Fatigue response of the spine to asymmetrical lifting and lowering.

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UMI
Fatigue Response of the Spine to Asymmetrical Lifting and Lowering

by

Derek Fredrick Fraser

A Thesis
Submitted to the College of Graduate Studies and Research through Human Kinetics
in Partial Fulfillment of the Requirements of
the Degree of Master of Human Kinetics at the
University of Windsor

Windsor, Ontario, Canada

2000

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0-612-52551-1
ABSTRACT

Low back disability due to manual material handling continues to be the leading cause of lost productivity for humans in the work force. While both asymmetrical lifting and repetitive lifting have been linked to an increase in the risk of low back injury, few studies have been performed biomechanically to assess if there are changes in spine mechanics with asymmetrical lifting over prolonged periods of time. Thus, the purpose of the current study was to monitor trunk muscle electromyography (EMG) and spine kinematics during repetitive asymmetrical lifting and lowering involving spine flexion, lateral bend and axial twist and to determine if there is a change in the risk of low back injury associated with fatigue. Ten male subjects were studied. Each was instrumented with a kinematic devise at the sacrum, as well as surface EMG electrode pairs to monitor the activity of the bilateral lumbar erector spinae (LES), thoracic erector spinae (TES), internal oblique (IO) and external oblique (EO) muscles. EMG signals were normalized to maximum voluntary contractions (MVCs) performed at the beginning of the session. Subjects were asked to perform lifts, with a load of approximately 11 kg, from a platform at knee height to a platform at shoulder height. Lowers were performed in the opposite direction. The platforms were spaced three meters apart and 4 lifts and 4 lowers were performed each minute for a total of 40 minutes. EMG data were sampled at 1000Hz, rectified and low pass filtered to approximate the relative muscle force. Data were pooled across the first and last 10 lifts (representing the "rested" and "fatigued" states) and for the first and last 10 lowers. Data were analyzed to determine the peak EMG for each channel. The average of the ten repetitions were averaged for each variable and each condition. A two way ANOVA with repeated measures was used to
determine the main effects of time (rest and fatigue) and direction (lift and lower) and their interaction. Significance was set at \( p<0.05 \). One of the findings of this study was that the asymmetrical lifting and lowering task decreased the trunks maximum extensor moment. The finding was based on the expectation that the trunk muscles would fatigue and result in a decrease in maximum extensor moment. To a certain degree, all subjects' maximum extensor moment decreased when comparing the pre and post trials and the group mean was significantly reduced. The protocol was sufficient to cause fatigue but the kinematic and EMG data during the dynamic session results do not support a consistent effect of fatigue on each subject. This was shown with an average increase in flexion, when comparing rest and fatigue, but not all subjects decreased flexion to the same degree. As well, the muscle recruitment pattern for the trunk muscles also varied for each subject. Lifting was observed to result in higher EMG amplitudes than lowering but it is not likely that there were substantial differences in the erector spinae muscle forces given the force-velocity relationship of muscle.
Acknowledgments

I would like to thank my committee members; Dr. Dutta, Dr. Marnio and Dr. Potvin for their time and work they contributed to my thesis. I would especially like to thank Dr. Potvin for his patience and guidance during the course of my research.
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List of Nomenclature

IO: Internal Obliques
EO: External Obliques
TES: Thoracic Erector Spinae
LES: Lumbar Erector Spinae
TCA: Test Contraction Apparatus
MVC: Maximum Voluntary Contraction
EMG: Electromyogram
CHAPTER I
INTRODUCTION

Low back disability due to manual material handling (MMH) continues to be the leading cause of lost productivity for humans in the work force. Seventy percent of the working population have had lower back pain at some time during their working lives (Pope et al 1986). Occupational low back injuries cost the US industry between 4.5 and 38 billion per year (Andersson, Pope, Frymoyer, & Snook 1990). Low back injuries due to MMH tasks have also been an increasing research topic in ergonomics. Muscle fatigue, repetitive bending and twisting motions during MMH tasks are known to be major causes of low back disorders (van Dieen et al 1993, 1998 & Ayoub 1992). Still, the understanding of the biomechanical factors associated with mechanical loading of the torso is limited.

The mechanical overload or overuse of lumbar spine tissues has led to the development of mathematical models to determine the size of forces during MMH tasks. The combination of forward flexion and compression has been shown to cause the intervertebral disc to prolapse posteriorly, either by a sudden extrusion of the nucleus pulposus into the spinal canal or gradually in response to fatigue loading (Adams & Dolan 1991). Anterior shear forces in normal MMH movements are routinely in the order of 400N but in industrial load handling, shear forces can be much higher because of the introduction of a load to be lifted and the accelerations of that load and the upper body (Potvin, Norman and McGill 1991).

As well, different types of motions can increase the risk of injury to lumbar spine tissues. In addition to spine loading and flexion, twisting motions will cause extra stress on the invertebral disc. Further, it will increase activation of the abdominals and erector muscles as loads become
more asymmetric (Seroussi and Pope 1987). This simultaneous activation of agonist/antagonistic muscles is referred to as coactivation. Even though coactivation increases the stiffness of the joint, thereby increasing stability, the coactivation of the muscles surrounding a joint increases the mechanical load (Lavender et al. 1992). It should also be noted that as early as 1905, Lovett observed that axial rotation and lateral bending motions do not occur independently at the individual vertebral level because they work more as a column (Cholewicki et al. 1996). The introduction of lateral bending motions has been shown to also increase the risk of injury to lumbar spine tissues. Marras et al. (1993) identified trunk lateral velocity as one of the five risk factors for the development of low back disorders. Average lateral bend velocity increased the odds of being classified as high risk (Marras et al. 1993).

With lower back injury studies, the act of lifting is the most observed movement but Lamonde (1987) showed that 2 out of 3 injuries occur during lowering activities. In lifting and lowering similar body positions, muscle coordination and net joint moments are produced but in lowering, these forces are distributed over a smaller cross-sectional area of active muscle, which may increase the chance of injury (de Looze, Toussaint, van Dieen, & Kemper, 1993). de Looze et al. (1993) also concluded that repetitive lowering requires less metabolic energy than lifting. Consequently, workers in manual material handling are very aware of the reduced metabolic stress in lowering and underestimate the biomechanical stress, which may increase the risk of injury.

Not all MMH injuries to the spinal muscles and ligaments are due to acute loads that exceed tissue tolerance limits. Injuries with loads that are under the acute tolerance limits are related to fatigue due to repetitive movements. Biological tissues that are subjected to wear and tear and repeated submaximal applications may develop cumulative trauma which reduces their
stress bearing capacity (Kumar 1990). Consequently, repetitive lifting has been shown to be a risk factor for the development of low back pain (Garg and Moore 1992).

Potvin and Norman (1993) have shown that repetitive lifting can cause a rapid development of back extensor muscle fatigue. Even with repetitive lifting of a constant load, the sharing of the load amongst tissues changes with fatigue and overloads a tissue that would normally not be used (Potvin 1992). This change in lifting technique due to repetitive lifting has also been shown and discussed by van Dieen et al (1998). The changes in trunk kinematics and hip trunk coordination are a consequence of the inadequate control caused by back muscle fatigue. It is assumed that under rested conditions, motor control of the lumbar spine is optimized or constrained such that the resulting kinetics and kinematics for the given motor action involve a relatively low risk to injury (van Dieen 1998). However, muscle fatigue can impair coordination and increase the risk of musculoskeletal injuries (Roy, DeLuca & Casavant 1989).

Neuromuscular fatigue also appears to affect the stability of the spine. Back extensor fatigue may lessen the muscles capacity to generate force and maintain stability of the spinal column and control of spinal movement. This in turn may also increase stress on the passive tissues of the spine and any severe perturbation in load may not be attenuated as well by the muscle in this fatigued condition (Sparto et al 1997).

The findings of Sparto et al (1997) would have even more serious implications during asymmetrical lifting. During dynamic asymmetrical movements there is substantial coactivation of the antagonist lumbar muscles (Thelen, Schultz & Ashton-Miller 1995). This substantial coactivation implies that when a load is applied externally, the mechanical forces acting on the spine will be generated primarily by the muscles supporting the torso rather than the external load
itself (Lavender et al 1992). The biased load distribution during asymmetric lifting disadvantages the contralateral muscles and escalates the fatigue accumulation of the muscle (Kim and Chung 1995). Parnianpour et al (1988) have shown the effects of fatigue with “isoinertial” trunk flexion-extension movements in the sagittal plane. They found that the range of movement of the secondary planes of motion (axial twist and lateral bend) increased when the subjects became fatigued. This signified that the stability of the spine had been diminished by the loss of muscular coordination. When considering an asymmetric lifting task, the onset of fatigue may serve to further increase triaxial rotations of the spine putting it at an even greater risk of injury.

In summary, fatigue has been shown to change spine mechanics during sagittal lifting and these changes have been associated with an increased risk of injury. Even though lifting and lowering have similar body positions, muscle coordination and net joint moments, the reduced metabolic energy required in lowering may consequently have a worker underestimate the biomechanical stress, which may also increase the risk of injury. Evidence has further shown that asymmetrical lifting can predispose the low back tissues to injury but no studies have been published to assess whether repetitive, asymmetrical lifting and lowering are associated with changes in lifting mechanics that may precipitate injury.

