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Bienvenido Pablo. Parcero

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The in vivo dynamic fatigue response of the spine to sudden loading in the sagittal plane: effects of pre-load and step input magnitude

By

Bienvenido P. Parcero

A Thesis
Submitted to the Faculty of Graduate Studies and Research
Through Human Kinetics
In Partial Fulfillment of the Requirements of
The Degree of Master of Human Kinetics at the
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Windsor, Ontario, Canada

2000

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Abstract

THE IN VIVO DYNAMIC FATIGUE RESPONSE OF THE SPINE TO SUDDEN LOADING IN THE SAGITTAL PLANE: EFFECTS OF PRE-LOAD AND STEP INPUT Magnitude

This purpose of the study was to investigate the effects of fatigue on the response of the trunk to a variety of conditions of initial static loads and step inputted loads causing rapid sagittal trunk flexion. The variables measured in the study were the pre-trial and peak EMG levels from 5 bilateral trunk muscles, peak extensor moments, peak trunk flexion angle, trial duration, time to peak moment and times to peak EMG levels.

The findings provide evidence that there were some changes in the response of the spine to sudden loading during fatigue conditions. However, the only main effect of fatigue was for the TES response to the different added loads. There was also a significant interaction between fatigue level and pre-load for the maximum trunk flexion variable, with the fatigued 4% pre-load trials having greater values than the rested 4% pre-load trials. The expected increase in LES and trunk flexor (co-contraction) EMG activity during fatigue was not observed.

A review of the individual subject data revealed that two subjects did demonstrate increases in agonist and antagonist muscle activity for the fatigued loading conditions, when compared to the rested conditions, and one subject was observed to have increased peak flexion values as a result of fatigue. Overall, the data seem to indicate that fatigue affects individuals in different ways, with respect to spine mechanics. It may be hypothesized that some subjects were more susceptible to the effects of fatigue. This may help explain why some, but not all, individuals are injured while manual material handling over prolonged periods.
Acknowledgements

I would like to thank all of the subjects who participated in this study. Jeff Jones deserves my thanks for his assistance during my data collection. I want to thank the members of my thesis committee for the work and time they contributed to my thesis, and they are Dr. Dave Andrews, Dr. Peter Frise, and Dr. Jim Potvin. I would especially like to thank Dr. Jim Potvin for his guidance and patience throughout the duration of my University of Windsor experience. A large thank you goes out to my parents, Pablo and Marilou Parcero, for their continued support and positive encouragement. To my sister, Mahalia, I would like to thank you for being a great sister. To my family and friends (especially Stephanie Jakubo), thank you for your love and support.

Lastly, I would like to dedicate this thesis to my late grandfather, Colonel Bienvenido Parcero. I can only hope that you are as proud of me as your grandchild; as I am proud of having you as a grandfather and the opportunity to carry your name.
# Table of Contents

Abstract iii

Acknowledgements iv

List of Figures vii

List of Tables x

Introduction 1

Statement of Purpose 5

Statement of Hypotheses 6

Review of Literature 9

The Lumbar Vertebrae 9

The Mechanism of Forward Flexion of the Lumbar Spine 11

Normal Ranges of Motion 12

Muscles of the Trunk 13

The Abdominal Musculature 13

The Lumbar Back Musculature 14

The Electromyography – Force – Fatigue Relationships 16

Stability of the Spine 19

Muscle Force and Muscle Stiffness 21

The Role of the Trunk Musculature in Spine Stability 21

The Role of Intra-Abdominal Pressure in Spine Stability 22

The Effect of Fatigue on Spine Stability 23

Sudden Loading and the Spine 24

Methods and Procedures 26

Subjects 26

Experimental Apparatus 26

Data Acquisition 29

Orientation Session 30

Collection Session 31

Data Treatment 35

Trial Selection and Windowing 35

EMG, Force & Displacement Data Treatment 36

Dependant Variables 37

Data Analyses 38

Results 40

Grouped Subject Data 42

Fatigue Assessment 42

Mechanical Variables from the Sudden Loading Trials 44

EMG Amplitudes 50

Individual Subject Data 59

Reproducibility 60
Discussion
Hypotheses
  1. Anticipation of Added Loads
  2. Effect of Fatigue of Pre-Trial EMG Activation
  3. Effect of Fatigue on Maximum Trunk angular displacement
  4. Effect of Fatigue on Peak Extensor Moments
  5. Effect of Fatigue on Peak Muscle Activity
  6. Effect of Fatigue on Time to Peak Angle, Time to Peak Moment and Time to
     Peak EMG Activity
  7. Effect of Pre-Load on Maximum Trunk angular displacement
  8. Effect of Added Load on Maximum Trunk angular displacement
  9. Comparison of the [4+24] and [16+12] Loading Conditions
Limitations and Assumptions
Main Findings
Additional Findings
Case Studies
Implications of the Research Findings

Conclusions
Recommendations for Future Research

References

Appendix A

Vita Auctoris
## List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Figure 1</td>
<td>Vertebral column (lateral view). From Agur, 1991.</td>
<td>10</td>
</tr>
<tr>
<td>Figure 2</td>
<td>Lumbar vertebra. A. Superior view. B. Lateral view. From Agur, 1991.</td>
<td>11</td>
</tr>
<tr>
<td>Figure 3</td>
<td>During flexion, the lumbar lordosis unfolds and the lumbar spine straightens. From Bogduk and Twomey, 1987.</td>
<td>12</td>
</tr>
<tr>
<td>Figure 4</td>
<td>The abdominal musculature (only right side shown). A is the rectus abdominus (anterior view); B is the external oblique (lateral view); and C is the internal oblique (lateral view). From Stone and Stone, 1990.</td>
<td>14</td>
</tr>
<tr>
<td>Figure 5</td>
<td>The lumbar back musculature. A shows the left and right thoracic lumbar erector spinae and B shows the left lumbar erector spinae (posterior view). From Bogduk and Twomey, 1987.</td>
<td>16</td>
</tr>
<tr>
<td>Figure 6</td>
<td>Frontal view of experimental apparatus (not to scale).</td>
<td>27</td>
</tr>
<tr>
<td>Figure 7</td>
<td>Sagittal view of experimental apparatus with subject (not to scale).</td>
<td>28</td>
</tr>
<tr>
<td>Figure 8</td>
<td>Representation of back harness with subject in place (overhead view).</td>
<td>28</td>
</tr>
<tr>
<td>Figure 9</td>
<td>(Top) Load carriage system. Weight plates are stacked on the middle pole. (Bottom) Altered weight plate with handles to allow added loads to be dropped to load carriage system.</td>
<td>29</td>
</tr>
<tr>
<td>Figure 10</td>
<td>(Left) Posterior view. The electrode locations and lines of action for the right lumbar erector spinae (LES) and the right thoracic erector spinae (TES) are shown. (Right) Anterior view. The electrode locations and lines of action for the right rectus abdominus (RA), right external oblique (EO) and left internal oblique (IO) are shown.</td>
<td>32</td>
</tr>
<tr>
<td>Figure 11</td>
<td>Representation of a sample trial with the extensor moment, trunk angle and EMG data of the extensor muscles depicted in the graph. This figure includes a timeline of the start and end of a trial.</td>
<td>36</td>
</tr>
</tbody>
</table>
Figure 12  Schematic of the 2*2*2 ANOVA model used for statistical analysis.

Figure 13  MnPF values for start and end of the fatiguing session. The muscles investigated were the right and left LES, and the right and left TES (n=13). Standard error bars are shown.

Figure 14  a) Maximum trunk angular displacement for the 12% and 24% added load conditions. The values presented are pooled across pre-loads, and rest and fatigue trials (n=52). Standard error bars are shown. b) The pre-load and fatigue interaction for the across-subject means of the maximum trunk flexion angle. The values are pooled across added loads (n=26). Standard error bars are shown.

Figure 15  Trunk extensor moment for each loading condition (n=26). Means are indicated for the pre-trial values (4% or 16%) and trial values (12% or 24%). The trial values represent the maximum values observed in response to the added loads. Standard error bars are shown.

Figure 16  Average trial duration (TTPK angle) for two pre-load conditions, pooled across added loads and time (n=52). Standard error bars are shown.

Figure 17  Average time to peak value for each of the measured variables, with the values pooled across added loads and time (n=52 for extensor moment and n=104 for EMG). Standard error bars are shown. The * indicates a significant difference (p<0.05) between pre-load conditions for a variable.

Figure 18  Average time to peak value for each of the measured variables, with the values pooled across pre-loads and time (n=52 for extensor moment and n=104 for EMG). Standard error bars are shown. The * indicates a significant difference (p<0.05) between added load conditions for a variable.

