaigle et al. 1995; McGill 1991; Panjabi 1992; Panjabi 1992b; Wilke, et al. 1995), but is also important in assisting in extension of the lumbar spine (Panjabi 1992; Panjabi 1992; Ng and Richardson 1996; Ng et al. 1997).

Using diagnostic ultrasound, Hides et al. (1994) investigated the cross sectional area of the multifidus in subjects with acute unilateral LBP (n=26) and controls (n=51). Subjects with acute or subacute LBP demonstrated most marked decreases in cross sectional area of the multifidus well localized segmentally at the injured level and ipsilateral to the side of symptoms. Because of the specificity of atrophy segmentally and asymmetrically, the authors proposed that disuse atrophy or spinal reflex inhibition was an unlikely cause, and that inhibition from nociception or perceived pain via a long loop reflex is a more viable explanation. Interestingly these atrophic changes and decreases in cross sectional area did not spontaneously recover after symptoms subsided (Hides, Richardson et al. 1996). In addition to morphological changes, functional differences between multifidus and iliocostalis lumborum have been observed through electromyography (EMG) (Ng et al. 2001; Ng et al. 1997). During the Sørensen test, the multifidus demonstrates more electromyographic activity (Ng et al. 1997), and in LBP

groups faster rates of fatigue than the iliocostalis lumborum (Ng et al. 1997; Ng and Richardson 1996). This suggests the importance of the multifidus in resisting flexion in the sagittal plane and that the iliocostalis lumborum acts not only in extension, but also in lateral flexion. It has been observed that during strength exercises with extension there is decreased EMG activity in both the multifidus and the iliocostalis lumborum in LBP patients compared to healthy controls (Danneels et al. 2002). The authors postulated that the selective atrophy of the lumbar multifidus with LBP and its established role in extension and stability of the lumbosacral spine, may contribute to the high recurrence rate of LBP (Danneels et al. 2002). The authors also compared the normalized EMG activity of the multifidus and the iliocostalis thoracic during coordination, stabilization and strength exercises in subacute (n=74) and chronic LBP subjects (n=51) and healthy controls (n=77). The chronic LBP patients displayed significantly lower (P=0.013) EMG activity of the multifidus during coordination exercises, and in both multifidus (P=0.017) iliocostalis thoracis (P=0.003) during a strength exercise. There was no significant difference however in normalized EMG activity of the two muscles in the stabilization exercises (Danneels et al. 2002). Daneels et al. (2002) suggests that the decreased activity of the multifidus during coordination exercises indicates that, back pain patients may have a decreased ability to recruit the multifidus to obtain a neutral lordosis which is known to be a protective mechanism for reducing low back injury (McGill 2002).

2.3.2 Pain Inhibition and Psychological Factors:

Maximal muscle activation, frequently used as part of normalization procedure for EMG signals, is known to be adversely affected by pain and neuromuscular inhibition

(Zedka et al. 1999; Sohn et al. 2000; Babenko et al. 2000). Non-physiologic factors such as motivation (Hirsch et al. 1991), and fear of re-injury (Hirsch et al. 1991; Vlayen et al. 2005) are also noted to play a role.

Decreased muscle activation moving from a standing flexed posture to standing extension has been observed in subjects with LBP (Zedka et al. 1999). The author ssuggests that decreased muscle activation may be associated with either voluntary avoidance or from nociceptive input resulting in segmental inhibition of the descending voluntary α-motoneurones. In this study, deep pain changed descending motor commands, but had little effect on segmental stretch reflexes. Lund et al. (Lund et al. 1983) has described such a pain-adaptation model of afferent nociceptive input facilitating inhibitory pathways. Several animal studies where voluntary drive was overridden with descending electrical stimulation and muscle modulation persisted, offers support that voluntary drive alone is not solely responsible for observed muscle activation changes (Westberg et al. 1997).

A defining characteristic of chronic LBP is decreased trunk strength and endurance associated with a cyclical pattern of deconditioning through pain, avoidance and inactivity (Mayer and Gatchel 1988). Patients with chronic LBP are shown to have decreased trunk strength and endurance as compared to healthy counterparts (Mayer et al. 1985; Nelson et al. 1995; Smidt et al. 1983). Further, weak trunk musculature is considered an important risk factor for lower back trouble (Cady et al. 1979). Conversely, individuals with increased muscle strength, endurance and cardiovascular fitness have fewer spinal problems (Cady et al. 1979; Nelson et al. 1995).

2.4 MEASUREMENT, RELIABILITY AND VALIDITY OF MEASURES OF STRENGTH, AND FATIGUE IN THE LUMBAR SPINE:

There have been investigations into many methods to assess physical parameters involved in LBP. Strength, flexibility, motor control, range of motion and endurance have all been studied. Weak trunk musculature is considered an important risk factor for lower back trouble (Cady et al. 1979). Fatigue has also been shown to correlate well with LBP (Biering-Sørensen 1984; Hultman et al. 1993; Alaranta et al. 1995; Mayer et al. 1985; Nelson et al. 1995; Smidt et al. 1983). Endurance of the paraspinal muscles are thought to be of particular importance due to action of counteracting the constant effects of gravity (Kalimo et al. 1989; Roy, et al. 1989).

2.4.1 Fatigue and Endurance:

Neuromuscular fatigue is a complex process, modulated by a number of factors. Neuromuscular fatigue has been defined as "any exercise-induced reduction in the maximal voluntary force or power output" (Vollestad 1997), however it is clear that fatigue is not a benchmark that is reached but a process that begins with the onset of muscle contraction (Philip and Gardiner 2001). A key feature that highlights this point is the progressive decrease in firing frequency of motor neurons. Decreases in frequency are frequently used as an index to quantify neuromuscular fatigue. The mechanisms of fatigue that ultimately result in decreases in firing frequency may be of particular importance when looking at LBP and potential treatment options. Fatigue may occur both centrally and peripherally. Discharge patterns of spinal motoneurons are regulated by output of the motor cortex reflecting volitional control as well as afferent reflex activity. Decreased excitation from the motor cortex can decrease motoneuron output. In addition

to decreased supraspinal output, motoneuron inhibition and decreased firing rates may occur through spinally mediated reflex pathways. Because the decreased firing frequency is matched to optimize force output as fatigue progresses it is theorized that feedback closely regulates this process (Bigland-Ritchie and Woods 1984).

Peripherally there are a number of additional mechanisms that contribute to fatigue. Some mechanisms, such as metabolite accumulation, may directly affect the firing frequency, while others have a more indirect action. Increased metabolite concentrations such as extracellular hydrogen and potassium negatively affect the sodium potassium pump (Philip and Gardiner 2001). In the motor neuron, decreased pump activity may lead to decreased nerve conduction velocity directly affecting firing frequency. At the muscle itself, decreased pump activity will negatively impact the Ca2+ kinetics, thus lengthening relaxation times. ATP production is required for all aspects of cross bridge cycling including bonding of Ca2+ - troponin, troponin – tropomyosin, and actin – myosin, as well as for the Ca2+ and Na+/K+ pumps. Decreased firing frequencies are thought to occur to match the slowing cross bridge cycling and slowing Ca2+ kinetics to optimize maintenance of maximal force (Bigland-Ritchie and Woods 1984). This phenomenon of decreased firing rate in response to fatigue, coined "muscle wisdom" (Marsden et al. 1983), is further supported by animal models that demonstrate maximal force outputs optimized with electrical stimulation using continually decreasing stimulation frequencies (Bigland-Ritchie and Woods 1984).

With isometric muscle contraction at sufficient intensity (approximately 30% of MVC) internal muscle pressure may increase, collapsing capillaries and compromising the muscles' blood supply. Yoshitake et al. (2001) demonstrate that restriction of blood

flow from increased intramuscular pressure is one of the most important factors in lumbar muscle fatigue. If such ischemic conditions occur during sustained muscle contractions, reliance on anaerobic metabolism is more pronounced and associated metabolites may accumulate. Substances commonly associated with muscle fatigue such as lactic acid, (Jovanovic et al. 1990; Djupsjobacka et al. 1995) bradykinin, (Jovanovic et al. 1990; Djupsjobacka et al. 1995), arachadonic acid (Rotto and Kaufman 1988), and potassium chloride (Djupsjobacka et al. 1995; Rybicki et al. 1985; Kaufman and Rybicki 1987; Hirche et al. 1980; Vyskocil et al. 1983) have been shown to increase afferent discharge.

2.4.2 Fatigue Measures:

Fatigue and endurance is evaluated in the lumbar spine most commonly with isometric, isotonic and isokinetic testing (Moreau et al. 2001). Of the three, isometric testing is most frequently used in a clinical setting to evaluate the lumbar spine and has most recently been used in Canadian Society for Exercise Physiology's (CSEP) Canadian Physical Activity Fitness & Lifestyle Approach (CPAFLA) fitness appraisal (CSEP 2004.). There are various methods of evaluating lumbar spine fatigue using isometric extension testing. Some of the testing methods reported in the literature include: Prone Isometric Chest Extension, Seated Dynomometry, Pulling or Nicolaisen Jørgensen Test and the Beiring-Sørensen and Modified Beiring-Sørensen Tests.

2.4.3 Prone Isometric Chest Raise:

In the prone isometric chest raise, subjects lie prone and raise their chest off the table. The duration of time that the chest is held off the table is recorded. One study had

subjects lie with a pillow placed under the abdomen, and the subject was instructed to keep their buttocks tightened and their cervical spine flexed. The subjects in this study were asked to maintain the extended position up to a maximum of 300s. The authors report a mean endurance time of 208.2s in healthy males (n=37) and 85.1s in male subjects with chronic LBP (n=40). Unlike other studies examining fatigue in males and females, the authors report shorter endurance times for both healthy females 128.2s (n=53) and females with chronic LBP 70s (n=60) (Ito et al. 1996). In a study of similar design, the Canadian Back Institute had subjects perform the chest raise without a pillow and with their hand placed at their temples. Subjects would be asked to hold this position as long as possible while the endurance times were recorded. This study demonstrated significantly different endurance times for different age groups (McIntosh 1998). Ito et al. (Ito et al. 1996) report test-retest r values of 0.97 for healthy males and 0.94 for healthy females versus r values of 0.93 for males with chronic LBP and actually better (r=0.95) for females with chronic LBP. Additionally they report intraclass correlation coefficient (ICC) values of 0.97 for both healthy males and females and 0.93 for both males and females with chronic LBP. McIntosh et al. (1998) reports a test-retest r values of 0.633 for the prone isometric chest raise however no ICCs were reported.

2.4.4 Sitting Dynamometer

Sitting dynamometry is not widely used for testing isometric fatigue. Using a Biodex medical dynamometer, Van Dieën and Heijblom (1996) had subjects perform a maximal isometric voluntary contractions in order to gauge 50% of maximum. Subjects were then asked to perform sustained exertion at this intensity until the force fell below

90% of the target force. They report poor test-retest reliability across days (ICC=0.54), but excellent reliability (ICC=0.94) when test was performed within 5 minutes of the original test.

2.4.5 Pulling or Nicholaisen Jørgensen Test

The Pulling Test or Nicolaisen Jørgensen test is performed while standing, having the lower body braced while the upper body exerts an extension force against a dynamometer. Unlike the Beiring Sørensen test, the Nicolaisen Jørgensen test requires the subject to first perform a baseline MVIA. After sufficient rest the subject then maintains 60% of the predetermined MVIA with use of a visual feedback device. Authors report mean pull times of both 54s (n=24) (Nicolaisen and Jørgensen 1985; Jørgensen and Nicolaisen 1987), and 52s (n=53) (Hultman et al. 1993) in healthy males and 80s (n=8) (Nicolaisen and Jørgensen 1985) and 73s (n=23) (Jørgensen and Nicolaisen 1987) in healthy women. Consistent with findings of prone isometric extension, (Kankaanpaa et al. 1998; Mannion and Dolan 1994; Mannion et al. 1997) women had longer endurance times than men (Nicolaisen and Jørgensen 1985). Because this test requires maximal efforts to generate a percentage of maximum, there has been suggestion that it may not be appropriate for acute LBP populations (Moffroid et al. 1993). The Nicolaisen Jørgensen test has been purported by the authors to have better general test–retest reliability than that of the Beiring Sørensen test, however they use an unspecified statistic to report their test-retest reliability (Jørgensen and Nicolaisen 1986). While such conclusions are criticized as premature, (Moreau, Green et al. 2001), the Nicolaisen Jørgensen test has

demonstrated better median frequency slope reliability than the Beiring Sørensen test (Koumantakis et al. 2001).

2.4.6 Beiring Sørensen Test

The most widely reported endurance test in the literature is the Biering–Sørensen test (Moreau et al. 2001). For the Sørensen test, subjects would have their lower body secured to a table with three wide straps and positioned such that their anterior superior iliac spine was at the edge of the table. Subjects are asked to hold their unsupported head arms and trunk parallel to the floor until exhaustion to a maximum of 240-300 seconds. Administration of the Sørensen Test is varied in the literature, including differences in arm position, number of straps (or no straps) and conclusion criteria. These variations have been grouped together as Modified Sørensen tests (Moreau, Green et al. 2001). This test is generally considered safe for both healthy and clinical populations (Biering-Sørensen 1984; Moffroid 1997; Nordin et al. 1987; Alaranta et al. 1994; Peltonen et al. 1998; Alaranta et al. 1995; Mannion and Dolan 1994). Forces required to maintain a horizontal position are well below MVIA in healthy populations, (Mayer et al. 1995; Moffroid, et al. 1993; Jørgensen and Nicolaisen 1986) but may rise to as much as 85% in a patient with chronic LBP (Hultman et al. 1993). It has been suggested by one author that performance of MVIA in patients with LBP could compromise safety (Moffroid et al. 1993).

There is considerable range of mean endurance times reported for the Sørensen test in the literature ranging from 84s to 180s in healthy males (Jørgensen and Nicolaisen 1986; Sparto et al. 1997; Jørgensen and Nicolaisen 1987; Biering-Sørensen 1984;

Kankaanpaa et al. 1998; Mannion and Dolan 1994; Nicolaisen and Jørgensen 1985; Hultman et al. 1993) and from 142s to 220s in healthy females (Jørgensen and Nicolaisen 1986; Biering-Sørensen 1984; Kankaanpaa et al. 1998; Mannion and Dolan 1994; Nicolaisen and Jørgensen 1985; Mannion et al. 1997; Nordin et al. 1987; Moffroid et al. 1993). For males with LBP mean endurance times range from 80s-194s (Jørgensen and Nicolaisen 1987; Biering-Sørensen 1984; Nicolaisen and Jørgensen 1985; Hultman et al. 1993).