1.1 Statement of Purpose

The purpose in this investigation was to study the effects of fatigue during repetitive, asymmetrical lifting and lowering. Trunk muscle EMG, trunk kinematics and hand forces was used to evaluate the effects of fatigue and the difference between lowering and lifting. For the purpose of this investigation, fatigue was defined as a progressive loss in force generating capacity of the total neuromuscular system regardless of the force required by the task (Simonson
and Weiser, 1976; Bigland-Ritchie and Wood, 1984). As such, fatigue was considered to be a continuous process and not a single event which must be defined to occur when the task can no longer be performed (Edwards, 1981).

1.2 Statement of Hypotheses

1) It was hypothesized that repeated asymmetrical lifting and lowering would result in fatigue of the muscles of the trunk.

2) It was hypothesized that repeated lifting and lowering would precipitate numerous changes in lifting mechanics that can be hypothesized to increase the risk of low back injury.

3) It was hypothesized that the effects of repeated load handling would be more pronounced during lowering.
Chapter II
REVIEW OF
LITERATURE

2.1 Risk Factors with MMH

MMH activity has been shown as a risk factor for low back disorder (LBD) with industrial work (Andersson 1981). In particular, lifting (Fogleman and Smith 1995), lowering (deLooze et al 1993), twisting (McGill 1991), and repetition (van Dieen 1996) have been shown to be associated with LBD. However, most studies have been done in laboratories which may overlook causal factors that may be experienced by the worker. Marras et al (1993) explored dynamic trunk motions along with the previously mentioned risk factors under in vivo conditions. They showed that there is considerable three-dimensional trunk motion occurring in most industrial tasks and the assumption, such as those in current lifting guidelines, of sagittally symmetric, slow, smooth lifting are not consistent with the types of motions that are experienced in the work place. Secondly, Marras et al (1993) were able to identify high-risk groups as a function of both workplace and trunk motion factors. Five factors which are highly correlated to occupationally related risk factors are: moment, lift rate, lateral trunk velocity, sagittal trunk angles and trunk twisting velocity. Individually, these factors cannot predict the probability of high-risk groups but when these factors are considered in combination, the capability to predict high risk situations is greatly improved.

Marras and Mirka (1992) have also shown the significance of three dimensional trunk motion and its effects on the spine. Increased trunk motion during lifting can accentuate spine loading due to the coactivation of agonist/antagonist musculoskeletal system. As well, an increase
in trunk velocity will increase the degree of trunk muscle coactivation. This will magnify the loading on the spine because the muscles work against one another. Also the increasing of trunk velocity during asymmetrical lifting significantly increases lateral shear forces on the spine (Marras et al 1993).

The vertebral discs ability to tolerate strain also decreases under loading conditions. The lack of facet structural support that occurs while in flexed postures when the trunk is bent laterally, increases fiber strain (Broberg 1983). This increase in fiber strain is similar to the reduced tolerance experienced when the trunk is bent forward in combination with twisting (Gunzburg et al 1991).

2.2 Squat vs Stoop Lifting

With manual material handling, internal and external forces act on the spine. External forces generated by a hand held object creates an external moment about the spine and the internal forces cause reaction or counter moments from the trunk muscles (Marras and Mirka, 1990). As well, the amount of flexion in the lumbar spine governs the size of the contributions from both the passive structures (ligaments and the disc in bending) and the active musculature that is necessary to satisfy the extensor moment demand (Potvin et al 1991). In neutral lordosis, ligaments are not strained and muscular sources bear full responsibility for moment generation. But as the trunk progressively flexes, the ligaments are recruited and ultimately restrict the joint motion when the end range is reached (White and Panjabi 1978).

During manual material handling, there can be considerable functional differences in lifting styles. Stoop and squat lifting are considered to be functionally different styles of lifting. Squat lifting requires the subject to bend at the knees with their back erect and the majority of the work
being performed by the legs (Garg and Herrin 1979). On average trunk flexion during a squat lift is approximately 64 degrees, with the dominant contribution of 40 degrees from the lumbar spine (Potvin et al 1991). In a stoop lift posture the legs remain straight and the subject bends at the waist. Even with the increase of flexion of 48 degrees observed in stoop lifts, only 11 degrees occur at the lumbar spine as most of the increase in flexion occurs at the hips (Potvin et al 1991). This indicates that the increase of total trunk flexion is dominated by hip rotation rather than spinal rotation.

When considering EMG activity of the extensor musculature, the lumbar EMG is higher than the thoracic EMG for squat lifts and lower than thoracic EMG for stoop lifts (Potvin, McGill and Norman 1991). The increased lumbar EMG for squat lifts when compared to a stoop lifts is probably related to the reduced demand on the lumbar muscles during stoop lifting because of the additional moment provided by the ligaments. Also, the increase in thoracic EMG for stoop lifting may have resulted from the increased flexed posture resulting in higher moments around the thoracic joints (Potvin, McGill and Norman 1991).

2.3 Anatomy

When looking at control of a movement it is important to distinguish which muscles are involved in asymmetrical lowering and lifting movement. The trunk muscles that will be studied in this research are the abdominal muscles and trunk extensors. Information on the origin and insertion of abdominal muscles and latissimus dorsi is detailed by McMurtrie and Rikel (1991) and information on the origin and insertion for the erector spinae is detailed by Bogduk and Twomey (1987).
**Abdominal Muscles:** The three abdominal muscles active in asymmetrical lifting are the internal oblique (IO), external oblique (EO), and the rectus abdominus (RA).

**Internal Oblique:** The IO originates at the inguinal ligament, iliac crest, and lumbar oponeurosis and attaches to the lower 3 or 4 costal cartilages, linea alba, tendon of pubis, crest of ilium and lumbar fascia.

**External Oblique:** The fibers of the EO are opposite in angle to the IO on the same side of the body. EO originates on the lower 8 ribs at the costal cartilage and inserts into the crest of the ilium, linea alba through the rectus sheath, pubic crest and Poupart’s ligament.

**Rectus Abdominus:** The RA originates at the Pubic crest and symphysis. It then inserts into the xiphoid process and the 5th, 6th, and 7th costal cartilages.

![Abdominal Muscles Diagram](image_url)

**Figure 1.** The abdominal musculature (only right side shown). A: Rectus Abdominis, B: External Oblique, C: Internal Oblique. From Stone and Stone, 1990.
**Back Muscles**: The back muscles involved with asymmetrical lifting are the latissimus dorsi (LD), the lumbar erector spinae and thoracic erector spinae.

![Image of back musculature](image)

*Figure 2. The posterior trunk musculature. A: left and right thoracic lumbar erector spinae, B: left lumbar erector spinae. From Bogduk and Twomey, 1987.*

**Latissimus Dorsi**: The LD originates on the spines of the lower thoracic vertebrae, spine of lumbar and sacral vertebrae through attachment to thoracolumbar fascia, tip of the iliac crest, lower ribs and inferior angle of the scapula. It inserts into the medial lip of the intertubercular groove of the humerus.

**Erector Spinae**: The erector spinae lies lateral to the multifidus and forms the prominent dorsolateral contour of the back muscles in the lumbar region. It consists of 2 muscles: iliocostalis lumborum and the longissimus thoracis. They are then even further broken down into
the lumbar fascicles; which consists of fascicles arising from lumbar vertebrae and thoracic; fascicles arising from the thoracic vertebrae or ribs.

Lumbar Erector Spinae: The lumbar longissimus consists of 5 fascicles with each rising from the accessory process and the adjacent medial end of the dorsal surface. The lumbar iliocostalis has 4 overlying fascicles which each fascicle arising from L1 through L4 vertebrae transverse processes and to an area extending 2-3cm laterally to the middle layer of the thoracolumbar fascia. The fascicles then insert to the iliac crest. The fascicles form overlapping layers with the fascicle originating at L5 as the deepest and most medial layer. Succeeding fascicles lie superficial and lateral to the L5 fascicle. The fascicles lie inferiorly and slightly lateral.

Thoracic Erector Spinae: The thoracic longissimus consists of 11 or 12 pairs of small fascicles arising from the ribs and transverse processes of T1 down to T12. Each rostral tendon extends 3-4 cm before forming a small belly. Each muscle belly forms the caudal tendon that extends into the lumbar region. Fascicles from T2 level attach at L3 spinous process, while the remaining fascicles insert into spinous processes progressively lower. The thoracic iliocostalis consists of fascicles from the lower seven or eight ribs that attach caudally to the ilium and sacrum. Each fascicles arises from the angle of the rib via a ribbon-like tendon measuring some 9-10 cm in length and then forms a muscle belly. These fascicles then continue as tendons, contributing to the erector spinae oponeurosis, and ultimately attaching to the posterior superior iliac spine.

2.4 EMG-Force/Fatigue

Electromyography (EMG) allows us to estimate the dynamics of limb movements and to examine the physical mechanisms underlying neuromotor control. This led to testable predictions
for the neural control of limb movement (Zernicke & Smith 1996). Three applications dominate the use of the EMG signal: its use as an indicator of the initiation of muscle activation, its relationship to the force produced by a muscle, and its use as an index of fatigue processes occurring in the muscle (De Luca 1997).