Figure 19  EMG amplitudes for the EO muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

Figure 20  EMG amplitudes for the IO muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.
Figure 21  EMG amplitudes for the LES muscle for each loading condition (n=13). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

Figure 22  EMG amplitudes for the TES muscle for each loading condition (n=26). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

Figure 23  EMG amplitudes for the RA muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.
List of Tables

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Table 1</td>
<td>Ranges of segmental motion in males aged 25 to 36 years (measured in degrees). From Pearcy et al., 1984.</td>
<td>13</td>
</tr>
<tr>
<td>Table 2</td>
<td>Summary of post hoc findings comparing the maximum (trial) values observed in each variable in response to the added loads. Conditions are ranked from 1, having the largest maximum values, to 4, having the lowest. Cells with the same number indicate conditions that were not significantly different from each other for a particular variable. All significant differences are at p&lt;0.05.</td>
<td>41</td>
</tr>
<tr>
<td>Table 3</td>
<td>Summary of the significant main and interaction effects for the fatigue (F), pre-load (PL) and added load (AL) independent variables. Blank spaces indicate no significant effect was found. Significance is indicated at the level of p&lt;0.05 and p&lt;0.0001.</td>
<td>42</td>
</tr>
<tr>
<td>Table 4</td>
<td>Absolute increase and the relative percent increase from rest to fatigue for the peak angle, moment and EMG amplitudes. The values summarized are for Subject #1 and #6, and pooled across loading conditions (n=4).</td>
<td>60</td>
</tr>
<tr>
<td>Table 5</td>
<td>Reproducibility results for the rested trials. Values are averaged within each condition/time combination then averaged across subjects. %CV reported for trunk extensor moment, trunk flexion angle, and trial duration. The standard deviation reported for muscle EMG data. Each value represents the average %CV across 13 subjects.</td>
<td>61</td>
</tr>
<tr>
<td>Table 6</td>
<td>Reproducibility results for the fatigued trials. Values are averaged within each condition/time combination then averaged across subjects. %CV reported for trunk extensor moment, trunk flexion angle, and trial duration. The standard deviation reported for muscle EMG data (n=13).</td>
<td>61</td>
</tr>
</tbody>
</table>
Chapter I

Introduction

Low back pain (LBP) has been a major topic of concern in Western society due to the economic and health implications that arise from it. In the United States, it has been reported that the annual medical costs for managing LBP is 24 billion dollars (Frymoyer & Cats-Baril, 1991). There is not one group of society who is immune from developing LBP. In fact, at least a quarter of the working population has its symptoms at any given period of time (Frymoyer & Cats-Baril, 1991). LBP also has a psychosocial cost since it limits the activity level of individuals and can affect people during their most productive years. There have been many proposed factors that can contribute to the occurrence of LBP (Marras et al., 1993) but strong evidence exists for mechanical risk factors associated with lifting (Marras et al., 1993), whole-body vibration (Magnusson et al., 1993), sudden unexpected movements or loads (Marras et al., 1987), muscular fatigue and repetitive movements (Ayoub, 1990; Mundt et al, 1993; Sparto & Parnianpour, 1998; van Dieen et al., 1998).

The exposure of the spine system to a load perturbation, causing a sudden response of the neuromuscular control system, will significantly increase the mechanical loads placed on the vertebral structure when compared to static loading of similar loads (Lavender et al., 1993). The observed increases in mechanical loads are a result of two factors. First, the external force applied during the sudden loading is dynamic and requires increased muscle force to minimize the effect on the trunk posture, in order to maintain stability and control the movement of the external load (Lavender et al., 1993). The second factor is the startle response, where the system over-compensates in reaction to a sudden load (Greenwood & Hopkins, 1976). This over reaction by the
neuromuscular system, in response to sudden loading, has been implicated as the source of 12.3% of accidental injuries (Mitchell et al., 1983).

The process of fatigue has been shown to affect the spine in a variety of detrimental ways. Bigland-Ritchie et al (1986) defined neuromuscular fatigue as any reduction in the maximum force generating capacity, regardless of the type of work being done. Fatigue of the muscles of the spine results in muscle cocontraction and increases the loading on the intervertebral discs due to a heightened level of muscle activation (Sparto et al., 1997). It has been found that there is a significant increase in internal oblique and latissimus dorsi muscle activity when the erector spinae muscles become fatigued during isometric trunk extension (Sparto et al., 1997). When trunk rotators are isometrically fatigued, the antagonistic trunk rotators increase activation (O'Brien & Potvin, 1997). The increased spinal compression that manifests during fatigue can result in low back disorders such as vertebral end plate fractures (Bowman et al., 1994). Fatigue can cause increased passive tissue loading (Seidel et al., 1987) and place the trunk at an increased risk for injury. This is especially significant when one considers what occurs during a sudden load to the trunk. There may be an increased risk of low back injury if the trunk muscles that are responsible for generating the forces necessary to maintain spine stability are fatigued and cannot completely counteract the torque from the sudden load (Parnianpour et al., 1988).

In one definition of mechanical stability, the spine is considered to be stable or not stable. This statement is derived from the definitions of mechanical stability noted by Bergmark (1989) and Crisco and Panjabi (1992). However, stability can also be seen from a clinical viewpoint as "the ability of the spine under physiologic loads to limit patterns of displacement so as not to damage or irritate the spinal cord or nerve roots and, in addition, to prevent uncapacitating deformation or pain due to structural changes" (White & Panjabi, 1978). Unlike the definition of mechanical stability, the concept of
clinical stability introduces the idea that the spine can have different levels of stability. The maintenance of spinal stability is extremely important in decreasing the chance of low back disorders. In the absence of muscles, an in vitro ligamentous lumbar spine is unstable at compressive loads of only 88 N (Crisco et al., 1992). However, the in vivo spine may endure values greater than 6000 N while participating in daily tasks (McGill & Norman, 1986), and up to 18000 N in competitive power lifters (Cholewicki et al., 1991) due to the stability that is provided by the trunk musculature (Bergmark, 1989) and the neural system (Tesh et al., 1987).

Panjabi (1992) introduced a model of the stabilizing system of the spine, and the model was divided into three subsystems: 1) Control, 2) Passive, and 3) Active. The Passive subsystem is comprised of the vertebrae, intervertebral discs, spinal ligaments and the passive mechanical properties of the muscle of the trunk. The Active subsystem is composed of the muscles and tendons surrounding the spine. The Control subsystem consists of neural components that receive and process input, and sends out information to maintain stability. The Passive and Active subsystems are responsible for sending sensory input to the neural components of the Control subsystem, which then sends corresponding signals to the muscles of the Active subsystem. In support of Panjabi’s model, Cholewicki and McGill (1996) revealed that the local stabilizing system of the spine, which is comprised of the passive joint properties and intrinsic muscles, did not provide adequate structural stability to support normal loading conditions. Bergmark (1989) introduced a mechanical model of the spine that was in support of the findings of Panjabi (1992). Bergmark discovered that, by increasing muscle activation of the larger, more superficial muscles of the trunk, and maintaining some activation of intrinsic muscles, the spine demonstrated increased stability during movement about the three axes of the trunk.
To overcome the torque effects of an external load and maintain stability, the spine is largely dependent on the trunk muscles. Muscle cocontraction aids in the prevention of spinal instability (Panjabi et al., 1989; Lavender et al., 1992). Antagonistic muscle contractile forces generate trunk moments to provide resistance to movement and increase stability while simultaneously loading the spine in compression (Janevic, 1991; Granata & Marras, 1995; Cholewicki & McGill, 1996). Krajcarski et al. (1999) determined that an increase in pre-load values, which corresponds to an increase in joint compression, will decrease the angular displacement of the trunk during sudden loading because of an increase in trunk stiffness and spine stability. Krajcarski et al. (1999) used trunk angular displacement as a representation of spine stability. In addition, this study determined that a greater sudden load resulted in a greater initial acceleration of flexion by the trunk that required a larger maximum extensor moment to maintain stability. Marras et al. (1987) found that expected and sudden load (where the subject does not know when the load perturbation will occur) increased the required muscle force to maintain stability, compared with static loading. In fact, that study revealed that the mean and peak muscle force was up to 200% and 70% greater, respectively, during sudden loading when compared to expected loading.

When attempting to overcome the moments associated with sudden loading, the spine maintains stability by increasing joint compression forces (Cholewicki & McGill, 1996) and increasing the level of muscle coactivation, even if it may result in an increased rate of fatigue (Zetterberg et al., 1987). In a study that determined the response of the spine to sudden loads causing sagittal flexion, it was found that abdominal muscles rapidly responded to the sudden load with a large increase in muscle activation (Krajcarski et al., 1999). Also, Krajcarski et al. (1999) concluded that increases in the EMG amplitudes of abdominal and trunk extensor muscles were determined predominantly by the added load magnitude and not the pre-loads. Chiang (1997)
conducted a similar study of sudden loads causing lateral bending. Similar to Krajcarski et al. (1999), Chiang (1997) concluded that increased muscular activity due to pre-loads leads to increases in trunk muscle stiffness and spine stability.

Krajcarski et al (1999) and Chiang (1997) have provided a greater understanding of the active muscular response that is responsible for maintaining stability. These two studies have demonstrated that a greater pre-load will result in an increase in trunk stiffness, which is beneficial to the stability of the spine when exposed to sudden loads. Unfortunately, the increased trunk stiffness also contributes to trunk muscle fatigue because an increase in muscular activity is what causes the increased stiffness of the trunk. Since muscle fatigue has an effect on the ability of the spine to maintain stability, it would be of significant interest to determine whether a fatigued trunk system responds differently to the pre- and added loads than when it is in a rested state.