The reported r values for test-retest reliability in healthy subjects ranged from 0.20 – 0.91 (Jørgensen and Nicolaisen 1986; Mayer et al. 1995; Hyytiainen et al. 1991; Moffroid et al. 1993; Alaranta et al. 1994) In studies that reported reliability with ICCs, the values ranged from 0.54 - 0.99 (Mannion and Dolan 1994; Mannion et al. 1997; Simmonds et al. 1998; Ito et al. 1996). The reliability was higher in more physically active subjects (0.82 to 0.96) (Moffroid et al. 1994; Simmonds et al. 1998) but reported to be lower in inactive LBP patients (ICC=0.39) (Jørgensen and Nicolaisen 1986). Good correlation to LBP has been reported in association with low Beiring Sørensen scores. (Hultman et al. 1993; Alaranta et al. 1995)

Using Borg scales of perceived exertion, Dedering et al. (2000) revealed good reliability for endurance time (ICC=0.89), for initial and end median frequencies (ICC=0.75 - 0.89) and for median frequencies at Borg ratings of three (ICC 0.63 - 0.88), five (ICC 0.62 - 0.84) and seven (ICC 0.67 - 0. 87). In this study there was more agreement of the Borg ratings between the second and third testing session suggesting that a practice session, or orientation is needed.

The Sørensen test is used as a measure of low back function measuring overall lower back fatigue, although it has been argued that the activity of the biceps femoris and hip extensors contribute to endurance times (Kankaanpaa et al. 1998) and that there is significant correlation between Sørensen endurance times and EMG median frequency slopes of the biceps femoris (Moffroid et al. 1994; Moffroid 1997). EMG fatigue analysis also suggests that the gluteus maximus muscles are more fatigable in chronic LBP patients than in healthy control subjects during a sustained back extension endurance test (Kankaanpaa et al. 1998). Recent research of McKeon et al. (2001) also demonstrate that with prone isometric back extension, females with LBP used this muscle more than healthy females. Ng et al. (1997) however, demonstrated more activity in the multifidus than the iliocostalis lumborum during Sørensen testing. The multifidus fatigues at a faster rate than the iliocostalis lumborum during this test demonstrating a higher initial median frequency and normalized MF slope (Ng et al. 1997). Ng et al. (1996) suggests that the Modified Sørensen test with the use of EMG power spectral analysis may be a reliable method to measure the fatigue rate of the back muscles if cross-talk is minimized and adds that measuring the fatigue rate of the multifidus may be a useful clinical measure. Van Dieën et al. (1993) observed that the multifidus muscle at the L5 level appeared to show the most consistent changes of the EMG power spectrum as a consequence of fatigue.

2.4.7 Normalization of EMG

For meaningful evaluation of muscle activity with EMG, either between muscles or between conditions, some form of normalization is required. There are inherent

limitations to normalization techniques. Subcutaneous fat, skin impedance and electrode placement are of concern and could result in variability within and between subjects (Lehman et al. 2001; Mirka et al. 1997). Maximal muscle activations are most widely used as normalization references and are generally considered more reliable than submaximal activations (Anderson and Sweetman 1976). Nicholaison and Jørgensen (1985) described a submaximal technique with their upright pull test using 60% of a Maximal Voluntary Isometric Activation (MVIA) However the subjects had to perform at least one MVIA in order to determine the target of 60%. An entirely submaximal normalization technique was described by Marras and Davis (2001) which involved multiple exertions and the assumption of linear EMG force relationship to create a line to predict the maximal exertion.

2.4.8 Prone Back Extension for Normalization:

Various methods exist for eliciting MVIA in the trunk extensor musculature. Although there are reports of maximal isometric back extension performed with sitting (Taimela 1998; van Dieen and Heijblom 1996; Robinson et al. 1991) and standing dynamometry (Jørgensen and Nicolaisen 1986; Jørgensen and Nicolaisen 1987; Nicolaisen and Jørgensen 1985), the most commonly reported method of normalization for back extensor musculature in the literature is prone isometric back extension, in particular, the postures associated with the Sørensen test. While the Sørensen or Modified Sørensen is a test of endurance, the posture associated with it has been used for the purpose of normalizing EMG signals in the back extensors (Lehman 2002; Danneels et al. 2001; Plamondon et al. 2002). As with the endurance test, there exists considerable

variety in methods for eliciting MVIAs in this posture. Some authors did not measure the force directly, using the resistance of one or more examiners (Lehman 2002; Danneels et al. 2001), and/or used a fixed belt to offer resistance (Danneels et al. 2001). Plamondon et al. (2002) have used Modified Sørensen postures using a dynamometer to record maximal forces for the use of normalization, but has done so only on healthy subjects.

2.4.9 Maximal and Submaximal Testing:

Many studies show that weakness of the trunk extensors correlate well with chronic LBP (Gomez 1994; Kumar et al. 1995; Langrana et al. 1984; Mayer et al. 1985; McNeill et al. 1980). In fact Keller et al. (1999) found pain during exertion was the best predictor of strength. Roy et al. (1989) however, evaluated MVIA forces in control (n=12) and LBP (n=12) groups using a standing dynamometer and reported no significant difference between the control (105.8kg) and the LBP groups (112.1kg) .

Robinson et al. (1991) investigated test-retest reliability of torque for maximal (n=10) and submaximal efforts (n=10) in healthy subjects using a Med-X apparatus. The r values ranged from 0.89-0.96 in maximal efforts and 0.91-0.97 in submaximal efforts using a self rated 50% of maximum. In a more recent study by Dankert et al. (2004), it was reported EMG activity was consistent with Robinson's findings of reliable efforts in healthy controls in both maximal and submaximal conditions, but the authors further reported the reliability for LBP groups as well. Dankaert et al (2004) measured the reliability of trunk muscle activity using EMG in subjects with LBP (n=5) and healthy controls (n=6). Subjects were assessed twice with a week between sessions. They report excellent same day reliability for MVC and submaximal efforts for both LBP and

controls (ICC mean=0.91; range=0.75-0.98). Reliability for both groups across sessions was also excellent with submaximal efforts (ICC mean=0.88; range=0.78-0.97). Reliability is less with maximal efforts in both groups (ICC mean= 0.70; range=0.19-0.99). This data suggests that submaximal efforts are preferable for assessing EMG of the trunk across days.

Daneels et al. (2002) studied the normalized activity of the multifidus and iliocostalis thoracis during a prone isometric strength exercise in prone position similar to the Modified Sørensen test in healthy (n=77), subacute LBP (n=24) and chronic LBP (n=51) groups. Significantly lower normalized EMG was found in the chronic LBP group for both the multifidus (P=0.017) and iliocostalis (P= 0.003) than the healthy controls and the authors reported excellent reliability of the muscles: right iliocostalis pars thoracis (ICPT)=0.91, left ICPT=0.82, right multifidus=0.93 and left multifidus=0.92 for trunk extension and right ICPT=0.92, left ICPT=0.94, right multifidus=0.98 and left multifidus=0.92 for maximal isometric activation in prone extension.

The goal of any normalization procedure is to provide a relative reference point that is stable across muscles, across exertions as well as across subjects. The reference point need not be a maximal exertion as long as it relates to the relative contribution of the muscle. Yang and Winter (1983) used submaximal exertion to normalize muscle activity and found them more reliable than maximal exertions.

Barrata et al. (1998) describes a method for achieving MVIA by increasing exertions of 10% were performed until the subjects could no longer achieve exertion. This method while effective, has been criticized as time consuming and be inappropriate

in clinical groups (Marras and Davis 2001). Marras and Davis (2001) proposed a novel submaximal technique that plots submaximal efforts to predict maximal values. This method assumes linear force-EMG relationship. A linear force-EMG relationship in trunk muscles was reported by many authors, (Chaffin et al. 1980; Moritani and deVries 1978; Perry and Bekey 1981) however non linear relationship have also been found (Woods and Bigland-Ritchie 1983; Solomonow et al. 1990; Solomonow et al. 1986). Marras et al. (2001) went further to use this predictive method using the estimated MVIA value as a normalization reference point and demonstrated spinal load predictions that were matched to actual MVIA driven models.

One of the conceptual problems with normalization techniques that rely on maximal or even predictions of maximal exertions is that they are affected by pain or nociceptive inhibition, motivation, as well as fear-avoidance patterns which may manifest in variation in the measurement of EMG and force production. These methods may therefore not provide a stable baseline in populations where pain levels may vary from session to session.

2.5 SUMMARY

LBP accounts for the majority of musculoskeletal pain and is responsible for a disproportionate amount of the societal cost. Although sometimes self limiting, acute LBP may transition into chronic LBP 10-20% of the time. The longer the problem persists and the more chronic it becomes the less likely the individual will return to work, and the higher the economic burden becomes. Given the personal and social tolls of chronic LBP, one of the primary roles of the primary care physician that deals with LBP

should be to identify risk factors for developing chronicity and to prevent transition of the problem into a chronic stage.

It is well known that subjects with LBP have poorer strength and endurance than those without LBP. The Sørensen and Modified Sørensen test is used as a means to gauge back fitness and has been used in attempts to identify subjects with LBP, and in attempts to predict future occurrences of LBP. In addition to its reported clinical utility, prone isometric back extensions are frequently used in the laboratory setting for normalizing EMG activity with MVIA. While MVIAs are commonly used for normalization, there have been suggestions that they may offer risks to safety in LBP subjects. Moreover, because the maximal efforts are shown to be negatively affected by pain, variable pain conditions, may result in added variability in the normalization process. Adaptations in motor control and modulation of voluntary output observed in LBP populations add complexity and inherent variability to even simple tasks such as prone isometric activations. When we consider the multisegmental nature of the trunk with each segment allowed multiple degrees of freedom and with hundreds of segmental and global muscles acting across one, two, or multiple joints, we recognize that there are countless permutations of activation strategies in trunk extension. Using standardized procedures that minimizes variability in procedure methodology and reduces variability inherent in normalization procedures would be extremely desirable.

Chapter 3

Reliability of Electromyographic and Force Measures During Prone Isometric Back Extension in Subjects With and Without Low Back Pain.

3.1 ABSTRACT

Maximal voluntary isometric activations (MVIA) are frequently used as inputs for models attempting to predict muscle force and as normalization values in studies assessing muscle function. However nociceptive input and pain may adversely affect maximal activation of muscle. The purpose of this study was to assess reliability of MVIA force and electromyographic (EMG) during a prone isometric back extension in subjects with and without low back pain (LBP). A novel sub-maximal method using the percentages of the estimated mass of the head-arms-trunk (HAT) segment was also investigated. Repeated measures on twenty male volunteers divided into a LBP group of subjects with current LBP or a history of LBP that limited their activity (n=10) and a control group who have never experienced LBP that limited their activity (n=10) were made on four occasions. Force and EMG activity were recorded bilaterally upper lumbar erector spinae adjacent to L1 (ULES), lower lumbar erector spinae (LLES), adjacent to L5 and biceps femoris (BF). Subjects were asked to exert a maximal extension effort against a harness assembly that was attached to a force transducer anchored to the floor. Submaximal exertions were also performed with an additional 0, 10, 20, 30, 40, 50, 60 and 70% of HAT. Mean MVIA forces were significantly ($p \leq 0.05$) lower in LBP vs. controls. Intraclass correlation coefficients (ICC) for MVIA force, right and left ULES and LLES EMG indicated high reliability in controls ($R > 0.90$), but were significantly

less in LBP (R=0.36-0.80). EMG of BF demonstrated excellent reliability across both groups (R>0.90). Combining both groups, resistance at 100% HAT demonstrated the highest reliability.

3.2 BACKGROUND

Electromyographic (EMG) signals elicited with maximal voluntary isometric activations (MVIAs) are frequently used as input values to predict muscle force and as normalization values to compare muscle function across muscles and subjects. MVIAs are frequently used as reference values for normalization in studies investigating lumbar muscle function in individuals both with and without LBP. It is well known that maximal voluntary activation of muscle is negatively affected by pain and neuromuscular inhibition (Zedka et al. 1999). Non-physiologic factors of fear avoidance (Menard and Hoens 1994) and motivation (Hirsch et al. 1991) are also known to play a role. If varying levels of pain result in variability of the MVIA, it may not be the ideal reference for normalization studies that investigate LBP and especially those that evaluate the response of LBP to particular therapies.

The most commonly reported method of normalization for back extensor musculature in the literature is prone isometric back extension, in particular, the postures associated with the Sørensen test (Moreau et al. 2001). While the Sørensen or Modified Sørensen is a test of endurance, the posture associated with it has been used for the purposes of normalizing EMG signals in the back extensors (Lehman 2002; Danneels et al. 2001; Plamondon et al. 2002). As with the endurance test, there exists considerable variety in methods for eliciting MVIAs in this posture. Some authors have not measured

the force directly, using the resistance of one or more examiners (Lehman 2002; Danneels et al. 2001) and/or used a fixed belt to offer resistance (Danneels et al. 2001). Plamondon et al. (2002) have used Modified Sørensen postures using a dynamometer to record maximal forces for the use of normalization, but has done so only on healthy subjects.

Many studies show that weakness of the trunk extensors correlate well with chronic LBP (Shirado et al. 1995; Gomez 1994; Langrana et al. 1984; McNeill et al. 1980; Mayer et al. 1985; Pope et al. 1985; Newton and Waddell 1993). Keller and Colloca (2000) found pain during exertion was the best predictor of strength. Roy et al. (1989) however evaluated MVIA forces in control (n=12) and LBP (n=12) groups using a standing dynamometer and reported no significant difference between the control (105.8kg) and the LBP groups (112.1kg).