Within muscles, motor units are recruited according to the size principle. The size principle states that for low force contractions, smaller motor units are recruited and as more force is required, more motor units are recruited in order of size, with the largest motor units being the last ones recruited (Henneman et al 1965). This increase in recruitment of motor units causes an increase in EMG amplitude. The neuromuscular system can also increase force by increasing the rate of stimulation of the motor units, which in turn will increase EMG amplitude (De Luca 1997).

In isometric studies, the EMG-force relationship is linear for some muscles and non-linear for others. Petrofsky et al (1982) showed that there was a linear response in the handgrip muscles. The EMG amplitude increased as the tension increased. As well De Luca (1997) compared EMG-force relationship and found that there was a non-linear responses for the biceps brachii and the deltoid but a linear response for FDI.

Surface EMG can also be used as an indicator of fatigue. Kogi and Hakamada (1962) showed that when a muscle in isometric contraction began to fatigue, there was an increase in the amplitude in the low frequency range and a reduction in the amplitude in the high frequency range. This was shown as a decrease in the mean power frequency. Petrofsky et al (1982) also showed that as the muscle fatigued in isometric contraction, there was an increase linearly in amplitude for some muscles (handgrip) and non-linearly for other muscles (biceps brachii). One
reason that Petrofsky et al. gives for some muscles being non-linear is that the fibre composition is different.

Even though isometric contractions have been used as a standard for EMG-based fatigue quantification, recent evidence suggests that fatigue may also cause changes in EMG signals measured during dynamic contractions (Potvin & Bent 1997). Ament et al (1993) and Potvin and Bent (1997) are two of the few attempts to compare fatigue related EMG changes during isometric and dynamic contractions. Potvin and Bent compared biceps brachii MPF values from isometric and dynamic contractions during repetitive elbow flexion-extension task. They reported that the MPF is an effective variable for quantifying localized muscle fatigue because it is sensitive to modified force levels associated with changes in technique. Possible reasons for an increase in EMG amplitude when dealing with fatigue is that it may be attributed to new motor units recruitment. As well, fatigue may increase the synchronization of motor unit firing and subsequently increase EMG amplitude (Potvin and Bent 1997).

2.5 Electromyography of Axial Asymmetrical Movement

Many studies have used EMG to describe asymmetrical movement. Pope et al (1986) used a 3-D model of the trunk to determine the activity of the muscles during axial twist around the level of the 3rd lumbar. The subjects were instructed to perform an isometric twist MVC. A dynamometer was used for measuring the torque that was generated and surface EMG were attached to the RA, contralateral EO, ipsilateral IO and lumbar region of the erector spinae. This model allowed for the observation of both agonist and antagonist muscle activity. The proportions of activity of the muscles during the total generation of axial twist were ipsilateral IO 59%, erector spinae in the lumbar region 7%, RA 14% and contralateral EO 20%.
McGill (1991) did a similar study using 3-D modeling of the trunk. Pairs of surface EMG were attached to the right RA, EO, IO, LD and erector spinae in the thoracic and lumbar regions. The subjects were instructed to perform an isometric twist MVC in both the clockwise and counterclockwise direction. A Cybex II dynamometer was used for measuring the torque that was generated. McGill found that the ipsilateral IO contributed to 33% of total twist moments, the ipsilateral LD 11%, the erector spinae in the lumbar region 8% and the thoracic 3%, while the contralateral EO 42% and RA 3%.

Loads from the body itself (as in the previous studies), or externally applied loads that are not symmetric in relation to the midsagittal plane, can show that the internal muscle force requirements that are necessary for stabilizing the torso are not shared evenly between the left and right sides of the body (Lavender, Tsuang, Hafezi, Andaresson, Chaffin, & Hughes 1992). Under asymmetric loading conditions, the internal force contributions are complex. Biomechanically the torso needs to account for the ipsilateral and contralateral muscles with respect to the load position (Lavender et al 1992).

This simultaneous activation of opposing muscles is referred to as cocontraction. During twist exertions, antagonist muscles tend to have lower activation levels than the agonists (Thelen et al 1995, Potvin & O'Brien, 1997). In isometric studies, cocontraction has been shown to increase the stiffness of a joint thereby allowing increased stability but the cocontraction of muscles surrounding a joint increases the mechanical loading of the joint (Humphery and Reed 1983).

Chiang (1998) showed that during dynamic response of the spine to perturbations causing rapid lateral bending, the mechanical properties of the spine are constantly adapting to the
environmental conditions with an overall goal of maintaining stability. Peak angular displacement of the trunk was used as an analogue of spinal stability with the presumption that less angular displacement would be seen in more stable conditions but peak angular displacement was found to be inversely related to the initial load suggesting that, for conditions with lower initial loads, the spine was less stiff and, presumably, less stable. Krajcarski et al (1999) did the same with rapid flexion and found that the dynamic stability of the spine is consistent with Chiangs (1998) study of rapid lateral bend. The study suggested as well that the pre-activation during trunk loading and trunk muscle co-contraction may play an important role in enhancing spine stability.

2.5b Asymmetrical Lifting

Low back disorders have been linked to occupational activities which require lifting activities that are accompanied by twisting motions (Marras et al 1992). The addition of asymmetrical movement is anticipated to put large torsional loads on intervertebral discs and facet joints in addition to the large compression and shear loads already acting on these structures (Lavender 1993). The supporting structures within the torso are altered by asymmetrical postures even though the range of axial rotation in the lumbar spine is limited (Gunzburg et al 1991). This makes it very difficult to analyze as the recruited trunk muscles must support the external loads, twist the torso and stabilize the spine.

Dynamic models have been developed using EMG measures to estimate the force in the trunk muscles (McGill 1992). These models indicate that muscles are the most important in generating lateral-bending moments, but accomplish this assignment with moderate levels of co-contraction. Even with low antagonistic activity, such as 8%, MVC produces quite significant
compressive loads on the spine. Between antagonist and agonist counterparts, the additional 8\% MVC activation level is required in the agonist just to balance the moment at zero (McGill 1991).

The myoelectric patterns and the predicted muscle forces suggested by the dynamic model shows that the motor control system increases the mechanical stability of the spine by increasing the bending stiffness but does so at the expense of additional compressive load. The latissimus dorsi is active, and is in fact, sometimes more active on the ipsilateral side to stabilize the load-supporting shoulder. This type of co-activation suggests that there is a significant compressive penalty to the lumbar spine.

2.6 Spine Stability

In order to understand stability, many researchers have modeled the ligamentous spine as a slender column. Stability of this column has been defined by Euler, a Swiss mathematician in the 18th century, as “a column carrying a concentric compressive load is said to be stable, if the column returns to its original vertical position after it is perturbed, by a lateral force for example” but this column when subjected to a concentric compressive load will fail when stresses in the column exceed the failure stress of the material (Crisco & Panjabi 1992).

Lucas and Bresler (1961) considered an elastic rod clamped to a solid base at the bottom and subjected to a small vertical load. If a small lateral force is added in addition to the vertical load, a lateral displacement is produced and when the lateral force is removed the rod returns to its original position. This elastic equilibrium is called stable equilibrium. An unstable equilibrium occurs when a vertical load reaches a certain magnitude which will produce a lateral displacement which cannot be recovered merely by removing the lateral force. A very slight further increase in the vertical load will cause a sudden lateral bending of the rod without addition lateral force and
this phenomenon is called stability failure, or buckling and the vertical load at which this failure occurs is called the critical load.

This can be seen when an isolated thoracolumbar spine buckles under compressive loads exceeding 20N (Lucas and Bresler 1961) and the lumbar part of the spine buckles under 90N (Crisco, Panjabi, Yamamoto, & Oxland 1992). But in vivo the spine may experience compressive loads of 6000N which reveals how the motor control system and the osteoligamentous spinal linkage must operate within the range of mechanical stability (Cholewicki and McGill 1996).

It is the role of the trunk musculature to support the spine by increasing the stiffness, much like a guy wire spanning a bending mast (Bergmark 1989). In extreme ranges of motion, the ligaments will also be called on to support the spine. Crisco and Panjabi (1991) have shown that the bulk of the spinal column's stiffness is provided by the large muscles spanning between the ribcage and pelvis.

Muscle contractions not only increase spine stiffness but also lead to increased joint compression. Cholewicki and McGill (1996) showed that the stability of the human lumbar spine seemed to increase during the most demanding tasks, as defined by the joint compression force, and diminished during the periods of low muscular activity.

While the large global muscles spanning between the pelvis and ribcage provide the dominate spine stiffness to the spinal column, activity of short, intrinsic local muscles that span only one or two joints are necessary to maintain stability of the whole lumbar spine. Increasing the activity (stiffness) of small muscles can increase passive joint stiffness and help to prevent instability. The increased activity of small muscles highlights the importance of motor control to
coordinate muscle recruitment of large musculature and small intrinsic spine muscles when handling loads.