**Statement of Purpose**

The purpose of this study is to investigate the effects of fatigue on the response of the trunk to a variety of conditions of initial static loads and step inputed loads causing rapid sagittal trunk flexion. Within this study, fatigue will be defined as a progressive loss in force generating capacity of the total neuromuscular system independent of the force required by task (Simonson and Weiser, 1976; Bigland-Ritchie & Wood, 1984). Therefore, as stated by Edwards (1981), fatigue will be considered to be a continuous process and not a single event defined when the desired action can no longer be completed.
Statement of Hypotheses

1. After being fatigued, the two loading conditions with a static pre-load of 4% will generate an equal level of muscular activity and equal moments. This statement is applicable for the other two loading conditions with a 16% static pre-load.

   When comparing loading conditions, if the magnitudes of the static pre-loads are the same, then the activity of the trunk muscles to counteract these loads should be equal. The experimenter will attempt to remove any cues as to when the added load will be suddenly applied. However, if the muscular activity and measured moments for the same pre-load magnitude are not equal, this may indicate that the subjects were anticipating the magnitude of the added loads.

2. The static pre-load of 4% during the rested trials will elicit less muscular activity compared to the static pre-load of 4% in the fatigued conditions. This statement is applicable for the other two loading conditions with a 16% static pre-load.

   Since the magnitude of the pre-load is based on a percentage of the maximum isometric extensor moment that is obtained while the subject is rested, the magnitude of the pre-load will represent a higher percentage of the maximum isometric extensor moment for a fatigued subject. A muscle that is fatigued is weaker and its force generating capacity is decreased (Parnianpour et al., 1987).

3. The trunk angular displacement will be greater for the loading conditions with the 4% pre-loads than the 16% pre-load conditions.

   It has been documented that lower pre-load magnitudes will result in lower spine muscle stiffness and decreased spine stability (Cholewicki & McGill, 1995). This hypothesis has been demonstrated previously for the rested spine (Chiang, 1997; Krajcarski et al., 1999)

4. The response time to the load perturbations will be longer in the fatigued state as compared to the rested state.
Wilder et al. (1996) presented data showing that the reaction time to a sudden load is longer for a fatigued trunk than it is for a non-fatigued trunk.

5. For similar added load conditions, the fatigued trunk will have greater trunk angular displacement than the rested trials.

Fatigue is expected to result in longer reaction times to the added load, slower muscle contractile speed (Bigland-Ritchie & Woods, 1984) and weakening of the trunk muscles (Parnianpour et al., 1987). Thus, the fatigued trunk will require more time to counteract the forward flexion moment caused by the added load.

This will subsequently result in larger trunk angular displacements.

6. The peak moments achieved in the fatigued trials will be higher than those achieved in the rested trials when comparing the 12% added load conditions. This statement can also be made for the 24% added load conditions.

It has been hypothesized that the reaction time to the added load and the angular displacement of the trunk will be greater during fatigue. Therefore, it could be expected that the acceleration of the trunk will be greater during the fatigued conditions. The increased need for deceleration will result in increased peak moments in order to regain spine stability.

7. Higher peak agonist and antagonist activity will be recorded when the 24% added load is introduced during the fatigued trials when compared to the rested trials. A similar relationship will be observed for the 12% step input load.

The fatigued trunk will require the recruitment of more muscle fibers to maintain stability than a rested trunk. During fatigue conditions, the antagonist muscles will increase activity, as the agonist trunk muscles become fatigued (Potvin & O’Brien, 1998). The fatigued agonist trunk muscles will have to provide resistance to the added loads, but the added load magnitudes are based on the rested maximum isometric extensor moments.
8. The change in agonist and antagonist muscle activity, in response to the added load, will be larger for the fatigued conditions than the rested conditions. The antagonist muscles are used to aid agonist muscles in the maintenance of spine stability, however, antagonist muscle activity levels are increased to protect the spine against injury from the decrease in neuromuscular control that is manifested during fatigue (Parnianpour et al., 1987). The fatigued agonist muscles will overcompensate the activity levels required to counteract the effects of a perturbation (Thelen et al., 1995).

9. When comparing the two loading conditions (4+24% and 16+12%) that both have a cumulative load of 28%, the loading condition of 4+24% will have larger values for trunk angular displacement, peak moments, peak muscle activation and time to peak trunk angle. This relationship will be present in the fatigued and rested conditions.

The 4% pre-load will require a lower level of agonistic muscle activation than the 16% pre-load. However, the lower pre-loads will result in a less rigid and stable spine (Cholewicki & McGill, 1995). The 24% added load will require a larger level of muscle activation to maintain stability than the smaller added load, and will result in larger accelerations and moments.
Chapter II

Review of Literature

The Lumbar Vertebrae

The adult vertebral column typically contains 26 vertebrae, and is divided into five regions: 7 cervical vertebrae; 12 thoracic vertebrae; 5 lumbar vertebrae; sacrum; and coccyx (Figure 1). The average adult male vertebral column is 71 cm in length and accounts for approximately 40% of the total height of the body (Tortora & Grabowski, 1996). Between adjacent vertebrae from the second cervical vertebrae to the sacrum are intervertebral discs. Each disc has an outer ring called the annulus fibrosus and an inner structure called the nucleus pulposus. The annulus fibrosus consists of alternating concentric lamellae of collagen fibers that connect adjacent vertebral bodies. The collagen of the annulus fibrosus are type I, which are fibroelastic and tensile (Bogduk & Twomey, 1987). The nucleus pulposus consists of proteoglycans and collagen. The presence of the proteoglycans allows the nucleus pulposus to retain a large amount of water. The collagen fibers of the nucleus pulposus are mainly type II, which is an elastic form of collagen commonly found in cartilage (Bogduk & Twomey, 1987).
Figure 3: Vertebral column (lateral view). From Agur, 1991.

The vertebrae of the lumbar region are the largest and strongest in the vertebral column since the amount of body weight that is supported by each vertebra increases toward the inferior end of the vertebral column. There are a number of distinctive features that characterize a lumbar vertebra (Figure 2). The superior articular processes are located medially instead of superiorly. The inferior articular processes are directed laterally instead of inferiorly. The spinous processes are quadrilateral in shape and project straight posteriorly instead of inferiorly (Tortora & Grabowski, 1996).
Figure 4: Lumbar vertebra. A. Superior view. B. Lateral view. From Agur, 1991.

The Mechanism of Forward Flexion of the Lumbar Spine

Forward flexion of the trunk requires the entire lumbar spine to rotate forward and this straightens the natural lordosis of the lower back (Figure 3). This movement requires each of the lumbar vertebrae to rotate from their backward tilted position in upright lordosis to a neutral position causing the upper and lower surfaces of vertebral bodies to be parallel to one another (Bogduk & Twomey, 1987). Also, forward rotation of the upper lumbar vertebrae increases the range of forward flexion by compressing the anterior portion of the intervertebral discs (Bogduk & Twomey, 1987).
Figure 3: During flexion, the lumbar lordosis unfolds and the lumbar spine straightens. From Bogduk and Twomey, 1987.

Rotation in the sagittal plane is not the only movement that is responsible for the forward flexion of the lumbar spine. The vertebrae of the lumbar spine also experience a forward translation during flexion (Pearcy et al., 1984). The translation movement is possible because the vertebral rotation opens a small gap between the inferior articular facet and the superior articular facet in the zygapophysial joint. The gap is closed as the vertebra slides forward due to the anterior rotation of the lumbar spine.

During spine flexion, the zygapophysial joints provide spine stability by resisting and controlling the forward rotation and forward translation of each lumbar vertebra (Pearcy et al., 1984). The ligaments of the intervertebral joints also aid in providing stability to the lumbar spine during forward flexion. The movement of each vertebra causes the supraspinous and interspinous ligaments and the ligamenta flava to become tense and provide resistance to the flexion movements (Twomey & Taylor, 1983).

Normal Ranges of Motion

The introduction of biplanar radiography technology has allowed researchers to accurately quantify the segmental motion of each vertebra within a living subject (Pearcy et al., 1984). The average sagittal rotation value for each segment of the lumbar spine is presented in Table 1 (Pearcy et al., 1984). It should be noted that the greatest range of rotation generally occurs at the L4/L5 level.
<table>
<thead>
<tr>
<th>Level</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>L1-2</td>
<td>8</td>
<td>5</td>
</tr>
<tr>
<td>L2-3</td>
<td>10</td>
<td>2</td>
</tr>
<tr>
<td>L3-4</td>
<td>12</td>
<td>1</td>
</tr>
<tr>
<td>L4-5</td>
<td>13</td>
<td>4</td>
</tr>
<tr>
<td>L5-S1</td>
<td>9</td>
<td>6</td>
</tr>
</tbody>
</table>

Table 1: Ranges of segmental motion in males aged 25 to 36 years (measured in degrees). From Pearcy et al., 1984.

**Muscles of the Trunk**

The muscles that are described in this section are those that are pertinent to this particular study. The information presented in this section has been referenced from Bogduk and Twomey (1987), Stone and Stone (1990), McMurry and Rikel (1991), and Warfel (1993).

**The Abdominal Musculature**

The muscles of the abdominal region that are of interest in this study are the rectus abdominus (RA), internal oblique (IO) and external oblique (EO).