Test-retest reliability of torque for maximal (n=10) and submaximal efforts (n=10) in healthy subjects using a Med-X apparatus ranged from 0.89-0.96 in maximal efforts and 0.91-0.97 in submaximal efforts using a self rated 50% of maximum effort (Robinson et al. 1991). In another more recent study by Dankaerts et al. (2004) it was reported that EMG activity was consistent with Robinson's findings of reliable efforts in healthy controls in both maximal and submaximal conditions, but also reported excellent reliability for LBP groups as well. Dankaert et al. (2004) measured the reliability of trunk muscle activity using EMG in subjects with LBP (n=5) and healthy controls (n=6). Subjects were assessed twice with a week between sessions with high same day reliability for MVIA and submaximal efforts for both healthy and controls (ICC mean=0.91; range=0.75-0.98). Reliability for both groups across sessions was also

excellent with submaximal efforts (ICC mean=0.88; range=0.78 - 0.97). Reliability is less with maximal efforts in both groups (ICC mean = 0.70; range=0.19 - 0.99). These data may suggest that submaximal efforts are preferable for assessing EMG of the trunk across days.

Daneels et al. (2002), in a study measuring the normalized activity of the multifidus and iliocostalis thoracis during a Modified Sørensen in healthy (n=77), subacute LBP (n=24) and chronic LBP (n=51) groups, found significantly lower normalized EMG in the chronic LBP group for both the multifidus (p=0.017) and iliocostalis (p=0.003) than the healthy controls. High reliability was reported with ICCs for right iliocostalis pars thoracis (ICPT)=0.91, left ICPT=0.82, right multifidus=0.93 and left multifidus=0.92 for trunk extension and right ICPT=0.92, left ICPT=0.94, right multifidus=0.98 and left multifidus=0.92 for maximal isometric activation in prone extension (Dannëels et al. 2002).

The goal of any normalization procedure is to provide a relative reference point that is stable across muscles, across exertions as well as across subjects. The reference point need not be a maximal exertion as long as it relates to the relative contribution of the muscle. Yang and Winter (1983) used submaximal exertion to normalize muscle activity and found them more reliable than maximal exertions.

Barrata et al. (1998) describe a method for achieving an MVIA where increasing exertions of 10% were performed until the subjects could no longer achieve exertion. This method, while effective, has been criticized as time consuming and may not be appropriate in clinical groups (Marras and Davis 2001). Marras et al. (2001) proposed a

novel submaximal technique that plots submaximal efforts to predict maximal values. Predictive methods rely on assumption of linear force-EMG relationship.

One of the problems with normalization techniques that rely on maximal or even predictions of maximal exertions is that they are affected by pain or nociceptive inhibition, motivation, as well as fear-avoidance patterns, which may manifest in variation in the measurement of EMG and force production. These methods may therefore not provide a stable baseline in populations where pain levels may vary from session to session. The purpose of this paper was to investigate the numeric stability of forces and the associated EMG activity during MVIA, and with a novel sub-maximal normalization technique. The reliability of these measures will be compared to each other and between the LBP and control groups.

3.3 METHODOLOGY

3.3.1 SUBJECTS

Twenty male volunteer subjects were recruited from the university population. Based on a self report of currently having LBP or having a history of chronic or recurrent LBP, the volunteers were grouped into a LBP group (n=10) and a control group (n=10) *(Table 3.1)*. Subjects in the LBP group had a mean age of 29.1 years (±8.2) and mean mass of 79.7 kg (±11.2) as compared to 24.7 years (±2.9) and 81.9 kg (±7.8) for controls.

All subjects completed an Oswestry Low Back Pain Disability Questionnaire (Fairbank et al. 1980; Roland and Fairbank 2000) as well as a numeric pain scale. The experiment was explained to the subject and any questions or concerns were addressed and the subjects were informed that they could withdraw from the experiment at any time. A consent form was read and signed prior to experimentation. The Memorial University of Newfoundland Human Investigations Committee approved the study.

3.3.2 EXPERIMENTAL DESIGN

Repeated measures were taken over four sessions. Each testing session was separated by at least 24 hours. Subjects were asked to avoid engaging in vigorous physical activity prior to the testing. Subjects were asked to perform at least three maximum voluntary isometric back extension efforts followed by a series of six submaximal isometric back extension efforts using their head, arms, and trunk mass (HAT), plus an additional percentage (10%, 20%, 30%, 40%, 50%, 60%, 70%) of their

head arms and trunk mass. Each subjects' HAT is calculated using the their body weight and normative data derived through regression equations (Zatsiorsky 2002).

3.3.3 INSTRUMENTATION

A series of 1cm diameter silver/silver electrodes spaced 1 cm apart were used as part of a bipolar, differential surface EMG collection system (ME3000P; Mega Electronics Ltd, Kuopio, Finland) and was used to collect the electrical activities of 6 muscles of the trunk and thigh. Channels were sampled at 1000 Hz, band-pass filtered between 20 Hz and 500 Hz and amplified (differential amplifier, common mode rejection ratio 130 dB, gain x 1000, noise 1 μV). They were converted from analogue-to-digital (12-bit), and stored on computer for analysis. Signal amplification was done at the reference electrode site to minimize signal artifacts caused by movements and external noise. The raw EMG signals were full-wave rectified and low-pass filtered at 4 Hz.

Electrodes were placed bilaterally over the lower lumbar erector spinae (LLES) 2 cm lateral to the L5-S1 spinous processes and over the upper lumbar erector spinae (ULES) 6 cm lateral to the L1-L2, spinous processes. There are a number of studies that have used similar L5-S1 electrode placement to acquire EMG activity for the multifidus (Hermann and Barnes 2001; Vezina and Hubley-Kozey 2000; Danneels et al. 2002). Recently however, Stokes et al. (2003) reported that intra-muscular needle electrodes were needed for accurate assessment of the multifidus. For this present study, the EMG activity collected by the electrode arrangement is referred to as LLES as it is expected that there is activity from more than just multifidus. In the same way it is expected to emphasize the measurement of the multifidus at the lumbosacral junction with narrow

electrode placement, it is expected to emphasize the longissimus thoracis with electrode placement more lateral to the L1-L2 spinous processes. With this placement it is likely that there will be interpretation of signals from iliocostalis lumborum and multifidus and thus the observed EMG activity in this study is referred to as ULES. Electrodes were also placed bilaterally in the mid belly of the BF. For all collection arrays, reference electrodes were placed 5-10 cm away from the collecting electrodes.

Electrodes were placed in the same location for each subject by using bony landmarks and careful palpation. Repeatability of electrode placement was enhanced by using both skin marking and measurement techniques. Thorough skin preparation for all electrodes included shaving local body hair and removal of dead epithelial cells with a very fine grade sandpaper around the designated areas followed by cleansing with an isopropyl alcohol swab.

Force transmitted through the harness assembly, placed at the T5/T6 level was collected through a Wheatstone bridge configuration strain gauge (Omega Engineering Inc. 55LCCA 250). The signal was converted from analog to digital (MP100 analog to digital converter; Biopac Systems Inc. Holliston, MA) and stored and analyzed through computer software. (Acqknowlege III, Biopac Systems Inc. Holliston, MA). (Figure 1)

3.3.4 PRONE BACK EXTENSION

The posture adopted for the test was a variation of the Bering-Sørensen test (Biering-Sørensen 1984). Subjects lay prone on a padded examination table, with the trunk of the body extended off the edge of the table at the level of the anterior superior iliac spine of the pelvis. The lower legs, thighs and mid-buttocks region were restrained

from motion using wide straps attached to the examination table. A pad placed under the ankles prevented subjects from bracing against the table with their feet. A harness was attached around the trunk at the T4/5 vertebral level. The strain gauge was attached to this harness at a midline location of the trunk while the other end was attached to an anchor plate at floor level. The harness/strain gauge assembly was adjusted so the subject maintained a trunk orientation parallel with the floor (Figure 3.2). The subject exerted isometric exertions for a period of 3-4 seconds during the maximal and submaximal (percentages of HAT) exertions while EMG signals and extension forces were recorded. A minimum of 2-minute rest period was given to the subject to allow sufficient recovery. The trunk was supported against gravity during rest periods.

3.3.5 MAXIMAL AND SUBMAXIMAL PROTOCOLS

Subjects were asked to exert three to five maximal isometric extension efforts of approximately 3-4 seconds each. Subjects would be cued to start and given standard verbal encouragement during the effort. Maximal efforts were not randomized and always presented at the beginning of the testing session. After maximal efforts, a computer screen was placed at a comfortable viewing distance from the subject to allow for visual feedback of submaximal efforts (Figure 3.2). The force displayed on the computer screen was calibrated so that 10% increments of HAT were visible to the subject for feedback. Subjects were then asked to hold their HAT mass plus a randomly presented percentage of HAT (0, 10, 20, 30, 40, 50, 60 and 70%) for 3-4 seconds. Each exertion was punctuated with a rest period of at least 2 minute to allow muscle recovery.

3.3.6 DATA ANALYSIS AND STATISTICS:

3.3.6.1 Subject Characteristics:

Subject age and mass were compared with independent T-tests between controls and LBP groups. A Mann-Whitney U-test was used to compare non-parametric data of the Oswestry and Pain scales between groups.

3.3.6.2 Force and EMG:

All signals were visually inspected during real time collection of EMG to ensure optimal signal quality. A 2 second interval of EMG was collected over the most stable segment and was rectified and average amplitude recorded. A daily average maximal force was calculated by averaging the 3 closest maximal force values. Intraclass correlation coefficients (ICC's) were calculated for extensor force, EMG of each muscle during each MVIA and each percentage of HAT. Reliability was assessed using an alpha (Cronbach) model intraclass correlation coefficient (ICC) (Cohen 1988). Force (N) output of the MVIA condition was compared between groups using an independent t-test.

The EMG of each extensor muscle at MVIA was compared between groups using a 2x6 (Group x Extensor Muscle) configuration ANOVA (SPSS 12.0 for windows, SPSS Inc., US). ICC's were calculated for the EMG of MVIA's and HAT for control and LBP groups. The ICC's were compared with a 2x8 (Group x HAT%) configuration ANOVA (SPSS 12.0 for windows, SPSS Inc., US) for each of the muscle groups. Bonferroni post-hoc tests were used to discriminate between individual and significant differences. Data in the text and figures include means and standard deviation (SD).

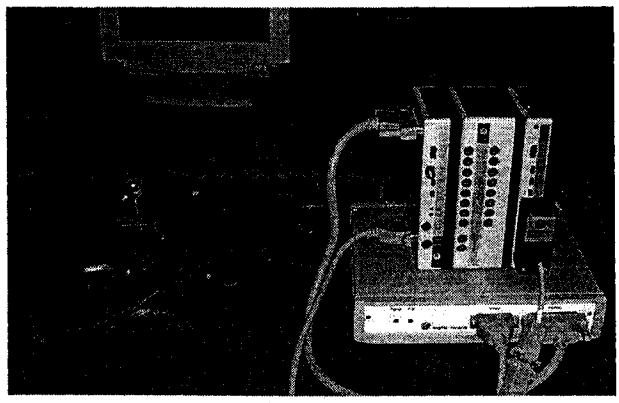


Figure 3.1. Wheatstone bridge configuration strain gauge *(Omega Engineering Inc. 55LCCA 250)*. MP100 analog to digital converter; *(Biopac Systems Inc. Holliston, MA)*

Figure 3.2: Harness/strain gauge assembly was adjusted so the subject maintained a trunk orientation parallel with the floor, with monitor placed for visual feedback.

3.4 RESULTS

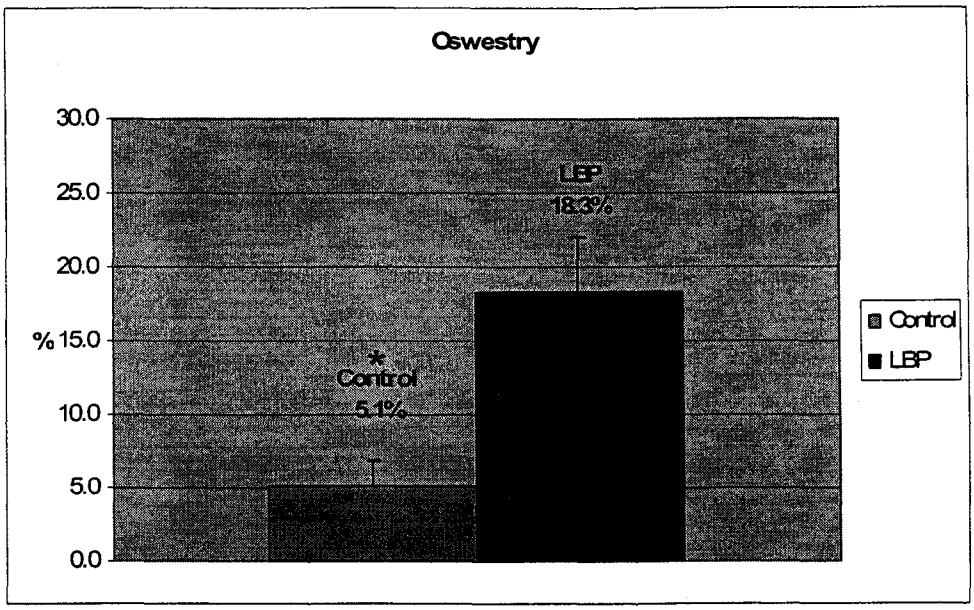
3.4.1 SUBJECT CHARACTERISTICS

Table 3.1 reports subject characteristics and mean scores of the Oswestry Disability Index and 0-10 Pain scale. Oswestry Low Back Disability Index scores were 73% lower and pain scores 97% lower in the Control group than LBP group. The LBP group had an Oswestry mean score of 18.3% (±11.8) and pain score 3.43 (± 2.0) as compared to Oswestry of 5.1% (±5.5) and pain score of 0.1 (±0.4) for controls (Figures 3.3 and 3.4). Using the Mann-Whittney Test significant differences ($p=0.007$) were found for Oswestry Low Back Disability Index scores between LBP and Control groups and significant differences ($p \leq 0.001$) in pain levels between LBP and Control groups.