The stability of the spine will increase as the moment demand and joint compression increase. A relationship can be shown in which injury risk, due to the loss of stability, increases with the decreased muscular effort. Thus, it appears that the trunk musculature contributes to

![Diagram of Subsystems]

**Figure 3.** Subsystems responsible for maintenance of spinal stability. Injury or dysfunction to any of the subsystems compromises spinal stability. The Passive subsystem consists of vertebrae, intervertebral discs, and ligaments. The Active subsystem consists of spinal musculature. The Control subsystem consists of neural components which receive and process input, and send out appropriate signals to maintain spinal stability (Panjabi, 1992).

spinal stability by modifying muscle stiffness and contributing to spinal compression. In
asymmetrical tasks, the stability of the spine will be highest when subjects are bent or twisted in such a way that the external load placed on the lumbar spine is acting through the shortest moment arm (Cholewicki and McGill 1996).

Panjabi (1992) hypothesized that the stabilizing system of the spine consists of three subsystems viewed in Figure 3. They conclude that the passive musculoskeletal subsystem includes, vertebrae, facet articulations, intervertebral discs, spinal ligaments, and joint capsules, as well as the passive mechanical properties of the muscles. The active musculoskeletal subsystem consists of the muscles and tendons surrounding the spinal column. The neural and feedback subsystem consists of various force and motion transducers, located in ligaments, tendons, and muscles and the neural control centers. These passive, active and neural control subsystems, although conceptually separate, are functionally interdependent.

The passive subsystem is passive only in the sense that it, by itself, does not generate or produce spinal motions. In the neutral position, the role of the passive system is to interact with the neural subsystem with sensory information. Towards the end ranges of motions, the passive system acts to stabilize the spine.

The muscles and tendons of the active subsystem are the means through which the spinal system generates force and provides the required stability to the spine. The active system also interacts with the neural subsystem by returning information regarding muscle tension and length. The neural subsystem receives information from the transducers from various locations and determines the specific requirements for the goal of stability. Feedback is measured and adjusted in each muscle until the required stability is achieved. These measures are dependent on the variation of lever arms and inertial loads of different masses and external loads. For normal
function of the stabilization system the subsystems must function and interact appropriately and any loss of stability may be due to fatigue or injury to any one of the subsystems.

2.7 Fatigue and its Effect on Spine Stability

Fatigue is an important aspect of increased risk of injury with spine stability. The chain of commands that leads to eventual voluntary contractions involves many steps from the brain to the formation of cross-bridges and fatigue may occur as a result of failure at any link in the chain (Ausmussen 1979). It is expected that as a result of fatigue neuromotor processes are going to change in some way but the exact cause of fatigue is unclear and could be peripheral and/or central.

Central fatigue may occur because of malfunction of nerve cells or inhibition of voluntary effort: action of the sensory pathways on the reticular formation (Wilder, Aleksiev, Magnusson, Pope, Spratt and Goel 1996). This occurs with the feedback loop from fatigued muscles to the reticular formation causing inhibition of voluntary effort. Central fatigue may also be caused by an inhibition of motor areas elicited by nervous impulses from chemoreceptors (Ausmussen 1979). Peripheral fatigue deals more with the structures of the body, more precisely the muscle. Three possible sites for peripheral fatigue exist: the neuromuscular junction and muscle cell membrane (excitation), the calcium release mechanism (activation), and the sliding filaments (contractile processes) (MacLaren, Gibson, Parry-Billings & Edwards 1989).

Besides the question of where fatigue occurs, the major concern with fatigue and spine stability is that there will be a change in movement coordination. The question is whether the adaptability of the neural control is sufficient to accommodate the strong changes of the input-
output characteristics of the muscle caused by fatigue so that an essentially constant performance of the movement act can be maintained (van Dieen 1996).

In fatiguing situations of the low back, the load on the joints and surrounding muscles may reach levels that exceed tissue tolerance. To avoid damage during repetitive lifting, the system must control the act so that the load on the musculoskeletal system is kept within bounds (Sihvonen 1997). With muscular fatigue, there is a reduction in motor performance and hampered motor control which will lead to increased loading on passive tissues in the lower back and cause mechanical damage. With repetitive isoinertial trunk flexion-extension movements Parniapour et al (1988) showed that fatiguing trunk muscles became weaker and slower while the neuromuscular system demonstrated less precision and control as the fatiguing activity continued.

2.8 Fatigue and Lifting

When lifting, linear and angular accelerations of body segments must be produced and reversed in time to end the movement while the inertial properties of the lifter-load system are changing as the object is picked up or released (van Dieen et al 1996). With manual material handling, the load on the joints and surrounding muscles, especially in the low back may reach levels that exceed tissue tolerance. To avoid damage, one must control the movement so that the load on the musculoskeletal system is kept within bounds. Muscular fatigue due to repetitive lifting has been shown to be a risk factor for the development of low back pain (Frymoyer et al 1983).

Muscle fatigue has been defined as any reversible decrease in performance capacity of a muscle that results from its activity (Bigland-Ritchie and Wood 1984). It is assumed that when
muscles fatigue, the motor control is not adjusted and changes in kinetics will result (van Dieen et al 1998). It has been shown that repetitive lifting leads to muscular fatigue and will decrease motor performance (van Dieen et al 1998). This will then increase loads on passive structures in the low back and mechanically damage the passive structures and lead to low back pain (van Dieen et al 1996).

The change in muscular coordination can be effected by fatigue and if the load being lifted is being held continuously. van Dieen et al (1996 b) has shown that back muscle fatigue in repetitive discrete lifting would cause a change in the coordination of the legs and trunk, more specifically in the phase lag between hip and spinal extension. Lifting and lowering with a load continuously in the hands revealed no such change. This was attributed to the continuous availability of feedback on the balance between the capacity of the muscles involved and the load mass. In discrete lifting this balance has to be estimated on the basis of memories from previous lifting movements. During repetitive lifting, the decrease in trunk extension velocity and the resulting increase in phase lag between hip and trunk extension is interpreted as a consequence of fatigue, more specifically of a decrease rate of force development of the back muscles (van Dieen et al 1996 b).
3.1 Subjects

Ten right handed males were recruited from the university population. All subjects were in good physical condition with no previous low-back injury. The mean age, height and body mass of the 10 subjects were 25.3 yrs, 179.2 cm and 80.4 kg, respectively, Table 1. The first subject repeated the testing procedure a second time after all subjects were completed. Informed consent were also obtained from all subjects attesting to their full understanding of the lifting and lowering requirement (Appendix 1). The research in this paper was approved by the Graduate Committee of the School of Human Kinetics.

Table 1: The Subject’s Anthropometric chart for age, weight and height.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Weight</th>
<th>Height</th>
</tr>
</thead>
<tbody>
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<td>82</td>
<td>184</td>
</tr>
<tr>
<td>2</td>
<td>25</td>
<td>76</td>
<td>173</td>
</tr>
<tr>
<td>3</td>
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<td>85</td>
<td>190</td>
</tr>
<tr>
<td>4</td>
<td>27</td>
<td>84</td>
<td>180</td>
</tr>
<tr>
<td>5</td>
<td>25</td>
<td>82</td>
<td>177</td>
</tr>
<tr>
<td>6</td>
<td>25</td>
<td>79</td>
<td>175</td>
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<td>7</td>
<td>26</td>
<td>78</td>
<td>180</td>
</tr>
<tr>
<td>8</td>
<td>25</td>
<td>77</td>
<td>167</td>
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<td>9</td>
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<tr>
<td>10</td>
<td>26</td>
<td>81</td>
<td>179</td>
</tr>
</tbody>
</table>
3.2 Data Acquisition

The asymmetrical lifting and lowering task was designed to fatigue the subjects with repetitive motion. The collection of the data was used to quantify the effects of fatigue due to loading of tissues of the lumbar spine by asymmetrical lifting and lowering of a load. It was expected that this research would give insight into the mechanisms of low back injury due to asymmetrical lifting and lowering. Each subject’s first session was the orientation session which provided the subjects with experience with the demands of the task. The next session was the collection session which required each subject to perform Maximum Voluntary Contraction (MVC) trials followed by a lifting and lowering session. The collection session was conducted 2-3 days after the orientation session.

3.2.1 Session Tasks

3.2.1.1 Orientation Session

No EMG or kinematic data were collected during the orientation session. First, the subjects were asked to perform isometric maximum voluntary contractions (MVC) and this will be described later in the collection session under “MVC trials”. The subjects were then asked to perform the asymmetrical lifting and lowering task, (described later), for 20 minutes to familiarize themselves with the task. The average 11 kg load in the box was used to allow the subjects to experience the loads that would be used during the collection session.

3.2.1.2 Repetitive Lifting Sessions

3.2.1.2.1 Lifting and Lowering Load and Frequency Characteristics

The selected load magnitudes for the dynamic lifting and lowering session were based on the Mital et al (1989) guide to manual materials handling. This guide’s design approach and
criteria provides different safe load recommendations for manual lifting and can be applicable to a wide variety of situations. Mital et al (1989) used four design approaches to develop their database. For the epidemiological criteria, the JSI value of 1.5 was used, with the maximum load that could be handled for this JSI value being 27 kg. The biomechanical criterion was the lumbar spine compression, which provided a safety margin of at least 30% of the lower back. The value for the males was 3930N with a corresponding load of 27 kg and for females the value was 2689N at a corresponding load of approximately 20kg. Physiologically the criteria were an energy expenditure rate of 4kcal/min for males and 3kcal/min for females for a 8h working duration. This represented approximately 29% and 28% of physical work capacity of males and females, respectively.