The rectus abdominus (Figure 4A) is the most superficial muscle of the anterior abdominal wall. The RA originates from the pubic symphysis and the crest of the pubis, and inserts onto the 5th, 6th, and 7th costal cartilages and the xiphoid process. The fibers of the RA run vertically and are responsible for creating flexion of the pelvis and the vertebral column. The RA has a role in compressing the abdomen and during forced expiration.

The external oblique (Figure 4B) originates from the external surfaces of the lower eight ribs. The fibers of the EO insert into a wide aponeurosis of the anterior abdominal wall, the anterior half of the iliac crest, the pubic crest, the pubic tubercle, the
linea alba, and the xiphoid process. The fibers of this muscle run anteriorly and inferiorly along the anterolateral abdominal wall. The EO aids in flexion and axial rotation of the vertebral column. Due to its orientation, the EO has a role in compressing the abdomen and during forced expiration.

The internal oblique (Figure 4C) arises from the lumbar fascia, the iliac crest, and the inguinal ligament. The IO attaches to the lower three costal cartilages, linea alba, tendon of the pubis, ilium crest, and the thoracolumbar fascia. The main function of the IO is to flex and axially rotate the vertebral column. Similar to the previous two abdominal muscles presented, the IO aids in the compression of the abdomen and during forced expiration.

**Figure 4:** The abdominal musculature (only right side shown). A is the rectus abdominus (anterior view); B is the external oblique (lateral view); and C is the internal oblique (lateral view). From Stone and Stone, 1990.

**The Lumbar Back Musculature**

The lumbar extensor muscles that are of interest in this study are the erector spinae muscle group. The group includes three muscles: spinalis, longissimus, and iliocostalis. The function of the erector spinae muscle group is to aid in extension, lateral bending
and axial rotation of the vertebral column. In the lumbar vertebral section, only the iliocostalis and the longissimus are present since the spinalis is only present in the cervical and thoracic vertebral sections. The erector spinae muscles found in the lumbar area are divided into the thoracic erector spinae (TES) and the lumbar erector spinae (LES).

The thoracic erector spinae (Figure 5A) is composed of two muscles: the thoracic iliocostalis and the thoracic longissimus. The thoracic longissimus consists of 11 or 12 pairs of small fascicles arising from the ribs and transverse processes of T1 to T12. Each muscle belly is 7 to 8 cm in length and has 3 to 4 cm of tendon from the origin. Inferior to each muscle belly is a caudal tendon that runs into the lumbar region. The tendons insert between the levels of L3 and S3, and the posterior superior iliac spine. The oblique orientation of the muscle acts directly to laterally flex the thoracic vertebral column, and a bilateral contraction of the muscle indirectly produces an increase in lumbar lordosis.

The thoracic iliocostalis originate from the angle of the lower seven or eight ribs and inserts into the ilium and sacrum. Each fascicle is attached to the ribs by a 9 to 10 cm tendon and has a muscle belly that is 8 to 10 cm in length. The thoracic iliocostalis have no attachment to the lumbar vertebrae but span the entire lumbar section. This allows the thoracic iliocostalis to increase the lordosis of the lumbar spine.

The lumbar longissimus and the lumbar iliocostalis combine to form the lumbar erector spinae (Figure 5B). The LES tendons lie deep to the tendons of the TES. The lumbar longissimus is composed of five fascicles. The fascicles originate from the accessory process and the adjacent medial end of the dorsal surface and insert into the medial aspect of the posterior superior iliac spine. The five fascicles overlap each other with the fascicle of L5 being the deepest and most medial. These muscles act to
produce lateral flexion when contracted unilaterally, and can cause extension during bilateral contraction.

The lumbar iliocostalis is composed of four fascicles originating from L1 to L4. Similar to the lumbar longissimus, the lumbar iliocostalis is layered with the fascicle from L4 being the deepest and most medial layer. The fascicles insert onto the iliac crest lateral to the posterior superior iliac spine. The lumbar iliocostaliss act to extend and laterally bend the lumbar spine.

![Figure 5: The lumbar back musculature. A shows the left and right thoracic lumbar erector spinae and B shows the left lumbar erector spinae (posterior view). From Bogduk and Twomey, 1987.](image)

**The Electromyography – Force – Fatigue Relationships**

The electrical signal associated with the contraction of a muscle can be observed with the use of electromyogram (EMG). In previous studies, EMG has been used as an indicator to the initiation of muscle activation (Lavender et al., 1992; Deluca, 1997), to measure tension developed in the muscle (McGill & Norman, 1986; Seroussi & Pope,
1987; Hughes et al., 1994) and to determine the amount of muscular fatigue

Muscle contraction is a result of the depolarizing of the sacrolemma which
produces an electric signal that can be measured (Guyton, 1991). These signals are
used in electromyography technology to approximate levels of muscle tension.
According to Henneman et al (1965), the motor units of the muscle are recruited
according to the Size Principle. This implies that smaller motor units are activated for
tasks requiring low force contractions. As the magnitude of force is increased, so are
the number and size of the motor units being recruited. The largest motor units are
recruited at the highest level of force contractions. The activation of an increasing
number of motor units results in the increase in force generation by the muscle as well
as the summation of electrical activity and larger EMG signal amplitudes (Fuglevand et
al., 1993). Another method for the neuromuscular system to increase the level of
muscle force is to increase the rate of stimulation to the motor units. The increased rate
of stimulation will result in the increased amplitude of an EMG signal due to the
overlapping of the motor unit potentials (Fuglevand et al., 1993).

A review of studies concerning the EMG-force relationship during isometric
contractions reveals a linear relationship for some muscles and a nonlinear one for
others. According to Lind and Petrofsky (1979), the forearm muscles displayed a linear
increase in EMG amplitude as force was increased. It was revealed in a study
investigating fatiguing isometric contractions of the first dorsal interosseous muscle, that
an increase in tension is accompanied by an increase in EMG amplitude (Fuglevand et
al., 1993). Woods and Bigland-Ritchie (1983) also investigated the first dorsal
interosseous muscle and reported a linear relationship between EMG amplitude and
force. A study on the bicep brachii revealed a nonlinear response (Petrofsky et al.,
1982). A nonlinear relationship was also observed in the deltoid muscle when
comparing EMG amplitude and force (Woods & Bigland-Ritchie, 1983). The reason for
the difference in the EMG-force relationship between different muscles, during an
isometric contraction, can be attributed to the fiber composition of the different muscles
(Petrofsky et al., 1982). The EMG-force relationship during dynamic task has also been
investigated. A linear relationship was found between the predicted muscle force and
measured EMG during trunk flexions and extensions (Thelen et al., 1995).

The process of muscular fatigue is associated with the increase in EMG
amplitude since fatigue is the reduction in the force-generating capacity of the
neuromuscular system that occurs during sustained activity (Bigland-Ritchie and Woods,
1984). Fatigue occurs when the muscle is not able to supply the necessary metabolites
because of a lack in oxygen supply or the local depletion of metabolic substrates
(Winter, 1990). During fatigued conditions, the maintenance of constant tension requires
increased recruitment of new motor units to overcome the decreased firing rate of the
already recruited motor units (Maton, 1981) and will increase the amplitude of the EMG.

Many studies have reported that fatigue is responsible for altering the shape of
motor unit action potentials (Viitasalo & Komi, 1977; Hagberg, 1981; Petrofsky et al.,
1982). These studies showed that fatigue caused the EMG spectrum to shift towards
the lower frequencies. The slowing of the conduction velocity, perhaps due to the build
up of lactic acid, has also been used to explain the EMG spectral shifts observed during
muscle fatigue (Mortimer et al., 1970; Eberstein & Beattie, 1985). Another change in
EMG, that has been observed as result of fatigue, is the tendency of motor units to fire
more synchronously (Deluca, 1984). During rested conditions, motor units fire
independently of the other motor units of the same muscle causing a summation of
motor unit action potentials. While fatigued, a tremor is present during force and EMG
measurements due to the activation pattern of the motor units, since they tend to fire in
synchronized bursts.
Historically, isometric contractions have been used to measure fatigue with EMG (Viitasalo & Komi, 1977; Petrofsky et al., 1982; Jorgenson et al., 1988), however, there is increasing support to suggest that fatigue may cause changes in EMG signals that are collected during dynamic contractions (Hagberg, 1981; Potvin & Bent, 1997). Potvin and Bent used a repetitive elbow flexion-extension task to compare biceps brachii mean power frequency values during isometric and dynamic contractions and concluded that the mean power frequency is the most effective variable for quantifying localized muscle fatigue.

**Stability of the Spine**

When attempting to understand what is responsible for the maintenance of spine stability, it is important to understand that stability can be explained with a mechanical and a clinical definition. The concept of the Euler column, named after an 18th century Swiss mathematician, provides an understanding of mechanical stability. It has been stated that “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”, however, the system will fail when it is exposed to stresses that are greater than the failure stress of the material (Crisco & Panjabi, 1992). This mechanical viewpoint of spine stability indicates that the system is either stable or not stable.