Table 3.1: Subject Characteristics

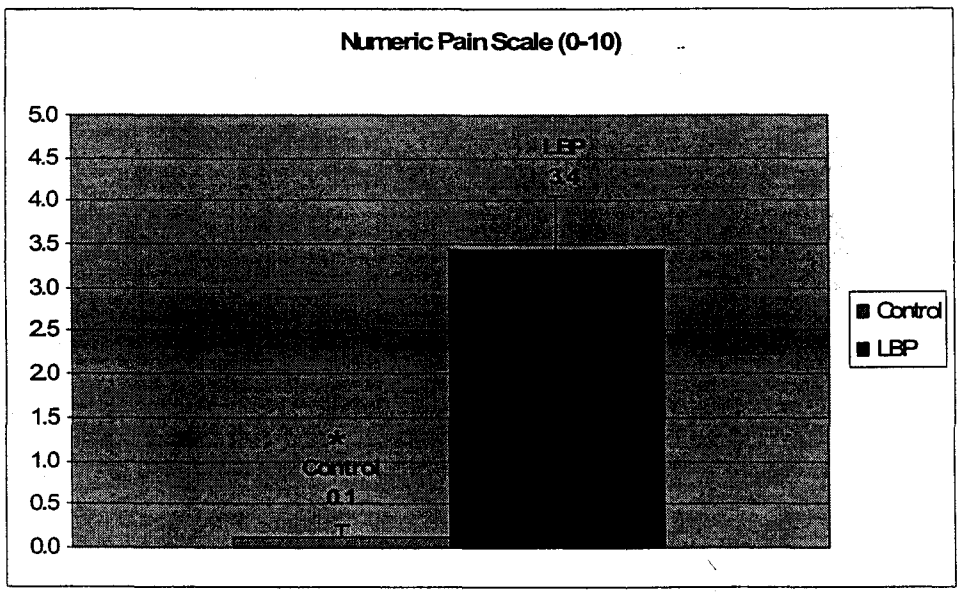
Variable	Group	Mean	SD	*p*
Age (years)	Control	24.7	2.91	0.401
	LBP	29.1	8.26	
Mass (Kg)	Control	79.7	11.17	0.785
	LBP	81.2	7.81	
Oswestry (%)	Control	5.1	5.45	0.007*
	LBP	18.3	12.38	
Pain Scale (x/10)	Control	0.13	0.35	0.001*
	LBP	3.4	1.98	

* Significant at $p \leq 0.05$



*Significant at p≤0.05

Figure 3.3: Oswestry Disability Index

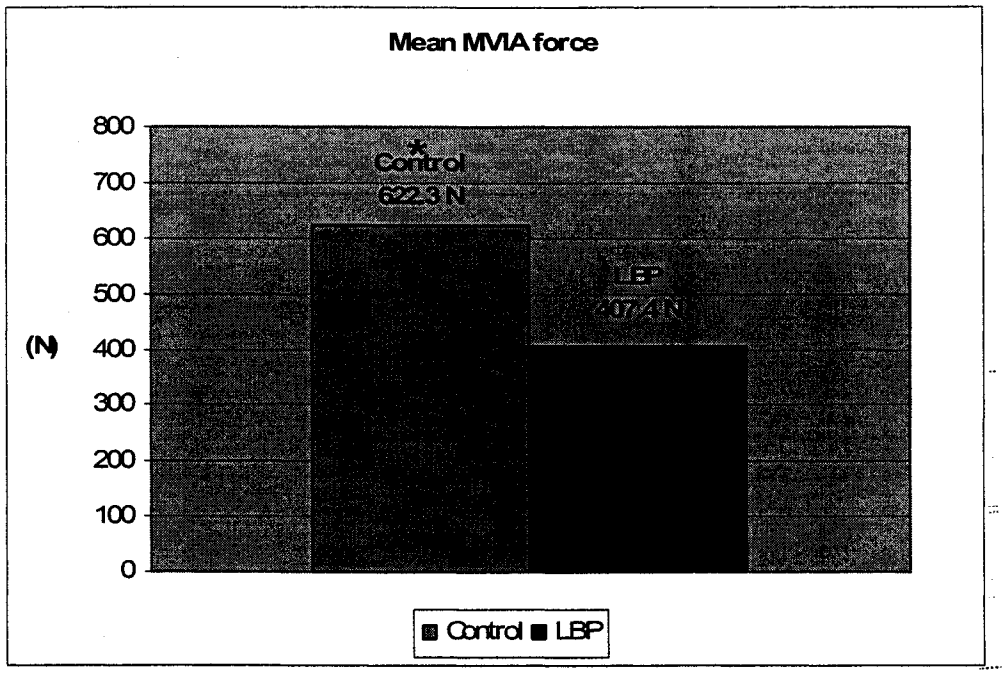


*Significant at p≤0.05

Figure 3.4: Pain Scale

3.4.2 MAXIMAL VOLUNTARY ISOMETRIC ACTIVATIONS (MVIA):

Figure 3.5 depicts MVIA's forces that were 34% lower in the LBP group than in controls. Differences between LBP and controls' EMG at MVIA were significant for lumbar extensor muscles: left ULES (p=0.001), right ULES (p=0.001) and the right LLES (p=0.003). EMG of left LLES and both right and left BF were not significantly different between LBP and Control groups.



*Significant at p≤0.05

Figure 3.5: Mean MVIA Force Output

3.3.3 RELIABILITY OF MAXIMAL VOLUNTARY ISOMETRIC ACTIVATIONS

ICCs for both force and EMG measures collected across 4 MVIA sessions are reported in Table 3.2. BF demonstrates the highest ICCs in both groups, and no significant differences were noted in the BF between groups. Right and left ULES and LLES muscles

demonstrated more marked difference between control and LBP groups. There was a significant difference in the EMG of right LLES between groups, but not in the EMG of the left LLES.

Table 3.2: ICCs for Maximum Force and EMG during MVIA

Muscle	Group	ICC	*p*
Force [a]	Control	0.98	<0.0001*
	LBP	0.80	
L ULES	Control	0.92	<0.0001*
	LBP	0.36	
R ULES	Control	0.96	<0.0001*
	LBP	0.52	
L LLES	Control	0.96	0.229
	LBP	0.72	
R LLES	Control	0.93	0.010*
	LBP	0.72	
L BF	Control	0.99	0.329
	LBP	0.93	
R BF	Control	0.96	0.329
	LBP	0.94	

*Significant at $p \leq 0.05$ [a] Evaluated with independent t-test

3.3.4 RELIABILITY OF ISOMETRIC ACTIVATIONS AT PERCENTAGES OF HAT

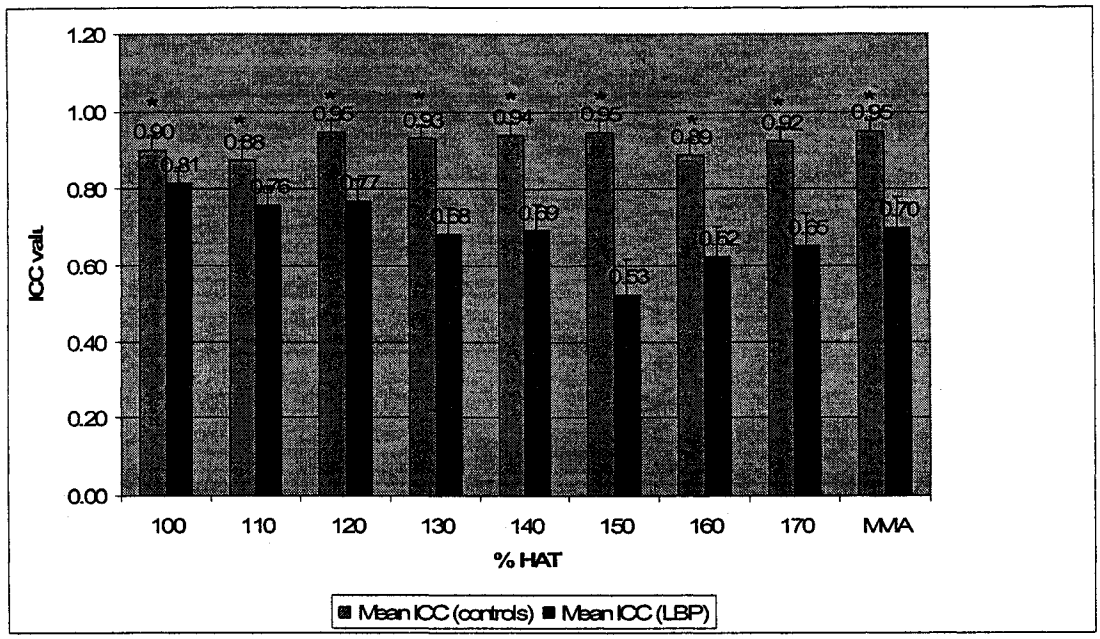
Table 3.3 reports the ICCs for all extensor muscles and compares the mean ICCs of the six extensor muscles for each %HAT and MVIA of Controls with that of the LBP group. There was excellent correlation in all muscle groups in all % HAT in the control group, but much less homogeneity in the LBP group compared to the control group. With ICCs combined over all muscle groups there were significant differences between LBP

and Control groups at all levels of HAT and at MVIA. Figure 3.6 depicts the mean ICCs for extensor muscles for each level of HAT and at MVIA.

Table 3.3: Comparison of intraclass correlation coefficients for back extensor musculature between groups

%HAT		L ULES	R ULES	L LLES	R LLES	L BF	R BF	*p*
100	Control	0.80	0.88	0.94	0.93	0.94	0.91	0.037*
	LBP	0.74	0.88	0.85	0.81	0.72	0.88	
110	Control	0.81	0.74	0.92	0.90	0.95	0.93	0.044*
	LBP	.60	0.79	0.75	0.72	0.83	0.86	
120	Control	0.88	0.93	0.97	0.95	0.98	0.96	0.005*
	LBP	0.57	0.80	0.73	0.74	0.88	0.88	
130	Control	0.85	0.91	0.96	0.94	0.98	0.93	0.008*
	LBP	0.37	0.71	0.63	0.68	0.86	0.84	
140	Control	0.89	0.92	0.96	0.94	0.98	0.95	0.006*
	LBP	0.46	0.72	0.57	0.64	0.89	0.88	
150	Control	0.89	0.92	0.97	0.94	0.98	0.97	0.011*
	LBP	0.03	0.61	0.33	0.44	0.88	0.86	
160	Control	0.87	0.90	0.90	0.87	0.98	0.80	0.019*
	LBP	0.21	0.67	0.67	0.62	0.66	0.90	
170	Control	0.84	0.90	0.95	0.91	0.98	0.95	0.025*
	LBP	0.45	0.31	0.67	0.65	0.92	0.92	
MVIA	Control	0.92	0.96	0.96	0.93	0.99	0.96	0.021*
	LBP	0.36	0.52	0.72	0.72	0.93	0.94	
p		<0.0001†	0.002†	<0.0001†	<0.0001†	0.001†	0.043†	

*Significant at $p \leq 0.05$ between LBP and Controls Groups for each % HAT
†Significant at $p \leq 0.05$ between LBP and Controls Groups for each muscle group



*Significant at $p \leq 0.05$

Figure 3.6: Comparison of mean ICCs for control and LBP groups

3.4.5 HAT AS A PERCENTAGE OF MVIA

Figures 3.7 and 3.8 depict the differences between the force (N) elicited with MVIA compared to the subjects' HAT. In the control group there was only one instance (subject 7) where the HAT exceeded the mean force of the MVIA in contrast to the five of ten LBP subjects where HAT exceeded the mean force of the MVIA. Figure 3.9, expresses control and LBP subject's MVIA values as a percentage of HAT. The control group had mean MVIAs that were of a higher percentage of HAT than that of the LBP group. Interestingly, all subjects except for 1 LBP subject were able to attain 170% HAT in at least 1 of the 4 sessions. It is unclear as to why subjects could achieve the target force better on some days or on some sessions. Pain was not reported to be a factor, nor were there any technical problems. Rather, the subjects simply stated that they couldn't

produce or maintain the required force. A summary of inability to achieve the given percentage HAT is reported in Table 3.10.