From the work of Mital et al (1989) a load of 11 kg was determined for the 50th percentile. The load was then increased or decreased around this average load to match each subject’s maximum isometric trunk extensor strength. To determine each subject’s maximum isometric trunk extensor strength, the subject was placed in a test contraction apparatus (TCA) as designed by Potvin et al (1996) (Fig. 4). The TCA allowed the subject to be in a controlled upright standing posture and was adjustable to the size of each individual. This apparatus served to stabilize the pelvis as subjects isometrically activated the spine extensor muscles against the resistance of a chain running horizontally from the wall to a bracket holding the trunk in place. A linear variable differential transformer (LVDT) force transducer was placed in series with the chain so that the horizontal force could be recorded and viewed in a LabView program to provide
Figure 4: The TCA as designed by Potvin et al (1996) allowed subjects to perform a maximum isometric trunk extension with the use of an oscilloscope, force transducer and harness.

force feedback. This force was multiplied by its vertical moment arm to the L4-L5 joint to calculate the trunk extensor moment. The length of the chain could also be adjusted so that the subject could stand upright with almost no detectable force or extensor muscle EMG being recorded. The subjects activated the trunk extensors with maximum effort and then relaxed. This effort was repeated three times consecutively. The subjects load used during the dynamic lifting and lowering session is shown in Table 2.
Table 2: Loads used for each subject during the dynamic session

<table>
<thead>
<tr>
<th>Subject</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>Load (kg)</td>
<td>10.6</td>
<td>8.5</td>
<td>13</td>
<td>12.5</td>
<td>10.6</td>
<td>11.9</td>
<td>11.6</td>
<td>11.6</td>
<td>11.5</td>
<td>10.0</td>
</tr>
</tbody>
</table>

3.2.1.2.2 MVC Trials

A method of training the subjects to produce the largest amplitudes of myoelectric activity was required to provide a basis for normalization. McGill (1991) used a basic isometric restraint strategy to allow the subjects to try to produce maximum abdominal muscle activity. This strategy was used to get the MVC of the abdominals from the subjects. In this strategy, the subjects were in a bent-knee sit-up posture with the feet restrained in an attempt to recruit the abdominals. The subjects hands were placed behind the head and the trunk forming an angle with the horizontal of approximately 30 degrees. Resistance to the shoulders occurred during the maximum sit-up effort. The subject was then asked to do a maximum sit up effort in the sagittal plane followed by a maximum twist effort to the left then the right. To obtain maximum efforts of the lumbar and thoracic erector spinae, the TCA was used as described earlier to obtain subject's maximum isometric trunk extensor strength.

3.2.1.2.3 Dynamic Lifting and Lowering

Subjects were asked to wait for an audible tone then lift a load on their right side from knee height position and place it on a shelf on their left side, pause and wait for an audible tone, then lower the load back to the knee height position on their right side. The collection duration was 40 minutes. Subjects were provided with an audible tone every 7.5 seconds to help them pace the lifting and lowering sequence at a desired frequency of 4 full cycles per minute. The
subjects were asked to release the box's handle after every lift and lower. These two requirements were the only instructions regarding how the task was to be performed. After the subject performed the dynamic session the subject was then placed into the TCA to perform and record a maximum extensor moment.

3.2.2 Methods of Data Collection

3.2.2.1 Lifting and Lowering Equipment

The lifting tray used was 42 cm wide and 15 cm tall. The two handles were 7.5 cm from the bottom of the lifting tray and 4 cm from the top of the lifting tray. The lifting tray started at the "low position" with the handles being knee height on the subject. The subjects lifted the tray to a "high position" height at the subjects shoulder region. As seen in Figure 5, the subjects were placed between the low and high position where the low position is to the right of the subject and the high position is to the left of the subject. The high and low position were 120cm apart laterally from each other.

![Diagram showing high and low positions]

Figure 5: Subject position in dynamic testing procedure
3.2.2.2 EMG Preparation

Myoelectric signals were recorded by means of 8 pairs of bipolar AgAgCl surface electromyogram (EMG) electrodes attached to the skin with a separation of 5 mm. First the skin was prepared by shaving hair. The skin was then slightly abraded and swabbed with alcohol. The electrode placement followed the procedures of Pope et al. (1986) and McGill (1991). A ground electrode was placed around the right hip area. All electrode placement occurred for both the left and right side of the body. The sites were located as follows: **The internal oblique (IO)** - The electrodes were 4 cm from the midline at a vertical height 8 cm inferior to the umbilicus. **The external oblique (EO)** - The electrodes were placed on the sides at the level of the umbilicus but 6 cm dorsal to the ASIS. **Lumbar erector spinae**- (representing the iliocostalis lumborum (IL) and longissimus thoracis (LT) pars lumborum). Electrodes were placed 3 cm lateral to the posterior spinous process at the level of L3. **Thoracic erector spinae**- (representing the IL and LT pars thoracis). Electrodes were placed 4 cm lateral to the posterior spinous process at the level of T9. These electrode locations and arrangements have been shown to best represent the different muscle activity patterns of the trunk and minimize signal cross-talk between electrode pairs during bending and twisting tasks (Lafortune, Norman, McGill 1988).

3.2.2.3 Spine Kinematics

The magnetic Insidetrak rotation monitoring device (Polhemus Navigation Sciences Division, McDonnell Douglas Electronics Company) was used during the dynamic lifting and lowering tasks. The Insidetrak is a electromagnetic device with a sensor that monitors the high
frequency oscillating magnetic field emitted from the source. Rotational information can be obtained based on the three dimensional orientation of the sensor, with respect to the source.

In this study, one sensor was mounted on a plastic form and taped on the subject's back at the level of L5. The other sensor was also mounted on a plastic form and taped to the subject at the level of T6. This system monitored the rotations around three anatomical axes of the rib cage relative to the pelvis segment. This information was used to distinguish the flexion/extension, of the spine during the dynamic task.

The kinematic data from the one axis of rotation was saved digitally at a sample rate of 7.6Hz. The LabView software package was used to record the data from the EMG and another 386 IBM computer was used to record the data from the Insidetrak.

3.3 Data Analysis

The EMG and spine kinematic data were used to identify whether there were any technique changes in the asymmetrical lifting and lowering due to fatigue. The EMG data were used to estimate the muscle force-time histories and quantify muscle activation due to fatigue. The kinematic data were used to determine the rotation about the one axis (relative to the subject's range of motion). The MVC trials were used to normalize the EMG during dynamic lifting and lowering.

3.3.1 Lumbar Spine Kinematics

Data from the Insidetrak system were used to determine the bias of the neutral standing posture observed for the flexion-extension. Flexion rotation represented the rotation in the sagittal plane during the lifts. The peak angles for each lift were determined for flexion/extension axis. The average of the peaks for the first and last 12.5% of the dynamic session were calculated.
3.3.2 Normalization of EMG

The normalization of EMG was based on Potvin, Norman and McGill (1996). For each isometric MVC condition, there were three repetitions. The highest one of the three represented the maximal activation level of EMG. The dynamic EMG signals from the four erector spinae channels and the four abdominal channels were initially divided by their individual maximum values. For the four trunk and four abdominal EMG channels, a 6th low pass filtered digitally (second order Butterworth) with a cutoff of 2.7 Hz (Potvin et al 1996). This produced a linearly normalized signal which more closely approximated the muscle force patterns.

The linearly normalized EMG levels represented the relative effort of the abdominal and trunk muscles. For each individual lift (windowed from start to end), the mean and peak activation levels were calculated for the right and left channels for each muscle group. The pooled average of means and peaks for the first and last 12.5% of the dynamic session were calculated for each subject. The fatigue related change in Mean Power Frequency was determined from a regression analysis for all lifts and lowers performed in the dynamic lifting session.

3.3.3 Determining the Start and End of Each Lift and Lower

It was necessary to establish the beginning and end of each lift and lower. This was done by the use of a triaxial accelerometer on the lifting tray and a computer program LABView. When the subjects lifted and placed the lifting tray in the designated area the triaxial accelerometer recorded the vertical movement of the box. The resultant force was used to trigger the start and end of the lift/lower movement. The averages for the peaks of the first and last 5 minutes of the dynamic session were then calculated.
3.3.4 Statistics

The design was a 2x2 factorial ANOVA with repeated measures. The independent variables manipulated were fatigue and movement direction. There were two levels of each variables: 1) movement direction (lifting and lowering) and 2) time in the session (rest and fatigue).