The concept of spine stability in a clinical sense is not as rigidly defined. There have been several definitions that have been provided in the literature. White and Panjabi (1978) presented clinical stability as “the ability of the spine … to limit patterns of displacement so as not to damage or irritate the spinal cord or nerve roots and, in addition, to prevent uncapsicating deformation or pain due to structural changes”. Whether using the clinical or mechanical definition, stability is presented as a response to a perturbation by a system, and not as the actual perturbation.
An analysis of spine stability must include the effects of stiffness. The muscles of the trunk are responsible for supporting the spine by increasing stiffness (Cholewicki & McGill, 1996). Stiffness is a ratio of the magnitude of a perturbation (displacement) to the magnitude of the response (force) (Pope & Panjabi, 1985). Therefore, stable equilibrium will be maintained if the muscle stiffness is greater than the critical stiffness, or the minimum stiffness needed to maintain stability for a perturbation of a given magnitude (Bergmark, 1989).

The model proposed by Panjabi (1992) indicates that the stability of the spine is maintained by three subsystems. The subsystems include a Passive subsystem, an Active subsystem, and a Control subsystem. The Passive subsystem is comprised of the vertebrae, intervertebral discs, spinal ligaments and the passive mechanical properties of the muscle of the trunk. The Active subsystem is composed of the muscles and tendons surrounding the spine. The Control subsystem consists of neural components that receive and process input, and sends out information to maintain stability. The Passive subsystem is termed passive because it does not generate motion. In the neutral position, the Passive subsystem is responsible for interacting with the neural subsystem to provide information to the sensory components. When the spine is nearing the ends of the range of motion, the structures of the Passive subsystem aid in maintaining spine stability. The structures of the active subsystem are responsible for generating the forces required to keep the spine stable. The muscles and tendons are constantly relaying information to the neural subsystem regarding muscle tension and strength. The Control subsystem takes the information provided by the Passive and Active subsystems and processes the information so that the appropriate signals are sent out the control posture and movement so that stability is maintained. Each of the subsystems must function and interact appropriately with the other subsystems to ensure that the stabilizing system is functioning normally. If the
stabilizing system fails, it may be due to fatigue and/or injury to any of the three subsystems.

**Muscle Force and Muscle Stiffness**

It has been reported that short-range stiffness of a muscle is correlated to the number of myofilament cross-bridges within a given muscle (Rack & Westbury, 1974). Bergmark (1989) proposed that muscle stiffness is inversely related to muscle length. According to the sliding filament theory, muscle tension is a result of actin and myosin cross-bridges. Therefore, it should be no surprise that Rack and Westbury observed an increased in muscle stiffness with the increase in muscle tension, since both are dependant on the number of cross-bridges. Bergmark (1989) and Cholewicki and McGill (1995) also concluded that the increased stiffness due to loading is the result of a direct relation between force and stiffness of a muscle.

**The Role of the Trunk Musculature in Spine Stability**

Lucas and Bresler (1961) found that an isolated fresh spine cadaver from T1 to the sacrum, when placed in an upright neutral position with the sacrum fixed to the test table, cannot support a load that is greater than 20 N before it buckles and becomes unstable. In addition, the lumbar portion of the spine has been shown to collapse under an axial load of 90 N (Crisco et al., 1992). These loads are only a small fraction of the actual loads that the spine must be able to withstand to perform normal daily activities, revealing the importance of muscular support to provide spine stability. The musculature of the spine acts to support structures to provide stiffness and stability to an otherwise unstable spinal column (Bergmark, 1989). An increase in trunk muscle activation has been shown to decrease the range of motion and the neutral zone of the trunk (Wilke et al., 1995). A decrease in the neutral zone of the trunk is associated with an increase in spine stability (Panjabi et al., 1989). The increase stiffness may seem beneficial to
providing stability but if the stiffness is a result of the compression of the intervertebral disk, then spinal injury may occur (Dunlop et al., 1984; Janevic et al., 1991). The compression forces can cause impingements of the bony tips of the zygapophyseal joints. The increase in friction forces between the articular processes of the lumbar vertebrae will increase the resistance to motion and can lead to spondylolysis (Cyron & Hutton, 1978; Janevic et al., 1991).

Crisco and Panjabi (1991) conducted a study to evaluate the lateral stabilizing potential of intersegmental and multisegmental muscles. They revealed that the larger multisegmental muscles required lower muscle stiffness to stabilize the spine. This indicates that the larger superficial trunk muscles (eg. Erector spinae) are inherently better at stabilizing the spine. In support of this finding, it has been found that small muscles that are beside large muscles have a higher muscle spindle density than their larger counterparts (Peck et al., 1984). It is assumed that the role of the smaller muscles is to provide proprioceptive feedback and the large muscles would be responsible for the control of the spanned joints mechanically (Peck et al., 1984; Crisco & Panjabi, 1991).

**The Role of Intra-Abdominal Pressure in Spine Stability**

The intra-abdominal pressure (IAP) mechanism can increase spine stability (Tesh et al., 1987; Cresswell et al., 1994) without the additional coactivation of erector spinae muscles (Cholewicki et al., 1999). As mentioned, muscle stiffness is proportional to muscle force (Cholewicki & McGill, 1995). The activation of the muscles surrounding the lumbar spine can increase the stability and stiffness of the spine. The increased stiffness of the spine may be a result of an increase in IAP due to the cocontraction of the antagonist muscles (Cresswell et al., 1994). The cocontraction of the antagonist muscles increase the compressive forces on the spine, but the hydrostatic IAP forces acting on the spine cancels out the compressive forces due to the cocontraction. The
end result is a contracted abdomen without the increased compression associated with muscle activity. The contracted abdomen increases the overall stability of the lumbar spine (Cholewicki et al., 1999).

Cholewicki et al. (1999) determined that the activation levels of all trunk muscles determine the stability of the spine and it is not the IAP that is responsible for the stability. In an investigation into the relationship between the IAP and various trunk extension motions, the IAP correlated with the activity of the EO muscles (Marras & Mirka, 1996). The study concluded that the IAP might simply be a by-product of trunk muscle activation and co-activation.

The Effect of Fatigue on Spine Stability

The risk of injury increases when a task is performed many times throughout the day (Svensson & Andersson, 1983). It has also been shown that the isometric endurance of the trunk muscles has a greater correlation with low back injuries than trunk strength (Jorgensen & Nicolaisen, 1987). However, it is difficult to accurately pinpoint the mechanism of injury. It is unclear whether central and/or peripheral fatigue is responsible for the injuries due to fatigue. Central fatigue may occur because of a malfunction by nerve cells or inhibition of a voluntary effort (Wilder et al., 1996). Another possible cause of central fatigue may be an inhibition of motor areas elicited by nervous impulses from chemoreceptors (Aussmussen, 1979). Peripheral fatigue or muscular fatigue has also been suggested to increase the chance of injury (MacLaren et al., 1989). A fatigued muscle has a decreased force generating capacity when compared to a rested muscle, therefore, the fatigued muscle may not be able to generate the force that may be required to maintain stability.

Regardless of where fatigue occurs, the reason why fatigue is highly correlated to injury is due to the change in movement coordination and control (Wilder et al., 1996), and the decrease in maximum force capacity (Sparto & Parnianpour, 1998). During
repetitive trunk movements, it has been found that the fatigued trunk muscles became weaker and the neuromuscular system demonstrated less precision and control (Parnianpour et al., 1987). A common finding of many studies is the increased recruitment of alternate agonist and antagonistic muscle groups as a result of fatigue in the primary agonists (Thelen et al., 1995; O’Brien & Potvin, 1997; Sparto et al., 1997). The reason for the increased muscle recruitment has been hypothesized to stabilize the system by increasing lumbar compression, as the motor control of the agonist muscles declines (Parnianpour et al., 1988; O’Brien & Potvin, 1997). However, the increased activation that results in an increase in compression forces and stability is a risk factor for injury (Brinckmann et al., 1988). Another possible harmful effect of fatigue is that the loading may be shifted to passive tissues due the effects of lumbar compression (Roy et al., 1989).

**Sudden Loading and the Spine**

There has been considerable interest in understanding how the lumbar spine reacts to perturbations (Marras et al., 1987; Lavender et al., 1993; Cresswell et al., 1994; Wilder et al., 1996; Thomas et al., 1998; Krajcarski et al., 1999; Chiang, 1997). The spine reacts to perturbations as an underdamped system. The sudden need to regain stability will result in overcompensation by the trunk muscles (Lavender et al., 1993). Generally, unexpected perturbations lead to increased muscle activity and greater displacements of the trunk compared with expected perturbations (Marras et al., 1987; Lavender et al., 1993; Cresswell et al., 1994; Thomas et al., 1998). The increased muscle activity associated with unexpected perturbations is responsible for increased compressive loads on the lumbar spine, and increases the risk of injury (Lavender et al., 1993). However, repeated perturbations result in the person decreasing the muscular response and decreasing the trunk displacement (Lavender et al., 1993).
Krajcarski et al. (1999) and Chiang (1997) conducted studies to analyze the response of the spine to unexpected perturbations when exposed to different pre-load and added load magnitudes. Both studies concluded that pre-activation of the trunk extensor muscles aids in reducing the amount of angular displacement caused by a rapid loading perturbation. Also, common to both studies was the observation that antagonist muscles increase levels of activation rapidly in response to a sudden load. The increase in activation levels was especially great during the low pre-load conditions. It was evident from the studies that the low pre-load/high added load resulted in greater instability and peak moments about L4/L5 when compared to the high pre-load/low added load condition. The high pre-load conditions resulted in greater trunk stiffness, which provided increased spine stability (Cholewicki & McGill, 1996).
Chapter III

Methods and Procedures

Subjects

The subject pool for this study consisted of thirteen males recruited from the university population. Only individuals who have had no prior episode of a low back injury were allowed to participate in the study. A consent form was presented to all subjects prior to the start of the experiment. The completion of the consent form was an indication that the person had a full understanding of the requirements of the experiment. The Graduate Committee of the School of Human Kinetics approved the research study prior to the start of the data collection.