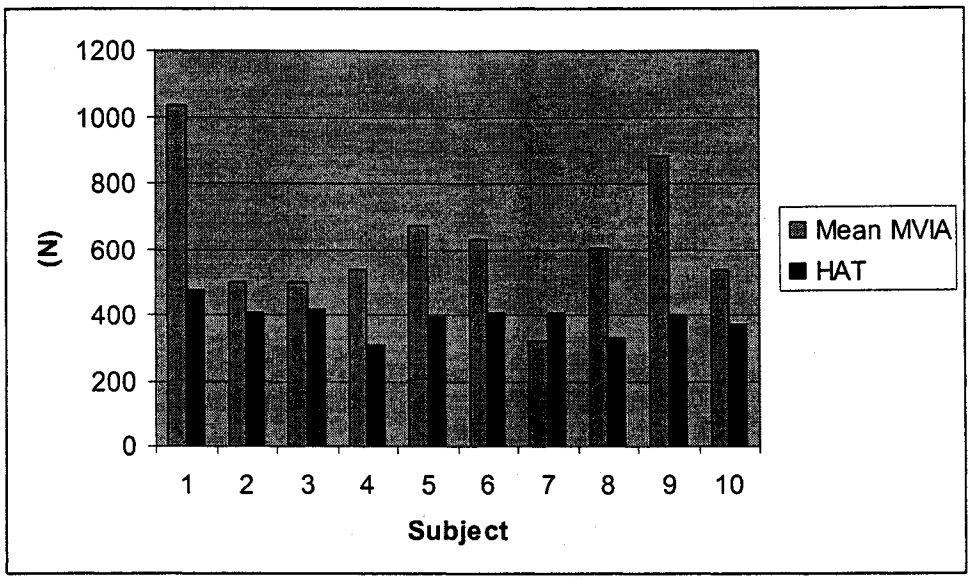


Figure 3.7: Mean MVIA of each Control subject compared with their HAT

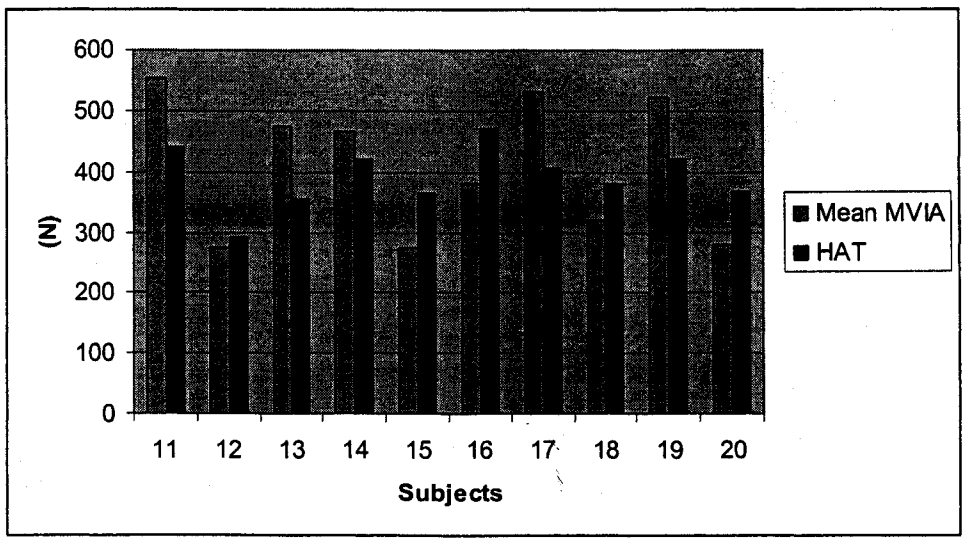


Figure 3.8: Mean MVIA of each LBP subject compared with their HAT

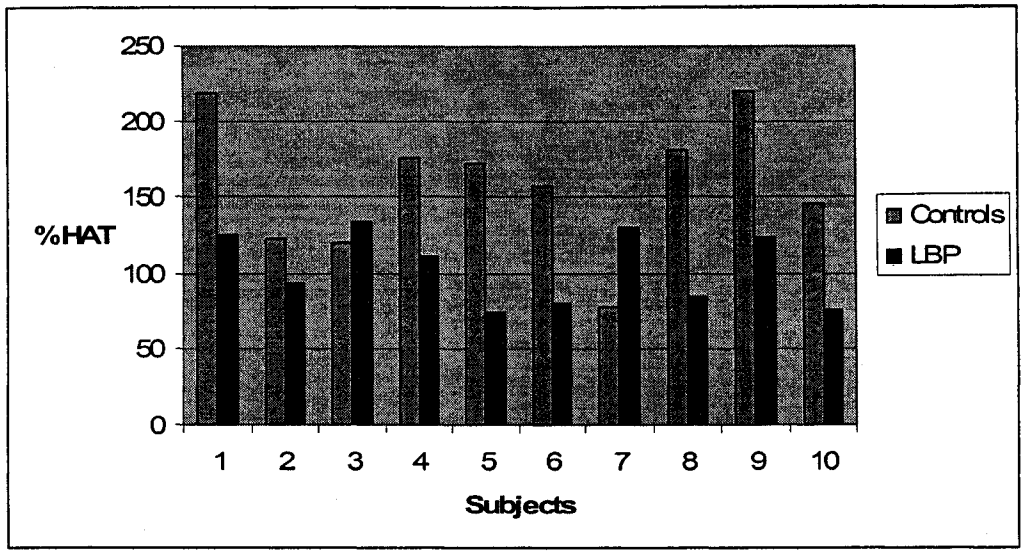


Figure 3.9: MVIA as a percentage of HAT in each subject

Table 3.4: Inability to achieve given percentage of HAT

%HAT		Subjects unable to complete all 4 sessions	Sessions unable to be performed
110	Control	-	-
	LBP	1	1
120	Control	1	1
	LBP	-	-
130	Control	-	-
	LBP	1	1
150	Control	-	-
	LBP	3	3
160	Control	2	2
	LBP	1	1
170	Control*	2	3
	LBP†	5	10

** Number of sessions unable to be achieved ranged from 1-2 with one subject unable to perform 2/4 and another unable to perform 1of 4*

† Number of sessions unable to be achieved varied from 1-4 with 1 subjects unable to achieve all 4, one subject unable to achieve 3 of 4 and three subjects unable to achieve 1of 4

3.5 DISCUSSION

3.5.1 STRENGTH DIFFERENCES WITH MVIA

The findings of significant differences between both MVIA force output and EMG activity between control and LBP groups are consistent with other studies using prone isometric extension (Biering-Sørensen 1984; Addison and Schultz 1980) but in contrast to other findings using standing dynamometry that reported no significant differences between force output of control and LBP groups (Roy et al. 1989; Jørgensen and Nicholaisen 1987; McNeil et al. Spine 1980).

3.5.2 RELIABILITY OF FORCE AND EMG WITH MVIA

The findings of excellent reliability (ICC≥0.90) of the force output with MVIA is consistent with the literature for prone extension postures (Plamondon et al. 1999) as well as for standing (Rytokoski 1994) and seated dynamometry (Robinson 1991; Smidt et al. 1983). The findings in this study show results similar to a study by Daneels (2002) that reported ICCs for EMG during maximal extension efforts in right and left iliocostalis lumborum pars thoracis (0.92, 0.94) and in right and left multifidus (0.98, 0.92). MVIAs are generally considered highly reliable in healthy controls and are effective for normalizing EMG and our findings show ICCs for EMG as excellent (ICC≥0.90) in all extensor muscle groups across 4 sessions with MVIA. This however does not hold true with the LBP group with ICC values ranging from 0.36 to 0.94 in the extensor musculature.

3.5.3 RELIABILITY OF EMG WITH PERCENTAGES OF HAT

Dankaerts et al. (2004) evaluated trunk extension with EMG in Control and LBP groups reporting excellent reliability in both control and LBP groups for submaximal efforts (ICC mean=0.88; range 0.78-0.97), but less reliability for maximal efforts in both groups (ICC mean=0.70; range 0.19-0.99). This is in direct contrast with findings in this study that indicate the control group had excellent EMG reliability in the maximal condition, (ICC mean=0.95; range 0.92-0.99) but less in the LBP group (ICC mean=0.70; range 0.36-0.94). Of note in the Dankaerts et al. study, is the wide range of ICC values (0.19-0.99) for maximal efforts compared to the very narrow range of ICC values in maximal efforts (0.92-0.99) in the present study. With lower number of subjects in the Dankaerts et al. study (Controls n=6 and chronic LBP n=5), the ICC mean may have been more sensitive to the low values in the listed range resulting in a lower value. Many levels of submaximal efforts at percentages of HAT were assessed in this study, but even in the most reliable condition in the LBP group (100% HAT), mean ICCs were much less than previously reported (ICC mean=0.81; range of 0.62-0.81). Although inclusion criteria allowed subjects with a current episode of LBP, all of our LBP subjects had either a history of recurrent or chronic LBP. As a result, the actual differences between groups may have been less marked than in similar studies that may have studied "more severely" injured subjects.

The highest average ICC value for the LBP group combining all extensor muscles occurred in the 100% HAT condition (ICC=0.81) compared to the highest average ICC value for the control group combining all extensor muscles occurring with the MVIA, 120% HAT and 150% HAT conditions (ICC=0.95). When averaging all extensor muscles

combining both groups, the 100% HAT condition demonstrated the highest overall reliability (ICC=0.86). The lowest averaged ICC for the LBP group occurred at 150% HAT (ICC=0.53) with the lowest ICC averaged across muscles for the control groups occurring at 110% HAT (ICC=0.88). These ICC values may have been affected by the inability to perform higher values of %HAT, as there were 3 sessions overall in the LBP group where 150% could not be achieved, and no sessions in the control that could not be achieved. Inability to achieve given percentage of %HAT was not a factor in the control group at 110% as all subjects were able to perform 110%.

3.5.4 RELIABILITY OF EMG OF ULES AND BF

For consistency, all subjects were strapped to the table in the same manner. The velcro fasteners were on the left side of the table and were cinched to that side. There is discrepancy in the ICC values between the left LLES and the right LLES at MVIA, whereas there is no discrepancy with other muscle groups. Although support from cinching the restraint system could potentially result in less variation in the left LLES, we would expect a side to side difference in ICC values to be seen in the ULES and BF also. It is suspected that the decrease in significance noted in both LLES, and more markedly the left LLES, may suggest that this group does not offer the most important contribution to extension efforts resulting in the difference in force output seen in the MVIA condition.

The BF EMG also demonstrated the highest ICCs values in both groups and was not significantly different between groups. It is proposed that if the contribution of the BF

was critical to the difference in force output extension effort between groups, a difference in EMG output would be expected.

As the most significant differences between groups and most disparate ICC values were observed in the ULES, it is possible that the discrepancy in the EMG of these muscles contribute to the marked MVIA force output observed between groups.

3.5.5 PAIN AND MOTOR CONTROL STRATEGIES

The thoracolumbar spine has many degrees of freedom with a multitude of segmental and global muscles crossing single and multiple segments. These mechanics coupled with all the available motor control strategies elucidate the complexity of a seemingly simple prone isometric extension. Pain is known to negatively affect maximal activation of muscle through neuromuscular inhibition (Zedka et al. 1999) and has been shown to alter motor control patterns in the trunk (O'Sullivan et al. 1997; Richardson and Jull 1995). Lower force output and ICC values in the LBP group could be due to factors such as neuromuscular inhibition, altered motivation or fear avoidance. The decreased reliability of LBP group EMG could be the product of altered trunk motor control strategies across days compared to potentially more consistent strategies employed by control groups.

3.5.6 MVIA AS A PERCENTAGE OF HAT

It has been suggested that performance of MVIA in patients with LBP could compromise safety (Moffroid et al. 1993). It has been reported that forces required to maintain a horizontal position are well below MVIA in healthy populations, (Moffroid et

al. 1993; Mayer et al. 1995; Jørgensen and Nicolaisen 1986) but may rise to as much as 85% in a patient with chronic LBP (Hultman et al. 1993). In this study the 100% HAT value exceeded the MVIA value in half of our LBP group, however some subjects were able to achieve and sustain %HAT values that were over mean forces associated with their MVIAs. While it is possible that these supramaximal values are partly a product MVIA variation values within and between days, it is also possible that visual feedback used to attain a given % HAT provides more substantial motivation than the verbal motivation provided during the MVIA sessions.

3.5.7 LIMITATIONS:

There were a number of limitations to this study. One of the largest limitations is that the study was a relatively small study containing only 10 subjects each group. Surface electromyography was used in lieu of needle electrodes for a number of ethical and practical reasons. This decision resulted in the necessity to describe muscles groups with more generalization than if needle electrodes were used.

Although inclusion criteria allowed subjects with a current episode of LBP, all of our LBP subjects had either a history of recurrent or chronic LBP. Many of the subjects with a history of LBP had very little discomfort at the time of the test. Additionally the LBP group's pain score were highly variable. As a result, the actual differences between groups may have been less marked than similar studies that may have studied "more severely" injured subjects.

3.6 CONCLUSIONS

There have been conflicting reports in the literature on whether differences in strength exist between healthy individuals and individuals with LBP. The findings of this study suggest that significant strength differences are present between control and LBP groups. In evaluating the numeric stability of EMG activity during these maximal efforts, it appears that since the BF are consistent in their activity in both LBP and control groups, they are not responsible for differences in force output seen between groups. Further the discrepancy in reliability was most marked in the ULES suggesting that they may contribute the most to the discrepancy between the observed force outputs.

Although MVIA is commonly used for normalization purposes, our results suggest that MVIA may only be ideal in healthy subjects and that submaximal efforts may be more reliable across sessions in subjects with LBP. Overall, it appears that use of a condition of 100% HAT would yield more overall reliable EMG results for both groups than use of MVIA.

The ability of some subjects to achieve supramaximal values suggests a need to investigate potential differences between providing verbal motivation and visual targets for eliciting MVIAs. Further research is needed to compare the use of normalized EMG values using MVIA and the submaximal HAT values as input values in biomechanical models and assess potential differences between the two methods.

Chapter 4

Neuromuscular Fatigue during a Modified Sørensen Test in Subjects With and Without Low Back Pain

4.1 ABSTRACT

Prospective studies employing modified Sorenson tests have reported that neuromuscular endurance of low back musculature is related to the potential for developing low back pain (LBP). Understanding the manner in which spinal musculature fatigues in people with and without LBP is necessary to gain insight into the predictive validity of the Sorenson test. Such information also helps provide direction for implementing preventative measures (e.g. back strengthening programs). Twenty male volunteers were divided into a LBP group of subjects with current subacute LBP or a history of LBP that limited their activity (n=10) and a control group who have never experienced activity limiting LBP (n=10). Spectral contents were calculated from bilateral surface electromyography (EMG) measures of the upper lumbar erector spinae (ULES), lower lumbar erector spinae (LLES) and biceps femoris (BF) as subjects with and without LBP maintained a prescribed Sørensen Test position and exerted isometric forces equivalent to 100, 120, 140 and 160% of the estimated mass of the head-arms-trunk (HAT) segment against a load cell. Time to failure was also investigated across the percentages of HAT. Endurance time decreased with increasing load and differences between groups increased as load increased, however these differences were not significant. Significant differences in the median frequency of the EMG signal between groups occurred only in the right BF ($p \leq 0.05$) with significant pairwise differences occurring only at 140% for the left BF and at 160% for the right BF. There were

significant pairwise differences at 120% for average EMG of the right BF and at 140% for the right ULES, and right and left BF ($p \leq 0.05$). The Modified Sørensen test as usually performed at 100% HAT is not sufficient to demonstrate significant differences between subjects without LBP, and those with a history of recurrent or chronic LBP.

4.2 BACKGROUND

Poor neuromuscular endurance of low back musculature has been related to the potential for developing low back pain (LBP) (Biering-Sørensen 1984; Hultman et al. 1993); (Alaranta et al. 1995; Mayer et al. 1995; Nelson et al. 1995; Smidt et al. 1983). Additionally, decreased trunk strength and endurance associated with a cyclical pattern of deconditioning through pain, avoidance and inactivity is noted as a defining characteristic of chronic LBP (Mayer and Gatchel 1988). The reported rate of recovery of LBP is high during the first few weeks (Coste et al. 1994), however the recurrence rate is also high (Smedley et al. 1998). Of the 8% of the working age population that experiences disabling LBP in a given year (Straus 2002), 10-20% of these are reported to become chronic (Waddell 2004).

Back pain accounts for 25% of all work-related injury, but a disproportionate 40% of lost time claims. Workers with back pain are less likely to return to work than with other injuries (Johnson 1998). A disproportionate distribution of costs is unmistakable with chronic cases being responsible for 42% of direct medical cost, 54% of indirect indemnity cost and 52% of the cost overall. Further, 25% of workers that lost time over 3 months accounted for over 75% of the costs (Williams et al. 1998). Data from the Quebec Workers Compensation Board reported in the Quebec Task Force show 7-10% of back

cases account for more than 70% of total health care and indemnity costs (Spitzer et al. 1987).