Peak and mean AEMG for the bilateral Internal Oblique, External Oblique, Lumbar Erector Spinae and Thoracic Erector Spinae, and peak flexion were used as dependent variables in a repeated measures ANOVA (Table 2). For each dependent variable measured two null hypothesis were set and they were evaluated by comparing computed F-ratios with critical values at p<.05. In each case the null hypotheses were:

Ho: DV (Fatigue) = DV (Non-Fatigue)

Ho: DV (Lifting) = DV (Lowering)

Table 3: Dependent variables used in the repeated measures ANOVA

<table>
<thead>
<tr>
<th>Dependent variables</th>
<th>Mean</th>
<th>Peak</th>
</tr>
</thead>
<tbody>
<tr>
<td>AEMG LEFT/RIGHT IO</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>AEMG LEFT/RIGHT EO</td>
<td>X</td>
<td>X</td>
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<tr>
<td>AEMG LEFT/RIGHT</td>
<td>X</td>
<td>X</td>
</tr>
<tr>
<td>LES</td>
<td></td>
<td></td>
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<tr>
<td>AEMG LEFT/RIGHT</td>
<td>X</td>
<td></td>
</tr>
<tr>
<td>TES</td>
<td></td>
<td></td>
</tr>
<tr>
<td>FLEXION</td>
<td></td>
<td>X</td>
</tr>
</tbody>
</table>
CHAPTER IV
RESULTS

The results have been divided into two sections 1) fatigue assessment and 2) dynamic session. These sections will present grouped data and then individual subject results will follow.

4.1 FATIGUE ASSESSMENT

In an experiment of this type it is necessary to show that the subjects became fatigued before conclusions are made about the effects of fatigue. One measure was used to quantify the presence of fatigue in each subject during the lifting sessions. It was expected that fatigue would be accompanied by a decrease in maximum extensor moment. The pooled results will be used to determine if fatigue was accompanied by reductions in maximum extensor moments.

4.2 Maximum Extensor Moment

Subjects performed maximum contractions in the TCA prior to, and after, the dynamic sessions. Each subject’s maximum extensor moments pre and post dynamic session are presented in Figure 6. A t-test indicated a significant reduction (p<.05) in the maximum extensor moment after the dynamic session from 251.35 Nm to 205.20 Nm. The average moment decreased 18.3% after the dynamic session. The task resulted in all subjects decreasing the maximum extensor moment capacity in spite of the difference in loads used by each subject.
Figure 6: The average maximum extensor moment Pre & Post lifting trials (n=10).
Figure 7: Subject 1's right lumbar EMG for an entire trial with windowing data indicating the start and end of each lower for the first 5 minutes.

4.3 Dynamic Lifting Data

4.3.1 EMG Amplitude

Figure 7 presents an example of the EMG signals from one muscle for an entire trial along with the windowing data indicating the start and end of each lower for the first 5 minutes. The average mean and mean peak levels for each channel were calculated from each lift and overall
averages were then calculated with the variables for the first and last 5 minutes of the lifts and lowers monitored in the dynamic session. The analysis of variance procedure indicated no significant differences or interactions in the EMG activity for the first 5 minutes when compared to the last 5 minutes peak means and average means for all muscles observed (Table 4). There were statistically significant differences in average mean and peak levels of various muscles among lifting and lowering. Subsequently then, all ANOVA results will refer to comparisons between the lifts and lowers.

Table 4: The difference of mean EMG activation from the first five minutes to the last five minutes for all subjects observed trunk muscles in both lifting and lowering.

<table>
<thead>
<tr>
<th>Lift</th>
<th>R-Lum</th>
<th>L-Lum</th>
<th>L-Thor</th>
<th>R-EOLEOR-IOL-IO</th>
<th>R-Lum</th>
<th>L-Lum</th>
<th>L-Thor</th>
<th>R-EOLEOR-IOL-IO</th>
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<tr>
<td>s1</td>
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<td>-3</td>
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<tr>
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</tr>
<tr>
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<td>5</td>
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<td>-1</td>
</tr>
<tr>
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<td>2</td>
<td>-2</td>
<td>-1</td>
</tr>
</tbody>
</table>

The pooled subjects results will be presented first and will then be followed by a more detailed presentation of case studies of individual subject responses.

4.3.1.1 Lumbar EMG

There were no statistically significant differences in the right lumbar mean or peak EMG activation. The difference in the mean values of the left lumbar when comparing lifting and lowering is statistically significant for both average mean (F=39.278, p<.05) and peak level (F=53.386, p<.05). The average mean was higher for lifting by 25%, while the peak levels for
lifting was 31% higher. (Figure 8a) The right lumbar levels tended to be lower than those of the left lumbar.

4.3.1.2. Thoracic EMG

The difference in the mean values of the left thoracic when comparing lifting and lowering is statistically significant for both average mean (F=37.7, p<.05) and peak level (F=11.5, p<.05). The average mean was higher for lifting by 27%, while the peak levels for lifting was 25% higher. (Figure 8b)

4.3.1.3. External Oblique EMG

External Oblique EMG activity was low for all subjects as the peaks rarely exceeded 10% of maximum. There was no change in average mean activation level with lifting and lowering for the right external oblique but the peak levels were significantly different (F=14.834, p<.05). The peak level for the right external oblique was higher in lowering by 26% MVC. The left external oblique peak means were not significantly different but the average means for lifts were significantly higher by 13% (F=6.958, p<.05). (Figure 8c)

4.3.1.4. Internal Oblique EMG

Internal Oblique EMG activity was low for all subjects as the peaks rarely exceeded 15% of maximum. The average mean and peak levels for the right internal oblique during lifting and lowering were higher than lowering by 27.58% (F=11.7, p<.05) and 21.75% (F=8.9, p<.05) respectively. The average mean for the left internal oblique during lifting and lowering were higher than lowering by 19.41% (F=22.3, p<.05). (Figure 8d)
Figure 8a: The left and right lumbar peak and mean EMG amplitude (across subjects) during lifting and lowering (n=10). The bars are standard deviation.

Figure 8b: The left thoracic peak and mean EMG amplitude (across subjects) during lifting and lowering (n=10). The bars are standard deviation. The right thoracic EMG is missing due to signal errors.
Figure 8c The left and right external oblique peak and mean EMG amplitude (across subjects) during lifting and lowering (n=10). The bars are standard deviation.

Figure 8d The left and right internal oblique peak and mean EMG amplitude (across subjects) during lifting and lowering (n=10). The bars are standard deviation.
Figure 9: Comparison of the first 5 minutes and the last 5 minutes mean relative peak flexion angle (across subjects) during lifting and lowering. (N=9) The bars are standard deviation.

4.3.2. Kinematic Data

4.3.2.1. Trunk Flexion

Flexion rotation was observed during the dynamic lifting movements. The average results for the peak relative levels of rotations observed in the flexion axis are presented in Figure 9. In the dynamic session there was a significant difference in the pooled lift and lower peak levels of spine flexion (p<.05). The peak levels for spine flexion during the first 5 minutes when compared to the last 5 minutes were higher by 9%. There was no statistically significant difference in relative peak levels of spine flexion during lifting and lowering.
4.4 Individual Subject Data

Subjects 7 and 9 were selected to represent the range of the observed responses to fatigue and differences between lifting and lowering. Subject 7 showed changes that were hypothesized to put him at a higher risk of injuring his back if he had continued the dynamic session for much longer durations. Conversely, subject 9 showed very little change in the lifting mechanics in response to repetitive lifting.

4.4.1. Individual Subject Responses to Fatigue (Subject 7 and 9)

All changes discussed are those that were significant (p<.05) and assumed to be substantial changes. For these descriptions the definitions of squat lifts and stoop lifts will be used as previously discussed in section 2.2.

4.4.1.1. Subject 7

Lifts were first done with a squat technique. The subject would step close to the load to pick it up from either the high or low position and set down the load. As the session progressed, the subject began to flex more in the spine. During the final 5 minutes of the dynamic session, a full stoop technique was used and he began to decrease the movement of the feet throughout each lift and lower. These changes were accompanied by increases in the left lumbar activation of 12% MVC in lifting and 6% MVC in lowering. There was also a decrease in the left thoracic recruitment during lifting by 9% MVC. As well, there was a substantial increase in relative flexion of 44% for lifting and 14% for lifting.

4.4.1.2. Subject 9

A squat technique was maintained throughout the session. As the session progressed, there was minimal change in the spinal flexion of the subject. Changes were only seen in the abdominal
muscles and left lumbar activation. There was a substantial decrease in the left lumbar activation of 7% MVC. Both the left and right internal obliques decreased activation by 31% and 21% respectively in lifting and 19% and 13% respectively in lowering.

4.5 Repeatability
To show reliability of the testing procedure, subject 1 repeated the dynamic session after all subjects completed the testing. As seen in Table 5, the comparison of subject 1's results from the two dynamic sessions show that there was little variation within the subjects results for the dynamic session. The results in Table 5 supports an assumption that this research protocol was reliable.

Table 5: % Change Comparing the First 5 Minutes to the Last 5 Minutes during lifting for Subject 1's two trials.

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<th>Subject 1 Repeat</th>
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CHAPTER V
Discussion

5.1 Main Findings

The current study was designed to determine the response of load handlers to fatigue during repetitive, dynamic asymmetrical lifting and lowering. Few such studies have been conducted previously. One of the findings of this study was that the asymmetrical lifting and lowering task decreased the trunk's maximum extensor moment. The finding was based on the expectation that the trunk muscles would fatigue and result in a decrease in maximum extensor moment. To a certain degree, all subjects' maximum extensor moments decreased when comparing the pre and post trials and the group average was significant. The protocol was sufficient to cause fatigue but the kinematic and EMG data during the dynamic session do not support a consistent effect of fatigue on all subject. There was an average decrease in flexion, when comparing rest and fatigue, but not all subjects increased flexion to the same degree. The muscle recruitment pattern for the trunk muscles also varied for each subject. Lifting was observed to result in higher EMG amplitudes than lowering but it is not likely that there were substantial differences in the erector spinae muscle forces given the force-velocity relationship of muscle.