The subjects had an average age of 23.5 ± 3.6 years, height of 178.0 ± 5.2 cm and mass of 74.3 ± 6.3 kg.

Experimental Apparatus

The base frame of the testing apparatus (Figure 6 and 7) measured 215 cm * 115 cm * 115 cm. The frame was bolted to the floor and to the wall using ½” concrete bolts. Two crossbeams spanned the width of the frame. These beams were height adjustable, with height settings every 5 cm. A displacement transducer was firmly mounted onto the upper crossbeam, at a location close to the midline. A ball-bearing pulley was attached to the upper crossbeam. Welded onto the lower crossbeam was a 78 cm metal bar with padding at its free end. The padding was the contact point for the subject’s pelvis during trials. The back harness was made from a 115 cm long wood piece. Eye socket bolts were screwed into the ends of the wood piece so that a ¼” plastic coated steel cord was fastened to both ends of the wood piece (Figure 8). The middle section of the wood piece was padded to provide comfort to the subject. The back harness was placed around the chest and arms of the subjects at armpit level. A chain system was attached
to the back harness. A force transducer was attached in series with the chain system. The chain system also included a turnbuckle to compensate for different chest sizes between subjects. This chain system was adjustable so that it could be easily attached to the upper cross bar for MVC trials or configured for sudden loading trials. When the testing apparatus was set up for sudden loading trials, the chain system was attached to a plastic coated steel cord that ran over the pulley and attached to the load carriage.

The 4.1 kg load carriage (Figure 9) was a welded steel object with a base that supported steel plates. The load carriage was designed to allow for the rapid addition and removal of steel plates. A steel plate with rope handles was used to efficiently add and remove the steel plates that represented the sudden load (Figure 9). To prevent anticipation of the load drop, a curtain was used to block the participants' view of the experimenter (Figure 6) and the load carriage. The collection computer was also located out of the subject's line of vision.

![Diagram of experimental apparatus](image)

**Figure 6:** Frontal view of experimental apparatus (not to scale).
Figure 7: Sagittal view of experimental apparatus with subject (not to scale).

Figure 8: Representation of back harness with subject in place (overhead view).
Figure 9: (Top) Load carriage system. Weight plates are stacked on the middle pole. (Bottom) Altered weight plate with handles to allow added loads to be dropped to load carriage system.

Data Acquisition

The protocol for the collection session was designed to determine the effects of fatigue on the response of the trunk to loading conditions causing rapid flexion in the sagittal plane. Recent studies have investigated the response of the rested trunk to different levels of muscle pre-activation and added loads causing rapid flexion and this study attempted to provide insight on the response of a fatigued trunk that was exposed to the same conditions.

Each subject was required to attend two sessions. The first session was an orientation session to allow the subject to become familiar with the task that needed to be completed during the collection session. The orientation session was used as the forum whereby the tester can fully explain the protocol of the experiment and the subject
can have any of their questions answered. The second session was the collection
session, and this took place one to two days after the orientation session. The collection
session consisted of a collection of MVC trials and the data collection for the induced
rapid flexion.

**Orientation Session**
The tester provided the subject with an overview of the testing protocol and thoroughly
explained what was required of the subject. The subject was fitted with the back
harness and placed in the testing apparatus. The apparatus' crossbeams were adjusted
so that they were at a proper height for the subject. The padded support on the lower
cross beam was in contact with the subject’s pelvis, and the upper cross beam was at a
height that allowed the chain system to be parallel to the ground when the back harness
was fitted at armpit level. The height settings for the crossbeams were recorded for
reference during the actual testing session. The chain system was set up for MVC trials
and adjusted in length using the turnbuckle to ensure that the subject was in the
apparatus comfortably and without trunk flexion before the start of the trials. The subject
was instructed to perform isometric maximum voluntary contractions so that they
became familiar with the action of exerting a MVC. For each of these exertions, the
subjects was asked to increase the force up to maximum, and maintain that level of
force for 2 to 3 seconds. The maximum force that was selected was the maximum
sustainable force. The moment arm from L4/L5 to the middle of the back harness was
measured. The maximum force and the moment arm distance were used to calculate
the maximum isometric extensor moment (IEM\text{MAX}). Then, the subject was required to
perform a short version of the rapid flexion task. The loads that were used for this short
task were 4%, 12%, 16% and 24% of IEM\text{MAX}. The four loading conditions for the task
consisted of a preload (4 or 16% IEM\text{MAX}) and an added load (12 or 24% IEM\text{MAX}). Each
loading condition was achieved by placing an added load on a preload (eg. 4 + 12 =
The purpose of the short trial session was to allow the subject to become familiar with being in the apparatus and to experience the loads used during the collection session.

**Collection Session**

The collection session was approximately 2 hours in duration and took place between 24 and 48 hours after the orientation session.

**Set-up**

The apparatus was prepared according to the crossbeam height settings that were recorded from the orientation session. Myoelectric signals were collected with the use of recording surface electrodes (AgAgCl bipolar electrodes, Medi-Trace disposable ECG electrodes, Graphic Controls). Prior to the application of surface electrodes, the skin was prepared by shaving hair and then slightly abrading the shaved area with cloth soaked in alcohol. Pairs of surface electrodes were placed on the muscle belly with a center to center distance of 3 cm for the five muscles. The muscles of the left and right sides of the body were simultaneously investigated. The locations of the electrodes were the placement sites described by McGill (1991) (Figure 10). The sites for the electrodes were as follows: lumbar erector spinae (3 cm lateral to spinous process at the level of L3; thoracic erector spinae (4 cm lateral to spinous process at level of T9); external oblique (between ASIS and caudal border of rib cage, 15 cm lateral to the umbilicus; internal oblique (below the external oblique electrode site and superior to the inguinal ligament); and rectus abdominus (3 cm lateral to the umbilicus). A ground electrode was placed over the ribcage on the subject’s right side.
**Figure 10:** (Left) Posterior view. The electrode locations and lines of action for the right lumbar erector spinae (LES) and the right thoracic erector spinae (TES) are shown. (Right) Anterior view. The electrode locations and lines of action for the right rectus abdominus (RA), right external oblique (EO) and left internal oblique (IO) are shown.

**MVC Trial Session**

This session began once the EMG electrodes have been placed on the proper locations of the body. The collection of MVC values for every muscle investigated was a very critical step for this study. The goal of this session was to generate the largest amplitudes of myoelectric activity for each muscle, which was used to normalize the EMG signals collected during the sudden loading trials. The subjects were required to perform four different MVC tasks, and there were three trials for each task (Krajacrski et al., 1999). For each MVC, the participant was asked to gradually ramp the force up to a maximum and to avoid ballistic movements. The first three tasks required the subject to be sitting on the floor with their trunk raised 60° from the floor and their knees bent at a 45° angle, and their feet firmly secured in place. The tasks performed in this position were a forward flexion, left twist, and right twist. During the trials for each of the three MVC tasks described, the experimenter provided resistance to the subject’s shoulders to
ensure that the task was isometric. For example, during the forward flexion MVC task, the resistance was placed on the shoulders preventing the subject from increasing the trunk floor angle from the original 60° position and the feet were anchored to prevent movement. These three MVC tasks were performed to collect MVC values for the rectus abdominus, internal oblique and external oblique muscles. The final MVC task took place in the apparatus. With the apparatus setup to collect IEM_{MAX} values, the subject performed maximal isometric trunk extensions. This MVC task generated maximum values for the thoracic and lumbar erector spinae. During this MVC task, the maximum isometric extensor moment (IEM_{MAX}) was also calculated. This value was collected by having the force transducer in series with the chain system that connects the harness to the apparatus. The IEM_{MAX} was calculated by multiplying the maximum force value with the vertical distance of the harness from the level of L4/L5. The IEM_{MAX} was used to determine the load magnitudes used for the preload and added loads in experimental trials.

**Testing Session**

The apparatus setup was modified from the MVC trial session. The chain system was no longer anchored to the cross bar of the apparatus. Instead, the chain system was extended to run over the pulley and connect to the load carriage system. The load carriage system was designed to support a preload and also allowed for placement of the added loads that caused forward flexion perturbations. The height, from which the added loads were dropped onto the load carriage system, was kept constant at 1 cm. Throughout the trials, the tester was responsible for ensuring that the drop distance of 1 cm was maintained. The load carriage system was designed to allow the tester to quickly add and remove weight plates from the load carriage system (Figure 9). The added loads were added and removed from the load carriage using a weight plate with a handle system (Figure 9). Once the subject had been placed into the apparatus, a
displacement transducer was attached to the front of the harness so that trunk
displacement could be recorded for each trial. The data collected from the displacement
transducer was displayed on the screen of an oscilloscope (Type 454, Textronix, Inc.,
Portland, Oregon). The subject was instructed to maintain a neutral upright posture, and
the tester adjusted the position of the signal on the oscilloscope screen so it rested
between the blacked out regions. The subject was required to keep the displacement
transducer between the blacked out regions of the oscilloscope screen prior to, and
immediately after, each loading trial.