Given the potential economic and social burden of developing chronicity in the lower back, preventing a transition from acute LBP to chronic LBP, or identifying potential risk factors, such as poor lumbar extensor endurance would be of paramount importance. The most widely reported endurance test in the literature is the Biering–Sørensen test (Moreau et al. 2001). A modified Biering–Sørensen test to measure back endurance is currently in use by the Canadian Society for Exercise Physiology (CSEP) in their Canadian Physical Fitness and Lifestyle Approach (CPAFLA) testing. (CSEP 2004.)

4.2.1 BIERING–SØRENSEN

In the Sørensen test, subjects have their lower body secured to a table with three wide straps and are positioned such that their anterior superior iliac spine is aligned with the edge of the table. Subjects are asked to hold their unsupported head arms and trunk (HAT) parallel to the floor until exhaustion to a maximum of 240-300 seconds. Administration of the Sørensen Test is inconsistently practiced in the literature, including differences in arm position, number of straps (or no straps) and conclusion criteria. These variations have been grouped together as Modified Sørensen tests (Moreau et al. 2001). This test is generally considered safe for both healthy and clinical populations (Biering-Sørensen 1984; Moffroid 1997; Nordin et al. 1987; Alaranta et al. 1994; Peltonen et al. 1998; Alaranta et al. 1995; Mannion and Dolan 1994). While forces required to maintain a horizontal position are well below forces of MVIA in healthy populations (Mayer et al. 1995; Moffroid et al. 1993; Jørgensen and Nicolaisen 1986), they may rise to as much as

85% in a patient with chronic LBP (Hultman et al. 1993). It has been suggested that performance of maximal activations in patients with LBP could compromise safety (Moffroid et al. 1993). There is considerable range of mean endurance times reported for the Sørensen test in the literature ranging from 84s to 180s in healthy males (Jørgensen and Nicolaisen 1986; Sparto et al. 1997; Jørgensen and Nicolaisen 1987; Biering-Sørensen 1984; Kankaanpaa et al. 1998; Mannion and Dolan 1994; Nicolaisen and Jørgensen 1985; Hultman et al. 1993). For males with LBP, mean endurance times range from 80s-194s (Jørgensen and Nicolaisen 1987; (Biering-Sørensen 1984; Nicolaisen and Jørgensen 1985; Hultman et al. 1993).

Although the Sørensen test is generally considered a measure of low back function measuring overall lower back fatigue, activity of the biceps femoris (BF) and hip extensors have been argued to substantially contribute to endurance times (Kankaanpaa, et al. 1998). Significant correlation has been observed between Sørensen endurance times and EMG median frequency (MF) slopes of the biceps femoris. (Moffroid et al. 1994; Moffroid 1997). EMG fatigue analysis of gluteus maximus muscles show they are more fatigable in chronic LBP patients than in healthy control subjects during a sustained back extension endurance test (Kankaanpaa et al. 1998). In a recent study by McKeon et al. (2006), it was observed that females with LBP had greater recruitment and more fatigue in this muscle as compared to the female without LBP.

Ng et al. (1997) demonstrated that the multifidus has more activity than the iliocostalis lumborum during Sørensen testing. The multifidus fatigues at a faster rate than the iliocostalis lumborum during this test demonstrating a higher initial MF and normalized MF slope (Ng et al. 1997). Ng and Richardson (1996) suggests that the

Modified Sørensen test with the use of EMG power spectral analysis may be a reliable method to measure the fatigue rate of the back muscles if cross-talk is minimized and adds that measuring the fatigue rate of the multifidus may be a useful clinical measure. Van Dieën et al. (1993) observed that the multifidus muscle at the L5 level appeared to show the most consistent changes of the EMG power spectrum as a consequence of fatigue.

Maintaining a horizontal position Sørensen tests (100% HAT) naturally result in a higher endurance times than at higher levels of resistance. With increased endurance times, motivation and pain levels and alternative muscle control strategies may be allowed to play a larger role. The purpose of this paper is to gain a better understanding of muscle activity during a modified Sørensen test in subjects with and without LBP, to determine if this test is sufficient to discriminate between groups, and to investigate the effects of higher than HAT resistance to elicit fatigue.

4.3 METHODOLOGY

4.3.1 SUBJECTS:

Twenty male volunteer subjects were recruited from the university population. These subjects were grouped into low back (LBP) group (n=10) and control groups (n=10) (Table 4.1). Subjects were included in the LBP group based on a self report of currently having LBP or having a history of chronic or recurrent LBP that limited activity. All subjects completed an Oswestry Low Back Pain Disability Questionnaire (Fairbank et al. 1980; Thomas et al. 1989) as well as a numeric pain scale. Subjects in the LBP group had a mean age of 29.1 years (±8.2) and mean mass of 79.7 kg (±11.2) as compared to 24.7 years (±2.9) and 81.9 kg (±7.8) for controls. The experiment was explained to the subject and any questions or concerns were addressed and the subjects were informed that they could withdraw from the experiment at any time. A consent form was read and signed prior to experimentation. The Memorial University of Newfoundland Human Investigations Committee approved the study.

4.3.2 PRONE BACK EXTENSION

The posture adopted for the test was a variation of the Bering-Sørensen test (Biering-Sørensen 1984). Subjects lay prone on a padded examination table, with the trunk of the body extended off the edge of the table at the level of the anterior superior iliac spine of the pelvis. The lower legs, thighs and mid-buttocks region were restrained from motion using wide straps attached to the examination table. A pad placed under the ankles prevented subjects from bracing against the table with their feet. A harness was attached around the trunk at the T4-5 level. The strain gauge was attached to this harness at a

midline location of the trunk while the other end was attached to an anchor plate at floor level. The harness/strain gauge assembly was adjusted so the subject maintained a trunk orientation parallel with the floor. The trunk was supported against gravity during rest periods (Figure 4.2).

4.3.3 EXPERIMENTAL DESIGN:

Using the subject's body mass and normative data derived through regression equations, (Zatsiorsky 2002) the subject's HAT mass was calculated. Using Zatsiorsky's calculations, it was found that subjects' HAT mass was 49.11% of their total body mass. The force displayed on the computer screen was calibrated so that 10% increments of HAT were visible to the subject for feedback (Figure 4.3). Repeated measures were taken over four sessions. On each testing day, subjects were initially asked to perform a series of 3-5 short 2-5 second MVIAs and then 7 randomly applied 2-5 submaximal exertions of 100% -170% HAT in increments of 10%. There was a rest period of at least 2 minutes between exertions and longer rest period of 5-10 minutes to minimize effects of muscle fatigue.

Subjects would then be cued for the fatigue protocol and given standardized verbal encouragement during the effort. On each testing day, subjects would exert one randomly chosen force equivalent to their HAT mass plus a given percentage (0, 20, 40 or 60%) of that HAT mass until volitional failure. The test was terminated if the subject could not maintain the given force as displayed on the screen, or if their torso fell below parallel to the floor (a conclusion criterion only necessary when assessing the 100% HAT

condition). Subjects used the visual feedback of a video monitor that demonstrated the target and actual forces. Electromyographic (EMG) signals and force were all recorded. Individual fatigue tests were separated by a minimum of 48 hrs.

4.3.4 INSTRUMENTATION

Surface EMG was colleted using a bipolar differential collection system (ME3000P; Mega Electronics Ltd, Kuopio, Finland) utilizing 1cm diameter silver/silver electrodes spaced 1 cm apart. This was used to collect the electrical activities of 6 muscles in the trunk and thigh. Channels were sampled at 1000 Hz, band-pass filtered between 20 Hz and 500 Hz and amplified (differential amplifier, common mode rejection ratio 130 dB, gain x 1000, noise 1 μV). They were converted from analogue-to-digital (12-bit), and stored on computer for analysis. Signal amplification was done at the reference electrode site to minimize signal artifacts caused by movements and external noise.

Electrodes were placed bilaterally over the lumbosacral erector spinae (LLES) 2 cm lateral to the L5-S1 spinous processes and over the upper lumbar erector spinae (ULES) 6 cm lateral to the L1-L2, spinous processes. While a number of studies have used the L5/S1 configuration of surface EMG electrodes for examination of multifidus, (Vezina and Hubley-Kozey 2000; Hermann and Barnes 2001; Danneels et al. 2002), others suggest the intramuscular needle electrodes are necessary for accurate assessment (Stokes et al. 2003). For the present study, the EMG activity collected by the electrode arrangement is referred to as LLES as we expect we may have activity from more than just multifidus. In the same way it is expected to emphasize the measurement of the

multifidus at the lumbosacral junction with our narrow electrode placement, we expect to emphasize the longissimus thoracis with our placement of electrodes more lateral to the L1-L2 spinous processes. We are aware that we may also be interpreting signals from iliocostalis lumborum and multifidus and in this paper refer to the observed EMG activity as ULES. Electrodes were also placed bilaterally in the mid-belly of the BF. Reference electrodes were placed 5-10 cm away from the collecting electrodes for all collection arrays.

Bony landmarks and careful palpation was used to place electrodes in the same location. Both skin marking and measurement techniques enhanced the repeatability of electrode placement. The subjects' skin was prepared prior to electrode placement by initially shaving local body hair, removing dead epithelial cells with very fine grade sandpaper and then cleansing the areas with an isopropyl alcohol swab.

Force exerted against the harness assembly placed at the T5/T6 level was collected through a Wheatstone bridge configuration strain gauge (Omega Engineering Inc. 55LCCA 250). The signal was converted from analog to digital (MP100 analog to digital converter; Biopac Systems Inc. Holliston, MA) and stored and analyzed through computer software. (Acqknowlege III, Biopac Systems Inc. Holliston, MA). (Figure 4.1)

4.3.5 DATA ANALYSIS AND STATISTICS

All signals were visually inspected during real time collection of EMG to ensure optimal signal quality. The median frequency (MF) was calculated using the Hamming Fast Fourier Transformation algorithm. This was a data reduction option available from the MegaWin software (Mega Electronics Ltd, Kuopio, Finland) employed in the EMG

data collection and analysis. A spectral estimate was calculated using a 1024 point moving window over the time from the initial marker flag representing the onset of activity to the final marker flag denoting the subject could no longer maintain the horizontal trunk position. The change in median frequency was calculated for the time period (Hz/sec) and employed as an estimate for muscular fatigue. Using the same time markers, the average amplitude of the EMG signal (aEMG) was also calculated. Descriptive statistics were reported for endurance time, change in MF, and aEMG. These measures were compared across the conditions of 100%, 120%, 140% and 160% HAT using an ANOVA of a 2x4 (group x resistance) configuration (SPSS 12.0 for windows, SPSS Inc., US). Data in the text and figures include means and standard deviation (SD).

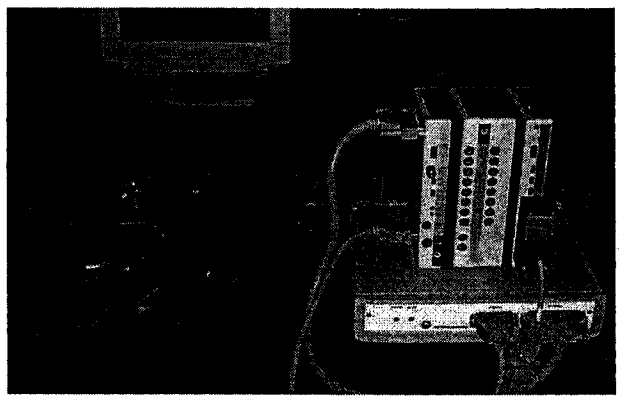


Figure 4.1 Wheatstone bridge configuration strain gauge *(Omega Engineering Inc. 55LCCA 250).* MP100 analog to digital converter; *(Biopac Systems Inc. Holliston, MA)*

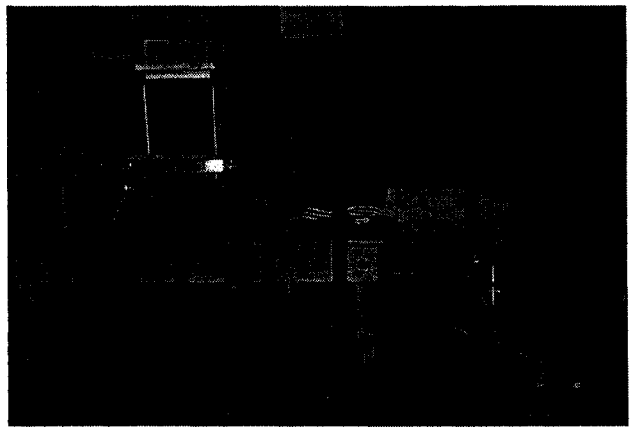


Figure 4.2 Posture for Sørensen Test

Figure 4.3 Harness/strain gauge assembly.

4.4 RESULTS

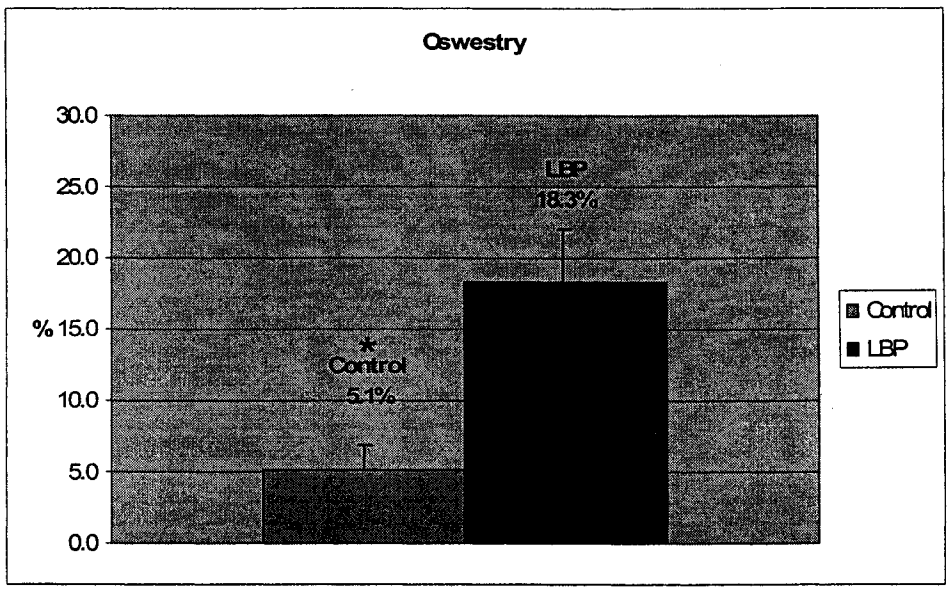
4.4.1 SUBJECT CHARACTERISTICS:

Table 4.1 reports subject characteristics and mean scores of the Oswestry Disability Index and 0-10 Pain scale. Oswestry Low Back Disability Index scores were 73% lower and pain scores 97% lower in the Control group than LBP group. The LBP group had an Oswestry mean score of 18.3% (±11.8), which is clinically categorized as "mild disability" as compared to control group that had an Oswestry of 5.1% (±5.5), which is also considered "mild disability". The mean pain score from the LBP group was 3.43 (±2.0) as compared to that of 0.1 (±0.4) for controls (Figures 3.3 and 3.4). Using the Mann-Whittney Test, significant differences ($p=0.007$) were found between Oswestry Low Back Disability Index scores between LBP and Control groups and significant differences ($p \leq 0.001$) in pain levels between LBP and Control groups.