5.2 Hypotheses

5.2.1. Repeated Lifting and Lowering and Fatigue

It was hypothesized that repeated asymmetrical lifting and lowering would result in fatigue of the muscles of the trunk. This fatigue has been hypothesized to have a number of potential
links to low back injury (Potvin, 1991). It has been hypothesized that fatigue, due to low back loading during repetitive lifting and lowering over a prolonged period of time, would result in changes of lifting mechanics that could increase the risk of low back injury. The potential consequences of this repetitive/prolonged loading could result in central fatigue and if one or more muscles groups is repeatedly loaded it would result in peripheral fatigue With the repeated loading on certain muscle groups there would be a decrease in strength and endurance of the muscle group. The combined effect of central and peripheral fatigue may lead to reduced motor coordination and result in a switch to loading of the passive tissues or other muscle groups. This increase in loading, on the passive tissues of the low back, can mechanically damage the passive structures and lead to low back pain (van Dieen 1996). The reduced motor coordination may also result in the shift to a more mechanically unsafe lifting technique that can result in acute tissue damage or chronic injuries after repeated loading.

The maximum extensor moment in the TCA was used as the direct measure of this capacity. In previous studies, it has been shown that repetitive lifting leads to muscular fatigue (Potvin and Norman 1993, van Dieen et al 1998). Potvin and Norman (1993) have shown that repetitive symmetrical lifting and lowering with similar loads and durations resulted in reduction in maximum extensor moment capacity in spite of differences. The current study resulted in similar findings when comparing pre and post maximum extensor moments. The trunk extensor moment decreased by 18.3% after the asymmetrical lifting and lowering task. This was a statistically significant difference at p<.05 and led to the rejection of a null hypothesis of no differences. Thus the alternate hypothesis was accepted.
5.2.2. Changes in Lifting Mechanics

It was hypothesized that repeated lifting and lowering would precipitate numerous changes in lifting mechanics that can be hypothesized to increase the risk of low back injury. This was quantified as an increase in flexion when comparing the first 5 minutes to the last 5 minutes of the dynamic session. Parnianpour et al (1988) found a decrease in flexion as subjects performed repetitive flexion-extension contractions against resistance in a triaxial dynamometer. The result from this present study, as well as Potvin and Norman (1993) are in direct contrast to those of Parnianpour. For all subjects, there was a increase in flexion when comparing rested to fatigued lifting and lowering. The difference can likely be due to the mode of contraction used in their study. Subjects were forced to contract concentrically for both flexion and extension. Subjects may have flexed less with abdominal fatigue due to the work required for each cycle. In this present study and Potvin and Norman (1993), the extensor muscles were required to contract concentrically in extension and eccentrically during flexion. As the load was lowered, the extensor muscles were required to contract concentrically in extension and eccentrically during flexion. This external flexor moment required a greater percentage of the subject’s current capacity as these muscles became fatigued (Potvin 1991). The subjects could have submitted more to the increase in the perceived flexor demands and allowed the spine to flex more than during the first 5 minutes of the dynamic session.

The EMG mean and peak levels were not significantly different when comparing the rested and fatigued lifting and lowering. This is consistent with the findings of Potvin (1991). This is due to the varied response to fatigue exhibited by each subject. Subject 7 increased extensor muscle activation and increased flexion as the subject became fatigued, whereas subject 9 showed
no significant change in flexion but increased abdominal activation with fatigued. Therefore the results of this study do not support a consistent effect of fatigue for all subjects. There were subjects, such as subject 7, who had significant changes in lifting mechanics due to fatigue, but others, such as subject 9, who showed no change in lifting mechanics. These different responses to fatigue will be discussed later in the chapter. There was no statistically significant difference at p<.05 and led to acceptance of the null hypothesis.

5.2.3. Lifting vs Lowering

It was hypothesized that the effects of repeated load handling would be more pronounced during lowering. Few studies have compared lumbar spine kinematics between lifting and lowering. DeLooze et al (1993) investigated lifting and lowering differences in the L5/S1 joint moments as a result of peak flexion angle and acceleration of the body’s center of mass. They observed that there were no significant differences in peak L5/S1 flexion angle. The research of McKeen and Potvin (1999) support this finding, as they found no significant difference in maximum flexion when comparing lifting and lowering. Results from this current study support the previous findings as peak lumbar spine flexion angle was only 2.1% greater during lifting compared to lowering.

The EMG mean and peak levels for TES and LES were significantly higher during lifting when compared to lowering of the load. This was consistent with previous studies of lifting and lowering (McKeen & Potvin 1999; Potvin et al 1996; de Looze et al 1993 Dolan & Adams 1993). This difference is attributed to the force-velocity relationship of the muscle. During concentric contractions of the erector spinae during lifting, the capacity to generate force is decreased relative to an isometric contraction. Whereas, during an eccentric contraction (during lowering)
there is enhancement in muscles ability to generate force (Hill 1938). The difference in the EMG mean and peak levels for lifting and lowering were not affected by the fatigue. There was no statistically significant difference at p<.05 and led to acceptance of the null hypothesis but the difference in EMG activity in lowering may increase injury.

Potvin and Norman (1993) have shown that muscle activation during lowering may increase the chance of an individual injuring themselves during such a task. Lifting and lowering have similar body positions, muscle coordination and net joint moments, but in lowering these forces are distributed over a smaller functional cross-sectional area of active muscle, which may increase the chance of injury (de Looze et al 1993). As well, Davis et al (1998) have shown that lowering tasks approach compression spinal tolerance limits and moments of 140.5 Nm, whereas the lifting exertions were well below compression tolerances and moments of 113.1 Nm. Thus, the lowering conditions were found to possibly increase the risk of a lower back injury.

5.3. Limitations

There are several limitations that should be addressed with regards to this study. First, the conclusions resulting from this current study must be made with the understanding that low back research lies in its application to the environment in which injuries are occurring. This study was performed in a laboratory setting with subjects who are not experienced in manual materials handling. Attempts to study and understand low back injuries are vital, given that there is both human pain and monetary losses associated with these injuries. It is not possible to design research with the subjects actually injuring themselves to allow for direct measures of this phenomenon. Only the initial stages of fatigue were assessed in this research with the subjects fully understanding that they could terminate the task whenever they felt their safety was in
jeopardy. The relevance of these findings to the effects of repetitive tissue loading over days, weeks, or months and the expected progressive changes in the lifting mechanics based on these early stages of a normal task can be only estimated.

The dynamic EMG was not corrected for changes in length and velocity. Thus, the interpretations are based on activation and not force. These corrections were not done due to the fact that the validity of the Insidetrak lateral bend and axial twist data were likely contaminated to some degree. After the collection of all subjects it was then realized that the data for the secondary planes did not represent the functional movement of the subjects. The secondary kinematic data was then discarded. The estimated trunk flexion angle was calculated to find the relative angle of one sensor to the other. As well, the use of external measures of trunk motion and the assumptions are limiting factors. Video analysis would not have provided an adequate sampling frequency unless high speed video was used. This technology was not available.

The missing right thoracic data has limited the study due to that fact that we are unable to assume influences of change from thoracic to lumbar EMG activation. A decrease in lumbar EMG can indicate lower posterior shear contributions from the lumbar erector spinae to compensate for anterior shear forces from the reaction forces and possibly the posterior ligaments when they were recruited (Potvin, 1991). An increase in compression loading of the spine can be identified with an increase in lumbar and/or thoracic EMG which would influence tissue fatigue failure of the vertebral bodies and intervertebral discs and may have implications for erector spinae muscle damage.
5.4. Additional Findings (Case Studies)

As stated earlier, there was a varying degree of how the subjects responded to repetitive lifting and lowering. Subject 7 demonstrated increases in the degree to which the spine was flexed as fatigue progressed during the lifting session. By the last five minutes of the dynamic session the peaks in flexion were significantly higher in both lifting and lowering. The increased flexion was accompanied by decreases in the left thoracic EMG activation. The data of this subject indicated an increased demand on the passive tissue, an increase in flexion and evidence of flexion-relaxation response of the spine. This was first identified by Floyd and Silver (1951) and has been shown in rare cases for rested symmetrical lifting task characteristics (Potvin et al 1991). The increase in flexion-relaxation response resulted in a increase in passive tissue demands in the thoracic region. Roy, DeLuca and Cassavant (1989) stated that this can occur in the presence of functional muscle insufficiency. Based on this previous research, there can be mechanical implications for the injury of ligaments, intervertebral discs and even extensor muscles. Flexion and lifting have been linked to the incidence of spine injury and pain in a number of studies (Chaffin and Park 1973, Frymoyer et al 1983). A shift to passive tissues, such as the ligaments, means that these tissues can be repetitively stressed and these structures are not generally recruited under rested conditions (Potvin, McGill and Norman 1991). Due to fatigue “wear and tear” with repeated loading applications it is hypothesized there will be a reduction in their tolerance to loading (Kumar 1990). A decrease in thoracic EMG associated with the flexion-relaxation response would have resulted in a net increase in the shear forces acting on the disc from the reaction forces and interspinaous ligaments (Potvin, 1991).
Injuries due to fatigue are most often found in the lower levels of the lumbar spine where changes in mobility of the lumbar joints can be most pronounced (Hsu et al 1990). In addition, Koeller et al (1984) have indicated that the disc height may be reduced by repeated loading. The increase in the left lumbar activity seen in subject 7 would have increased the compression loading of the lumbar spine. For subject 5 there was a substantial decrease in the right lumbar activity for both lifting and lowering. As well, subject 5 substantially increased flexion in both lifting and lowering. With extreme flexion in the lumbar spine it has been found that there is an increase of loading on the facets and annulus (Shirazi-Adl, 1991 and 1989). Prolonged stress can weaken the annulus until loads, which were previously tolerable, can cause complete fissure or prolapse.