The loads that were used for this session are based on the $IEM_{\text{MAX}}$. The testing
session consisted of four different load variations, using preloads of 4% or 16% and
added loads of 12% or 24% of the $IEM_{\text{MAX}}$. The final load values were 16% [4 + 12],
28% [16 + 12] and [4 + 24], and 40% [16 + 24] of the $IEM_{\text{MAX}}$. The different load
combinations were presented in a random order.

The testing session was organized so that data were collected for all four load
variations to determine the response of a rested trunk to the load conditions. For each
of the four load variations, 12 trials were collected. After the 12 trials, there was a two
minute rest period before the beginning of the next group of trials, and this prevented the
effects of muscle fatigue from affecting the "rested" trials. The purpose of the next
phase of the testing session was to fatigue the extensor muscles of the trunk. The
subject was required to resist forward flexion and maintain an isometric, upright posture
while a mass that is equivalent to 50% $IEM_{\text{MAX}}$ is in the load carriage system. The
subject was instructed to keep the displacement transducer signal in the viewable region
to the oscilloscope to ensure that the subject maintained an isometric, upright posture
throughout the fatiguing session. The fatiguing exercise was stopped once the subject
showed visible signs of fatigue near exhaustion levels. Once the subject was fatigued,
the trials for the four loading combinations were re-collected. One of the loading
conditions was randomly selected, and the 12 trials were executed. Immediately following, the next randomly selected load condition was carried out for 12 trials. Upon completion of trials for two of the loading conditions, the subject was again fatigued. The load carriage was loaded with a mass that represents 50% \( IEM_{\text{MAX}} \) and the subject was required to maintain an upright posture until the level of fatigue in the extensor muscles of the trunk near exhaustion levels. The sudden loading trials were repeated for the remaining two load conditions. Once the 12 trials for the fourth loading condition were completed, the testing session was also finished.

**Data Treatment**

**Trial Selection and Windowing**

All of the trials collected were scrutinized through visual inspection to determine if they were free of signal irregularities as a result of subject anticipation or an inconsistent drop of the load. Any trials with such irregularities were removed from the analysis. Trials were removed from analysis if upon investigation of the EMG data of the lumbar erector spinae or thoracic erector spinae, there were any burst of activity occurring during the pre-trial phase that may result from subject anticipation. The moment and angular displacement curves for each trial needed to be smooth to be considered as a trial suitable for analysis (Figure 11), since non-smooth data may have indicated ballistic loading.

During data processing, the beginning of each trial was determined as the instant the force increased beyond a predetermined threshold. This threshold was set to be a value that was 5% of the \( IEM_{\text{MAX}} \) above the preload level. Each windowed trial included data from 0.5 s before the threshold was exceeded (response to added load) to 1.0 s after the threshold was exceeded (Figure 11).
Figure 11: Representation of a sample trial with the extensor moment, trunk angle and EMG data of the extensor muscles depicted in the graph. This figure includes a timeline of the start and end of a trial.

EMG, Force & Displacement Data Treatment

All signal treatments were completed digitally post-collection using LabVIEW software, except for EMG signals which were initially analog high pass filtered at a cut-off frequency of 15 Hz. Using LabVIEW software, EMG was digitally rectified and low pass filtered using a 2\textsuperscript{nd} order Butterworth filter with a low cut-off frequency of 3.0 Hz. The filtered EMG was normalized to the % MVC by dividing the maximum activation of a muscle, derived from the MVC trials, and multiplied by 100%.

The force and displacement transducers were calibrated for accuracy prior to data collection, using known weights and distances. These tests provide calibration factors that were used to calibrate force measurements to Newtons and displacement measurements to centimeters. The data from the force and displacement transducers
were digitally filtered using the LabVIEW software. The collected data was low pass filtered using a 2\textsuperscript{nd} order Butterworth dual pass filter. A low cut-off frequency of 31.17 HZ (effective 25 Hz cut-off) is used for the force signals. The displacement data had a low cut-off of 7.48 Hz (effective 6 Hz cut-off, 4\textsuperscript{th} order filter) (Winter, 1990). As mentioned earlier, the forces were multiplied by the vertical moment arm (distance from L4/L5 to the centre of the chest strap) to determine the moments of force. The linear displacement values that were calculated from the displacement transducer were converted to angles to determine the magnitude of forward flexion of the trunk. The conversion to angles was achieved using a trigonometric relationship between the vertical moment arm and the instantaneous linear displacement at load application. This trigonometric relationship can only be made if it was assumed that the trunk was a rigid structure.

\textbf{Dependant Variables}

The dependant variables that will be used for this study were similar to those used by Krajcarski (1999). The dependant variables will be divided into two categories:

1) Pre-trial variables that will be calculated using the mean of the 15 ms prior to the start of each trial.
   a) Average EMG levels of the ten muscles.

2) Trial variables
   b) Peak EMG levels of the ten muscles.
   c) Peak extensor moments.
   d) Peak trunk angle.
   e) Trial duration.
   f) Time to peak EMG levels of the ten muscles
   g) Time to peak extensor moments.
Data Analyses

All analyses were based on a randomized complete block design as each participant was exposed to each loading condition, with the order of presentation randomized. Each participant served as a block.

To determine the reliability of the data collected, the measurements of variability (standard deviations and percent covariance) were gathered for each measured variable.

A multivariate analysis of variance (MANOVA) was completed on all of the dependant variables. If the MANOVA revealed significant differences due to the effect of fatigue, then the model used for the analysis of the measured dependant variables consisted of a 2*2*2 within subject design: pre-load (4%, 16% IEM$_{max}$), added load (12%, 24% IEM$_{max}$) and muscle fatigue state (rest, fatigue) (Figure 12). The statistical analysis was performed using 0.05 level of significance. If there was no differences found due to fatigue for a dependant variable, then the rested and fatigued conditions were pooled together for the data analysis. The model determined if there were significant main effects and/or interactions between independent variables. Orthogonal means comparisons were used as post hoc tests when interactions existed. The significance for these contrasts was based on the Bonferroni t-statistic to ensure a familywise error rate of $\alpha=0.05$. 

Figure 12: Schematic of the $2^3$ ANOVA model used for statistical analysis.
Chapter IV

Results

The data presented in this chapter are grouped into two sections: the grouped data from the thirteen subjects and case studies from individual subjects. The grouped data will be presented as mean ± 1 standard deviation.

Upon review of the data compiled, it was revealed that very few of the dependant variables revealed any significant effects of fatigue. Therefore, the general presentation of the grouped data sets, except for the TES EMG and trunk flexion data, will have the rested and fatigued data pooled together, to provide a more concise presentation. A main effect of fatigue was found for the TES peak EMG amplitude, therefore, the rest and fatigued data sets will be presented separately for that muscle. Also, a significant interaction was found between fatigue and pre-loads in the peak trunk flexion data and will be presented with the fatigued and rested data sets separated. The other variables that had significant differences between the rested and fatigued conditions will be highlighted throughout the course of the results section. In addition, the bilateral EMG data collected from the five muscles were pooled across right and left sides because side had no significant effect for any dependant EMG measures.

The post-hoc results comparing the peak values of each variable are presented in Table 2. The data sets being compared in the table are the group means, pooled across rest and fatigue and across the left and right sides for the EMG data.

A summary of the statistically significant findings in the study is displayed in Table 3.
<table>
<thead>
<tr>
<th>Dependant Variable</th>
<th>[4+12]</th>
<th>[4+24]</th>
<th>[16+12]</th>
<th>[16+24]</th>
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<td>1</td>
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</tr>
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</table>

Table 2: Summary of post hoc findings comparing the maximum (trial) values observed in each variable in response to the added loads. Conditions are ranked from 1, having the largest maximum values, to 4, having the lowest. Cells with the same number indicate conditions that were not significantly different from each other for a particular variable. All significant differences are at p<0.05.
<table>
<thead>
<tr>
<th></th>
<th>Main Effects</th>
<th>Interactions</th>
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<td>Pre-Load</td>
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<td><strong>Pre-Trial EMG</strong></td>
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<tr>
<td>RA</td>
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<td><strong>Maximum</strong></td>
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<td>p&lt;0.0001</td>
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<tr>
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</tr>
<tr>
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<td></td>
</tr>
<tr>
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<td>p&lt;0.0001</td>
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<tr>
<td>RA</td>
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<td></td>
</tr>
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</table>

Table 3: Summary of the significant main and interaction effects for the fatigue (F), pre-load (PL) and added load (AL) independent variables. Blank spaces indicate no significant effect was found. Significance is indicated at the level of p<0.05 and p<0.0001.