Table 4.1: Subject Characteristics

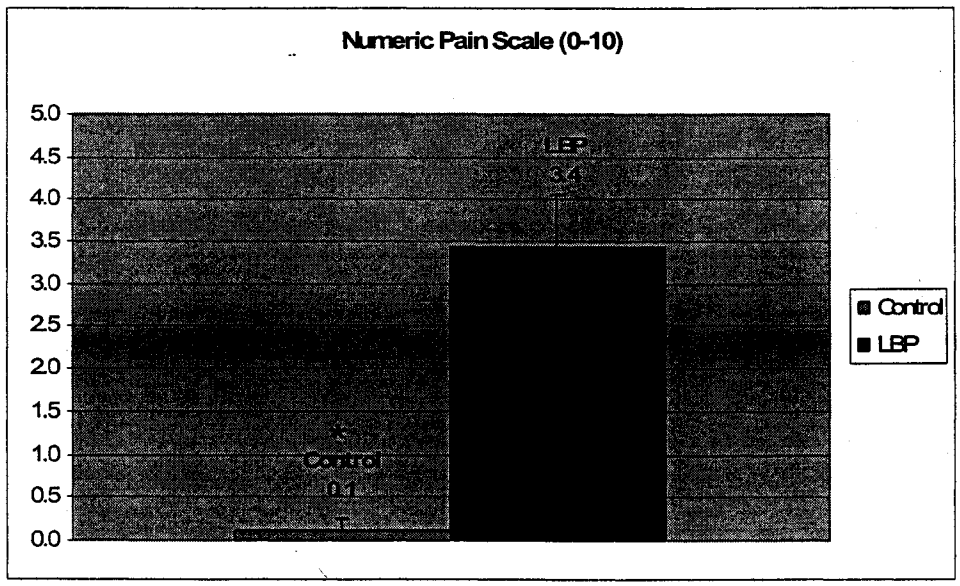
Variable	Group	Mean	sd	*p*
Age (years)	Control	24.7	2.91	0.401
	LBP	29.1	8.26	
Mass (Kg)	Control	79.7	11.17	0.785
	LBP	81.2	7.81	
Oswestry (%)	Control	5.1	5.45	0.007*
	LBP	18.3	12.38	
Pain Scale (x/10)	Control	0.13	0.35	0.001*
	LBP	3.4	1.98	

* Significance at $p \leq 0.05$



*Significance at $p \leq 0.05$

Figure 4.4: Comparison of Oswestry Pain Disability Index scores between LBP and controls

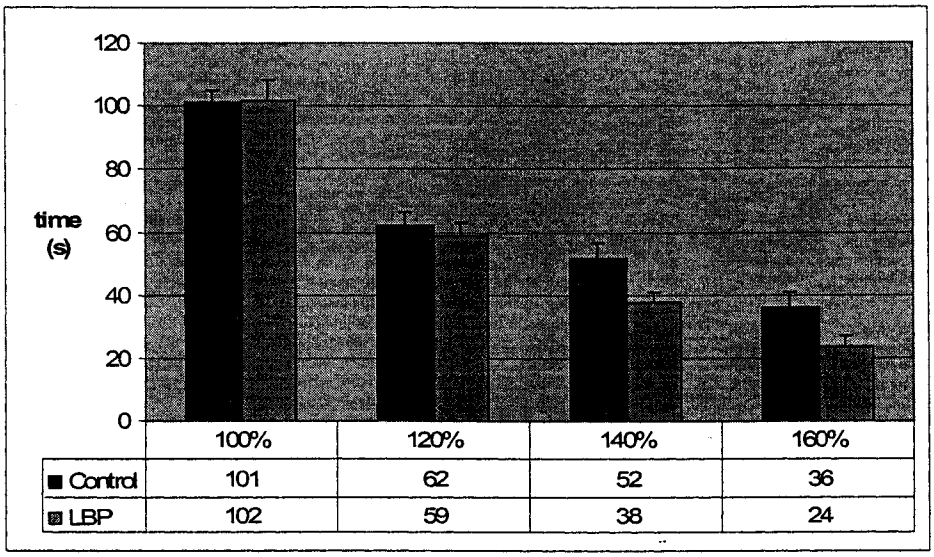


*Significance at $p \leq 0.05$

Figure 4.5: Comparison of visual analogue pain scale scores between LBP and controls

4.4.2 ENDURANCE TIME:

Figure 4.6 depicts the difference in endurance time as resistance increases from 100% to 160% HAT. Expectedly, endurance times decreased as resistance increased. The LBP group had 4.5%, 34.2%, 40.6% shorter times AT for 120%, 140% and 160% of HAT respectively however no significant differences were detected between groups.



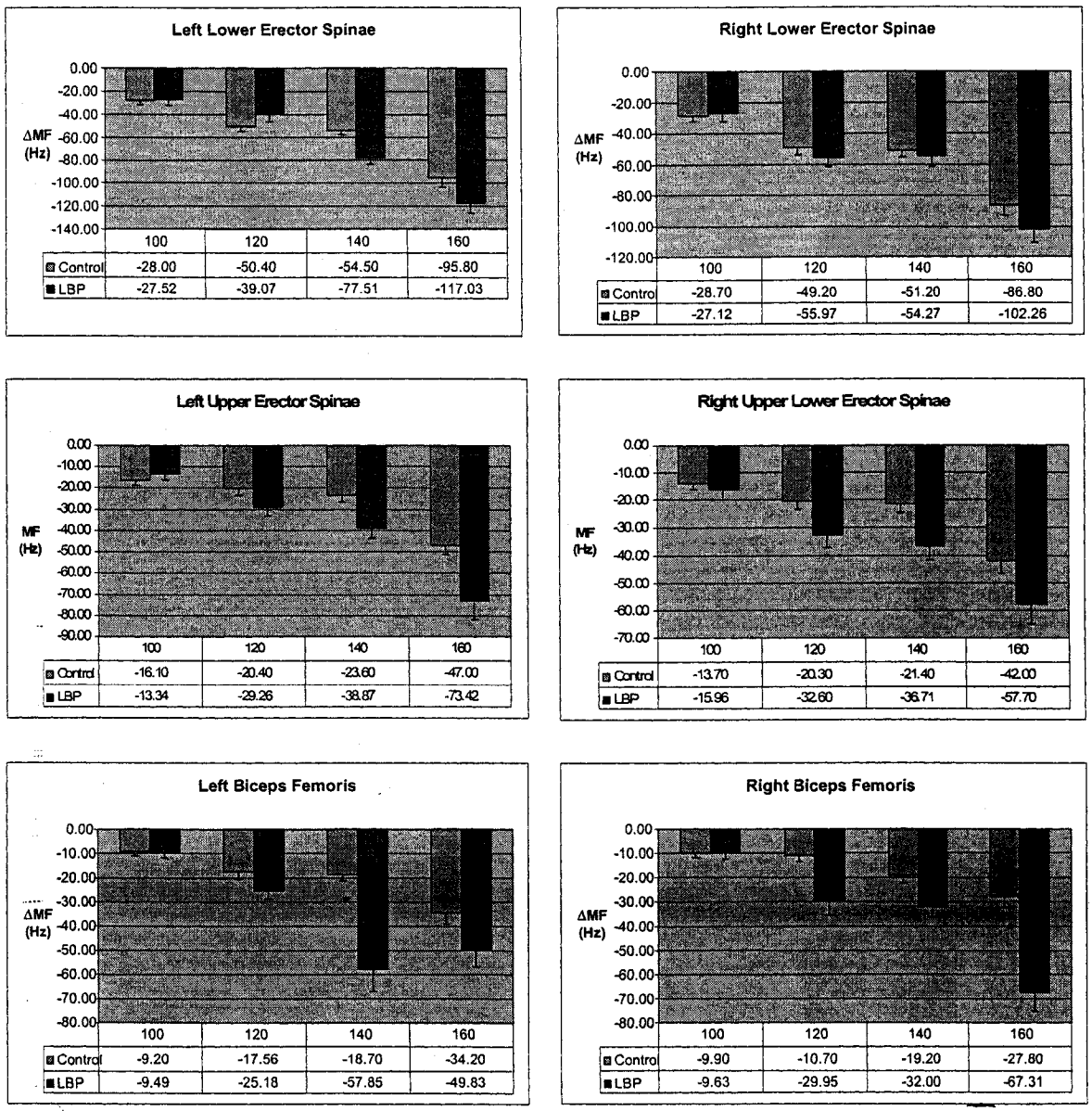
	100%	120%	140%	160%
Control	101	62	52	36
LBP	102	59	38	24

Figure 4.6: Comparison of mean endurance times (in seconds) between LBP and controls at given percentages HAT

4.4.3 MEDIAN FREQUENCY

Figure 4.7 illustrates differences in MF between Control and LBP groups for each extensor muscle group. MF decreased more as resistance increased from 100-160% HAT. Differences were observed only in the biceps femoris and only at higher percentages of HAT. Table 4.2 reports significant between group differences in the right BF. There were significant pairwise differences in the left BF at 140% HAT with 89% lower MF in controls. A significant pairwise difference was also evident in the right BF at 160% HAT

with 77% lower MF in controls and significance was approached (p=0.057) at 120% HAT with 107% lower MF in the control group.



*Significance at $p \leq 0.05$

Figure 4.7: Change in MF for each extensor groups between LBP and controls

Table 4.2: Comparison of p values for MF between LBP and Controls at each percentage of HAT

%HAT	LBP vs. Control	Between groups at each %HAT			
		100%	120%	140%	160%
Left ULES	0.176	0.615	0.233	0.101	0.236
Right ULES	0.267	0.463	0.147	0.086	0.287
Left LLES	0.453	0.850	0.968	0.090	0.402
Right LLES	0.685	0.867	0.708	0.160	0.540
Left BF	0.132	0.631	0.099	**0.037***	0.415
Right BF	**0.004***	0.677	0.057	0.065	**0.037***

***Significance at p≤0.05**

4.4.4 AVERAGE EMG

For the control group, the aEMG consistently increased from 100% to 160% HAT. Table 4.3 reports aEMG means for each group across percentages of HAT. The aEMG was markedly increased in the control group between the 140% to 160% of HAT condition in all extensor muscle groups. In the LBP group the 160% HAT condition only elicited marked changes in the left and right ULES, but failed to show marked differences in other muscles. Table 4.4 reports the interaction between groups and resistance for each muscle group. There was significant difference in the ULES aEMG with 54% less aEMG in control group than in the LBP group. There was a significant difference at 140% HAT in the left BF with 86% lower aEMG in controls. The right BF demonstrated significant

differences, with 65% lower aEMG in controls at 120% HAT and an 81% lower aEMG in controls at 140%.

Table 4.3: Average EMG for each muscle at each percentage of HAT

%HAT		100		120		140		160	
		Mean	sd	Mean	sd	Mean	sd	Mean	sd
Left ULES	Control	24.60	15.88	33.80	21.77	32.90	66.51	159.30	143.41
	LBP	10.79	19.95	6.21	67.23	78.53	56.82	152.26	183.71
Right ULES	Control	31.10	35.24	45.40	41.94	30.20	46.68	172.90	142.94
	LBP	14.24	30.45	35.46	51.25	84.48	30.76	217.43	331.06
Left LLES	Control	14.20	15.25	23.00	17.54	11.10	37.93	74.30	50.91
	LBP	3.46	16.67	15.71	39.95	15.84	32.42	29.80	49.64
Right LLES	Control	12.50	17.43	22.20	18.20	13.50	47.31	93.50	94.20
	LBP	2.01	21.03	18.48	28.76	50.31	68.26	53.51	108.52
Left BF	Control	7.00	15.18	19.13	27.35	-0.12	40.50	119.00	185.11
	LBP	18.71	29.79	58.09	109.67	124.09	139.24	125.27	354.92
Right BF	Control	19.60	24.74	16.80	36.05	23.10	45.06	129.70	186.95
	LBP	13.73	28.33	65.76	63.05	183.51	359.86	53.17	144.73

Table 4.4 Comparison of p values for aEMG between LBP and Controls at each percentage of HAT

%HAT	LBP vs. Control	Between groups at each %HAT			
		100%	120%	140%	160%
Left ULES	0.355	0.112	0.235	1.128	0.926
Right ULES	0.088	0.283	0.648	**0.009***	0.703
Left LLES	0.095	0.161	0.607	0.744	0.071
Right LLES	0.118	0.251	0.737	0.186	0.402
Left BF	0.312	0.333	0.345	**0.028***	0.965
Right BF	0.269	0.636	**0.05***	**0.018***	0.336

*Significance at $p \leq 0.05$

4.5 DISCUSSION

Back endurance as it relates to LBP has received much attention. Currently, a modified Sørensen test is used as part of the Canadian Society for Exercise Physiology (CSEP) Canadian Physical Activity Fitness and Lifestyle Approach (CPAFLA) test. Many studies have demonstrated that differences in endurance times are lower in those with LBP than those without. This study however did not find such a clear distinction. The more rigorous testing procedures outlined in our protocol may account for differences in overall fatigue times, but not in differences between groups. Differences in fatigue responses were observed through electromyographic evidence in select muscle groups at higher resistance of fatigue, but there were no differences at lower percentages of HAT. Further, endurance time did not appear to be a sensitive measure to discern between LBP and Control groups.