Subject 9 demonstrated no change in the degree to which the spine was flexed as fatigue progressed during the lifting session. The kinematic data from subject 9 is similar to van Dieen et al (1996) results where repetitive lifting caused no change in the degree of flexion. Even though there was no change in the kinematic data in subject 9, there was significant increase in EMG activity of the abdominals as the lifting session progressed. Preliminary evidence does show that fatigued muscles tend to increase cocontraction of antagonists (van Dieen 1996). An increased coactivation of abdominal muscles during lifting would increase compressive loads on the spine. The development of fatigue in a prolonged repetitive task such as the one in this study may lead to only minor changes in kinematic movement, but it may greatly reduce the ability of the subject to react to perturbations. An increase in reflex motor time has been reported to accompany muscle fatigue (Hakkinen & Komi 1983), as well, a rise in time of voluntary trunk extension has been shown to increase with fatigue (Parnianpour et al 1988). When the magnitude of the subjects response time is greatly reduced, there is the possibility that there may be excessive loading on
structures of the low back that are not normally stressed to that degree. Previous research has suggested that the pre-activation during trunk loading and trunk muscle cocontraction may play an important role in enhancing spine stability (Chaing 1998, Krajcarski et al, 1999). These reductions in one's ability to respond adequately in a timely manner may increase the chance of a low back injury.

In summary, subject 7 provided evidence to support the hypothesis that fatigue could result in changes in lifting mechanics which may place individuals at a higher risk of sustaining a low back injury. The data indicated an increased demand on passive tissues as the activation of the extensor muscles change from the lumbar to thoracic and vis versa. These changes were not observed for all subjects. Even though the kinematics of subject 9 did not change with fatigue, the possible changes in one's ability to react to perturbations may also lead to low back injury.
CHAPTER VI
Conclusion

6.1 Summary

Low back disability due to manual material handling continues to be the leading cause of
lost productivity for humans in the work force. While both asymmetrical lifting and repetitive
lifting have been linked to an increase in the risk of low back injury, few studies have been
performed to biomechanically assess if there are changes in spine mechanics with asymmetrical
lifting over prolonged periods of time. Thus, the purpose of the current study was to monitor
trunk muscle electromyography (EMG) and spine kinematics during repetitive asymmetrical lifting
and lowering involving spine flexion, lateral bend and axial twist and determine if there is a change
in the risk of low back injury associated with fatigue. Ten male subjects were studied. Each was
instrumented with a kinematic devise at the sacrum, as well as, surface EMG electrode pairs to
monitor the activity of the bilateral lumbar erector spinae (LES), thoracic erector spinae (TES),
internal oblique (IO) and external oblique (EO) muscles. EMG signals were normalized to
maximum voluntary contractions (MVCs) performed at the beginning of the session. Subjects
were asked to perform lifts, with a load of approximately 11 kg, from a platform at knee height to
a platform at shoulder height. Lowers were performed in the opposite direction. The platforms
were spaced 120 cm apart and 4 lifts and 4 lowers were performed each minute for a total of 40
minutes. EMG data were sampled at 1000Hz, rectified and low pass filtered to approximate the
relative muscle force. Data were pooled across the first and last 10 lifts (representing the
"rested"and "fatigued" states) and for the first and last 10 lowers. Data were analyzed to
determine the peak EMG for each channel. The average of the ten repetitions were averaged for
each variable and each condition. A two way ANOVA with repeated measures was used to
determine the main effects of time (rest and fatigue) and direction (lift and lower) and their interaction. Significance was set at p<0.05. One of the findings of this study was that the asymmetrical lifting and lowering task decreased the trunk's maximum extensor moment. The finding was based on the expectation that the trunk muscles would fatigue and result in a decrease in maximum extensor moment. To a certain degree, all subjects' maximum extensor moment decreased when comparing the pre and post trials and the group mean was significant. The protocol was sufficient to cause fatigue but the kinematic and EMG data during the dynamic session results do not support a consistent effect of fatigue on each subject. This was shown with an average decrease in flexion, when comparing rested and fatigue, but not all subjects decreased flexion to the same degree. As well, the muscle recruitment pattern for the trunk muscles also varied for each subject. Lifting was observed to result in higher EMG amplitudes than lowering but it is not likely that there were substantial differences in the erector spinae muscle forces given the force-velocity relationship of muscle.

Based on the results of this study and with the limitations of the study in mind, the following conclusion is warranted:

1: Even though there was no consistent changes across all subjects, the data in this research indicates that fatigue can compromise safety during an asymmetrical lifting task.

6.2. Hypothesis

1) It was hypothesized that repeated asymmetrical lifting and lowering would result in fatigue of the muscles of the trunk.

This hypothesis was accepted.
2) It was hypothesized that repeated lifting and lowering would precipitate numerous changes in lifting mechanics that can be hypothesized to increase the risk of low back injury. This hypothesis was not accepted.

3) It was hypothesized that the effects of repeated load handling would be more pronounced during lowering. This hypothesis was not accepted.

6.3. Contributions to the Literature

To the author’s knowledge this was the first study to biomechanically determine the response of subjects to dynamic asymmetrical lifting and lowering over a period of time. Evidence was provided to support the hypothesis that fatigue could result in changes in lifting mechanics which may place individuals at a higher risk of sustaining a low back injury. Even though large changes were not observed for all subjects, the different response of each subject to repetitive lifting and lowering may have led them to possible injury.

This research has increased the understanding of spinal mechanics and knowledge of the mechanisms that may contribute to low back injury during asymmetrical lifting and lowering. The evidence presented in this study can be used to alert the industry of the need to design occupational tasks so that there is minimal asymmetrical movement and control of fatigue. Even still, there is the need for more research in asymmetrical movement and fatigue to determine the missing information not obtained in this current study.
6.4 Future Directions

Future retesting of individuals should be done to obtain the missing kinematic data in this present study. The comparison of Panianpour et al (1988) finding of increased secondary plane movement with fatigue and this present format could possibly bring information not yet available. As well, there is more research needed in the area of asymmetrical movement and fatigue.
REFERENCE


APPENDICES
Appendix 1. Consent Form
Information and Consent Form

Ergonomics and Biomechanics Lab
School of Human Kinetics
University of Windsor

Study Title: Fatigue Response to Dynamic Asymmetrical Lifting and Lowering

Conducted by: Derek F. Fraser (Master’s student) and Dr. J. R. Potvin (supervisor)

I agree to participate in a study which is designed to add new knowledge concerning the mechanisms of musculoskeletal injury during prolonged asymmetrical lifting and lowering. The investigator has explained the procedures and the necessary time commitment to me. I understand that I may be asked to perform repeated asymmetric lifts and lowers for a session of 40 minutes or until exhaustion. The load for the session will not exceed 9 kg. The selected load magnitude for the dynamic lifting and lowering session is based on Mital et al (1989) Guide to Manual Materials Handling. I may be asked to perform maximal isometric exertions of the back extensor and abdominal muscles. These maximal isometric exertions are strength tests which have commonly been used in isometric strength evaluations. The dynamic lifting and lowering session may possibly result in muscle stiffness the following day but this will be no more than may be experienced after any unaccustomed physical exertion. The procedure will involve attachment of electrode to the surface of various muscles for monitoring their electrical activities as well as the attachment of magnetic movement tracking devices to the ribcage and pelvis. I understand that with each lift there is always a risk that I may experience discomfort or injury. I have been instructed to terminate lifting at any time I feel an injury may occur. I am aware that electrical shock is a possibility from EMG equipment or other electronic devices, to which I am connected, and that skin rash from the electrode adhesive sometimes occurs. These problems, however, are rare. This project has received ethical clearance from the Graduate Committee of the School of Human Kinetics at the University of Windsor.

Consent of Subject

I have read and understood the information presented above for the procedures and risks involved in this study and have received satisfactory answers to questions related to this study. The specific details of this study have been explained. I understand that my identity will be protected throughout my participation in this study. I am aware that I may report what I consider to be violations of my welfare to Dr. Bob Boucher 253-4232 ext. 2429 or the Office of Human Research, University of Windsor, and may withdraw from the study at any time. With full knowledge of all foregoing, I agree, of my own free will, to participate as a subject in this study and to allow photographs and/or other data to be used for teaching or research presentations.
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<th><strong>NAME:</strong></th>
<th>Derek Fredrick Fraser</th>
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