4.5.1 ENDURANCE TIME

Although numerically, endurance time was lower in LBP subjects than controls at higher % HAT, there was no significant difference. In fact, in the 100% HAT condition there was a non-significant mean difference of 1 second between groups. These findings are similar to that of Sparto et al. (1997) that measured 10 subjects without LBP with a mean of 109s. In a recent study by McKeon (2006), mean endurance time in the male LBP group was 115.3, but healthy males had a mean endurance time of 124.4. In the current study the initial series of MVIA and submaximal exertions were performed by all subjects thus should not play a role in differences between groups, however even with

adequate muscle recovery periods, the initial testing may account for lower endurance times than found in most studies. Biering-Sørensen (1984) reported endurance time with a mean of 195s controls (n=144) and 164s in subjects with LBP in the previous week (n=164) (Biering-Sørensen 1984). In a study by Hultman et al. (1993) subjects without LBP had a mean endurance time of 150s (n=36), while subjects with pain at least once but not within 2 months of testing had a mean of 134s (n=86). Hultman also measured subjects with chronic LBP and had taken more than 3 months leave within the past year having a mean of 86s (n=18). No subject in the LBP group in this study reported recent severe bouts of LBP within the past month, but all reported recurrent or chronic LBP that was reported to affect their activity. Validated outcome measures and visual analogue pain scales, while significantly different between groups, did not convey a sense of severe pain or marked physical disability. However subjects with similar pain history and ranges of discomfort are likely characteristic of people that are candidates for assessments such as the CPAFLA test.

4.5.2 MEDIAN FREQUENCY

Pairwise differences were only present at higher levels of resistance. Right BF demonstrated no difference in MF at 100%, but significant difference was evident at 120% and 160%. Significant difference was also found at the 140% HAT condition for left BF and right ULES. These findings may suggest that the lower resistance levels are not sufficient to delineate between groups, but as resistance increases, more extensor effort is required and the differences between groups occur primarily in the BF. Significant differences at the right ULES may also play a role.

4.5.3 AVERAGE EMG

The MF decreased as load increased. Since subjects were required to maintain a given force output, as fatigue developed, additional muscle recruitment is required to maintain the force output and increased EMG is observed (Behm 2004). During the fatigue testing, both recruitment and de-recruitment (represented by transient increases and decreases in aEMG activity) occur (Behm 2004). The net result is an increase in aEMG over the duration of the fatigue protocol. Differences in aEMG between groups were evident in the right ULES at 140% HAT. The only other significant differences occurred in the left BF at 140% HAT and in the right BF at 140 and 150% HAT. While the final product of force output through back extension is a composite of many synergistic muscles and recruitment strategies, it appears that the most marked differences in muscle recruitment between groups occurred in the BF at higher percentages of HAT. Based on the results of this study, using aEMG of erector spinae muscles in low resistance Modified Sørensen tests may not be ideal when attempting to discriminate healthy subjects from those with mild chronic or recurrent LBP.

4.5.4 MUSCLE SYNERGYSM

Due to the synergism of muscles used in back extension; there are various motor control strategies that may be employed during a low intensity fatigue test to maintain a desired static posture. It is suspected that at higher intensities (larger percentages of

HAT) there is less time for implementing a motor control strategy that coordinates load sharing across synergistic muscles. This may be the reason why endurance time differences are more pronounced at 140% and 160% HAT. The mean maximum voluntary contraction for the LBP groups was 408N as compared to 622N for the controls. For an 80kg subject, 140% HAT is 540N or 87% of maximum for controls and 132% of maximum for the LBP group. It is probable that at higher percentages of HAT that approach or exceed maximal values, there is less opportunity to employ alternative recruitment strategies.

In an isolated case, one of the control subjects had a higher endurance time at 160% than at the 100% condition. When EMG data streams were reviewed, it was evident that he had developed a load sharing strategy between his lumbar extensors and BF, alternating bursts of activity in each muscle group thus creating "micro-rest periods". This case highlights the idea that although the neuromuscular endurance of the trunk and hip extensors contribute to endurance time, motor control strategies may play an equal or superior role in the application of fatigue protocols.

4.5.5 LIMITATIONS

One of the most significant limitations of this study is having the subjects use self-report of LBP to delineate control and LBP groups. While the differences in the pain and Oswestry scores were significant between groups, there was considerable variability in the scores within the LBP group. Such variability may have reduced the discrimination between groups. Additionally, it should be noted that an Oswestry score of 18% classifies a subject as having only mild lower back disability. For future studies, it is suggested

that scores or other form of external assessment be used as grouping criteria groups independent of self classification as back pain sufferers or not. There were some limitations in the research design. Firstly we used a relatively small number of subjects with each group containing 10 subjects. Secondly, a series of maximal and submaximal tests were performed prior to the fatigue protocol. Although an adequate recovery times were used, this could have potentially led to shorter endurance times. Because this was done consistently on each session and for all subjects, it is not a factor influencing differences between groups.

4.6 CONCLUSIONS

Prone isometric back extension is frequently used as an assessment tool for LBP and has been suggested to have value as a predictive measure for first time LBP. Although the majority of studies show the Sørensen test as useful test of back endurance, the results in this study do not wholly support the modified Sørensen test utilizing resistance of 100% HAT to discern differences in endurance in subjects with mild LBP. No significant differences in endurance time between groups at 100% HAT or even at higher resistance levels are reported. Differences were evident with analysis of components including aEMG and MF, but only at higher percentages of HAT and predominantly in the BF.

The subjects with LBP participating in the present study had current subacute LBP or a history of recurrent or chronic LBP. Subjects in the LBP group generally did not have high pain levels and consideration must be given the variation in range of disability indices scores. Although the relatively low levels of disability and pain are a likely cause for decreased differences between groups, it can be argued that clients with similar pain and disability characteristics are likely candidates for conservative care treatment and likely to present to kinesiologists or trainers for fitness appraisals such as the CPAFLA.

The idea that load sharing strategies may be employed by a subject to increase endurance times, as was observed in this study is also important. The possibility exists that subjects with more sophisticated strategies could yield higher endurance times despite inferior neuromuscular endurance and the existence of LBP. Future research designs that evaluate motor control strategies during prone extension could yield important information for further design of assessment tools and rehabilitative procedures.

Chapter 5
Summary

This study was successful in gaining further understanding of the electrophysiology associated with performing the Modified Sørensen test both in its typical use evaluating fatigue, and also in its application for normalizing EMG profiles for comparative purposes. Trunk strength and endurance are thought to be important measures and even predictors for LBP. The findings in this study demonstrate significant strength differences, as indicated by force and EMG activity, between subjects with and without LBP. However little difference is found between groups in neuromuscular fatigue measured with time and analysis of EMG spectral contents.

Hypothesis 1 stated: EMG and force values elicited with MVIA will be less reliable for the LBP group than control subjects across four testing sessions. The control group demonstrated better reliability in EMG and force measures with MVIA than the LBP group, supporting hypothesis number 1 Although normalization techniques frequently use MVIA values in both healthy and clinical subjects, submaximal values, as those elicited using %HAT, had higher ICC's and appear more reliable than maximal efforts in the LBP group supporting hypothesis number 2 which predicted EMG values would be more reliable in the LBP group when the test is performed at set submaximal values with visual feedback.

Overall, resistance equivalent to 100% HAT yields higher overall EMG reliability for both groups than use of MVIA. Using percentages of HAT may be a safer and more reliable method to attain normalization values for trunk extensors.

During maximal efforts ULES demonstrated excellent reliability in the control group but significantly less in the LBP group. EMG activity of the BF was consistent in both LBP and control groups with discrepancy in reliability most marked in the ULES. This may suggest that the marked difference in force output between groups is more attributable to the ULES than BF.

Hypothesis 3 stated: Endurance time would be higher in the control group as compared to the LBP group. There were however no significant differences in endurance time between groups when using the Modified Sørensen test in its traditional configuration without additional resistance and at resistances higher than 100% HAT and does not support hypothesis number 3. Spectral content demonstrates fatigue only at high levels of resistance and predominantly in the BF. It is suspected that the longer duration of testing with fatigue protocols allows for alternate motor control strategies and differences in BF spectral contents between groups are a result of variation in the ability to execute motor control alternatives. There were increased differences between controls and LBP at higher resistance, but differences were not significant and thus fail to support hypothesis number 4 which stated differences in endurance time between LBP and control groups would be larger with higher muscle output demands.

The findings of this study raise additional questions and potential avenues for further research. The use of %HAT values for normalization procedures should be investigated and compared with MVIA as input values in biomechanical models. Evaluation of the extent neuromuscular control strategies play a role in prolonging endurance times naturally follows from this research and further begs the question: Can

training using feedback be used to enhance those strategies and improve lumbar endurance?

Although normalization techniques frequently use MVIA values in both healthy and clinical subjects, submaximal values, as those elicited with using %HAT appear to be reliable in the LBP group. Overall, resistance equivalent to 100% HAT yields higher overall EMG reliability for both groups than use of MVIA. Using percentages of HAT may be a safer and more reliable method to attain normalization values for trunk extensors. Although the sample size is small in this study, it seems that evaluating strength and MVIA with force or EMG may better assess function of the erector spinae than does using time measures with Modified Sørensen testing, in subjects with low back pain of low intensity. It appears that endurance times of Modified Sørensen testing are composites of muscle endurance, synergistic muscle load sharing and motor control strategy.

Chapter 6

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Appendix A

Oswestry Low Back Disability Questionnaire

The Oswestry Low Back Disability Questionnaire is a commonly used outcome tool that allows tolerance to activities of daily living to be assessed quantitatively. This outcome tool includes 10 questions regarding activities such as personal care (washing and dressing), lifting, walking, sitting, standing, sleeping, social life and driving. There are 6 answers to each question that are each assigned scores of 0-5. The test is scored out of 50. The scores are often used as a descriptor of relative disability. The raw score will fall into the following ranges:

0-4	No disability
5-14	Mild disability
15-24	Moderate disability
25-34	Severe disability
35+	Complete disability

The questionnaire is also frequently administered as an indicator of change in a patient's condition, though its sensitivity to real change is poorly understood. The strength of the questionnaire is in its practical relationship to daily tasks, making it an activity intolerance test. The Oswestry Low Back Disability Questionnaire is widely used and is considered a valid and vigorous measure. (Fairbank and Pynsent 2000).

Example of Oswestry Low Back Disability Questionnaire:

Please Read: This questionnaire is designed to enable us to understand how much your low back has affected your ability to manage everyday activities. Please answer each Section by circling the ONE CHOICE that most applies to you. We realize that you may feel that more than one statement may relate to you, but ***Please just circle the one choice which closely describes your problem right now.***

SECTION 1--Pain Intensity
A. The pain comes and goes and is very mild.
B. The pain is mild and does not vary much.
C. The pain comes and goes and is moderate.
D. The pain is moderate and does not vary much.
E. The pain is severe but comes and goes.
F. The pain is severe and does not vary much.

SECTION 2--Personal Care
A. I would not have to change my way of washing or dressing in order to avoid pain.
B. I do not normally change my way of washing or dressing even though it causes some pain.
C. Washing and dressing increase the pain, but I manage not to change my way of doing it.
D. Washing and dressing increase the pain and I it necessary to change my way of doing it.
E. Because of the pain, I am unable to do any washing and dressing without help.
F. Because of the pain, I am unable to do any washing or dressing without help.

SECTION 3--Lifting
A. I can lift heavy weights without extra pain.
B. I can lift heavy weights, but it causes extra pain.
C. Pain prevents me from lifting heavy weights off the floor.
D. Pain prevents me from lifting heavy weights off the floor, but I can manage if they are conveniently positioned, e.g. on the table.
E. Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned.
F. I can only lift very light weights, at the most.

SECTION 4 --Walking
A. Pain does not prevent me from walking any distance.
B. I have some pain with walking but it does not increase with distance.
C. Pain prevents me from walking more than one mile.
D. Pain prevents me from walking more than 1/2 mile.
E. I can only walk while using a cane or on crutches.
F. I am in bed most of the time and have to crawl to the toilet.

SECTION 5--Sitting
A. 1 can sit in any chair as long as I like without pain.
B. I can only sit in my favorite chair as long as I like.
C. Pain prevents me from sitting more than one hour.
D. Pain prevents me from sitting more than 1/2 hour.
E. Pain prevents me from sitting more than ten minutes.
F. Pain pevents me from sitting at all.

SECTION 6 -- Standing
A. I can stand as long as I want without pain
B. I have some pain while standing, but it does not increase with time.
C. I cannot stand for longer than one hour without increasing pain.
D. I cannot stand for longer than 1/2 hour without increasing pain.
E. I can't stand for more than 10 minutes without increasing pain.
F. I avoid standing because it increases pain right away.

SECTION 7--Sleeping
A. I get no pain in bed.
B. I get pain in bed, but it does not prevent me from sleeping.
C. Because of pain, my normal night's sleep is reduced by less than one-quarter.
D. Because of pain, my normal night's sleep is reduced by less than one-half.
E. Because of pain, my normal night's sleep is reduced by less than three-quarters.
F. Pain prevents me from sleeping at all.

SECTION 8--Social Life
A. My social life is normal and gives me no pain.
B. My social life is normal, but increases the degree of my pain.
C. Pain has no significant effect on my social life apart from limiting my more energetic interests, e.g., dancing, etc.
D. Pain has restricted my social life and I do not go out very often.
E. Pain has restricted my social, life to my home.
F. Pain prevents me from social, life at all.

SECTION 9--Traveling
A. I get no pain while traveling.
B. I get some pain while traveling, but none of my usual forms of travel make it any worse.
C. I get extra pain while traveling, but it does not compel me to seek alternative forms of travel.
D. I get extra pain while traveling which compels me to seek alternative forms of travel.
E. Pain restricts all forms off travel.
F. Pain prevents all forms of travel except that done lying down.

SECTION 10--Changing Degree of Pain
A. My pain is rapidly getting better.
B. My pain fluctuates, but overall is definitely getting better.
C. My pain seems to be getting better, but improvement is slow at present.
D. My pain is neither getting better nor worse.
E. My pain is gradually worsening.
F. My pain is rapidly worsening.

Disability index score: ____/ 50 = ____%