**Grouped Subject Data**

**Fatigue Assessment**

The purpose of this experiment was to determine the effects of fatigue on the response of the trunk to different combinations of initial static loads and step inputed loads causing
rapid sagittal trunk flexion. Therefore, it is important to demonstrate that the subjects were adequately fatigued from the prolonged contraction during the fatiguing session. The measurement of EMG mean power frequency (MnPF), from the trunk extensor muscles, was used as an indicator of muscular fatigue. The MnPF values for the start and end of the fatiguing session are shown for trunk extensor muscles (Figure 13). Changes are shown as a percentage of the rested values. During the fatiguing session, the right LES had a MnPF decrease of $24.6 \pm 14.1\%$ and the left LES decreased by $24.2 \pm 13.5\%$. The MnPF decrease for the right TES was $15.5 \pm 11.6\%$ and the decrease for the left TES was $15.3 \pm 10.7\%$.

![MnPF Values](image)

**Figure 13**: MnPF values for start and end of the fatiguing session. The muscles investigated were the right and left LES, and the right and left TES (n=13). Standard error bars are shown.
Mechanical Variables from the Sudden Loading Trials

Maximum Angular Displacement of the Trunk

Figure 14a summarizes the group means for the maximum angular displacement of the trunk for the added load conditions, pooled across pre-loads and fatigue levels. Figure 14b presents the pre-load and fatigue interaction for the group means of the maximum angular displacement of the trunk. The post hoc results for the maximum angular displacement data are presented in Table 2. There was a significant main effect of pre-load with the 4% condition having greater maximum angular displacement than the 16% condition (p<0.05). There was also a significant main effect of added load with the 24% condition having higher maximum trunk angles than the 12% condition (p<0.001). When comparing the maximum trunk angle measured for the two loading conditions with similar final loads, the [4+24] loading condition had a maximum trunk angle that was 96% greater than the [16+12] loading condition’s maximum trunk angle (p<0.05).

The peak trunk flexion angle was the only mechanical variable that was affected by fatigue. There was a significant interaction (p<0.05) between fatigue and pre-load with maximum trunk flexion for the fatigued 4% pre-load trials being 6.5% greater than the peak flexion during the rested 4% pre-load trials (Table 3).
Figure 14: a) Maximum trunk angular displacement for the 12% and 24% added load conditions. The values presented are pooled across pre-loads, and rest and fatigue trials (n=52). Standard error bars are shown. b) The pre-load and fatigue interaction for the across-subject means of the maximum trunk flexion angle. The values are pooled across added loads (n=26). Standard error bars are shown.
**Trunk Extensor Moments**

Among the 13 subjects, the average maximum isometric extensor force was 939.2 ± 150.4 N and the average moment arm length from the back harness to the L4/L5 joint was 0.29 ± 0.01 m. The average IEMmax was calculated to be 267.5 ± 42.1 Nm.

Maximum moment increased with increasing pre-load (p<0.0001) and added load (p<0.0001). Figure 15 summarizes the across subject means of the maximum moments of force for each loading condition, and the post hoc results are presented in Table 2. When the data were pooled across pre-loads, the 24% added load trials had a 56.6% greater maximum extensor moment than the 12% added load trials (p<0.0001). For the two loading conditions with the same final load, the [4+24] condition had a significantly greater maximum extensor moment than the [16+12] condition (p<0.05).
Figure 15: Trunk extensor moment for each loading condition (n=26). Means are indicated for the pre-trial values (4% or 16%) and trial values (12% or 24%). The trial values represent the maximum values observed in response to the added loads. Standard error bars are shown.
**Trial Duration**

The trial duration for each condition is shown in Figure 16 and the post hoc results are presented in Table 2. The 4% pre-load trials had shorter trial durations than the 16% pre-load trials ($p<0.05$), when pooled across added load and time. However, this average difference was only 27 msec (6%). The [16+24] loading condition had a significantly longer trial duration when compared to each of the other loading combinations ($p<0.05$). There was no significant difference observed between the [4+12], [4+24] and [16+12] conditions.

![Trial Duration](image)

**Figure 16:** Average trial duration (TTPK angle) for two pre-load conditions, pooled across added loads and time ($n=52$). Standard error bars are shown.
Times to Peak

In Figure 17, the TTPKs for the different variables are shown as the values are pooled across added loads and time. Peak moment and all five muscle groups showed significantly higher times to peak for the 16% pre-load than the 4% pre-load ($p<0.05$).

![Time to Peak (Effect of Pre-load Magnitude)](image)

**Figure 17:** Average time to peak value for each of the measured variables, with the values pooled across added loads and time ($n=52$ for extensor moment and $n=104$ for EMG). Standard error bars are shown. The * indicates a significant difference ($p<0.05$) between pre-load conditions for a variable.

However, only the EMG activity of the EO, IO, and LES had TTPKs that were significantly longer for the 24% added load conditions than the 12% added load conditions ($p<0.05$) (Figure 18). Conversely, the 12% added load conditions had TTPK values that were longer than the TTPKs for the extensor moments of the 24% added load.
Figure 18: Average time to peak value for each of the measured variables, with the values pooled across pre-loads and time (n=52 for extensor moment and n=104 for EMG). Standard error bars are shown. The * indicates a significant difference (p<0.05) between added load conditions for a variable.

When comparing the [4+24] and [16+12] conditions, the only muscle that had a significant difference in TTPK values for EMG activity was the LES, with the [16+12] combination having a greater TTPK than the [4+24] combination (p<0.05). The TTPKs for the extensor moments were greater for the [16+12] compared the [4+24] condition (p<0.05).

EMG Amplitudes
The pre-trial and trial maximum EMG amplitudes for each loading condition of the EO, IO, LES, TES and RA are presented in Figures 19 through 23, and the post hoc results are summarized in Table 2. All the muscles except for the RA showed a main effect of added load on peak EMG amplitudes. All the muscles except for the RA and IO showed a main effect of pre-load on peak EMG amplitudes. Only the TES and LES showed a main effect of pre-load on pre-trial EMG amplitudes.
As indicated in the post hoc findings (Table 2), only the LES was found to show evidence of subjects anticipating the magnitude of the added load due to the significant difference in pre-trial activation levels between the 12% and 24% added loads (p<0.05).

In the following subsections, each of the muscles analyzed will be discussed in detail. It should be noted that the muscles were analyzed bilaterally. However, the trunk movement associated with this study was purely trunk extension and flexion, and there were generally no differences between left and right EMG amplitudes for a muscle. Thus, bilateral data were pooled for each muscle.

External Oblique

With regards to the maximum activation level, there was a significant main effect of pre-load magnitude (p<0.05) (Figure 19). However, the loading conditions with the 16% pre-load had maximum activation levels that were an average of only 1 %MVC greater than the 4% pre-load. This difference was not considered to be functionally relevant. The loading conditions with the 24% added load demonstrated significantly higher maximum activation levels compared to the 12% added load conditions (p<0.0001). However, the average difference was small at 1.8 %MVC.

When comparing the maximum activation levels of the two loading conditions with the same final load, a significant difference was observed to exist (p<0.05). However, the maximum muscle activation during the [4+24] condition was only 1.1 %MVC greater than the [16+12] condition.

The EO was the only muscle that had a main effect of fatigue for the pre-trial muscle activation levels (p<0.05). However, the fatigued conditions had pre-trial EMG activity levels that only an average of 0.3 %MVC greater than the rested conditions.
**Figure 19**: EMG amplitudes for the EO muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

**Internal Oblique**

An analysis of the maximum activation level shows that the loading conditions with the 24% added load had 2.4 %MVC higher peak levels than the 12% added load conditions (p<0.05) (Figure 20).

A comparison of the two loading conditions with the same final load shows a significant difference (p<0.05) in peak activation levels for the IO. The [4+24] condition
elicited a 2.1 %MVC greater peak muscle activation level than the [16+12] condition (p<0.05). The pattern of peak muscle activation for IO is very similar to the peak levels of the EO. For both muscles the peak values rarely got above 10 %MVC. Generally, these two muscles did not increase pre-activation in response to an increase in pre-load magnitude.

Figure 20: EMG amplitudes for the IO muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.
Lumbar Erector Spinae

The muscle activity response of the LES to the pre-loads and added loads (Figure 21) was similar to the response observed in the trunk extensor moment data (Figure 15).

The magnitude of the pre-load had a significant effect on the pre-trial muscle activation levels \((p<0.0001)\), with the 16\% pre-load having muscle activity levels that were greater than the 4\% pre-load by an average of 6.7 \%MVC.

The data from the maximum EMG activation level also showed significant effects due to pre-load and added load magnitude. The loading conditions with the 16\% pre-load had an 8.4 \%MVC greater peak activity level than the 4\% pre-load condition \((p<0.0001)\). The loading conditions with the 24\% added load demonstrated maximum activation levels that were 12.7 \%MVC higher compared to the 12\% added load conditions \((p<0.0001)\).

The \([4+24]\) condition had a peak level that was 4.2 \%MVC\% greater than the \([16+12]\) condition \((p<0.05)\).

There was a significant interaction between fatigue and pre-load in the pre-trial muscle activity for the LES. The rested 4\% pre-load condition had a greater level of pre-trial EMG activity than the fatigued 4\% pre-load condition, but this difference was only 1.4 \%MVC.
Figure 21: EMG amplitudes for the LES muscle for each loading condition (n=13). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

Thoracic Erector Spinae

The muscle activity of the TES (Figure 22), much like the LES, exhibited a pattern that was similar to the response of the trunk extensor moment to the pre-loads and added loads.

The TES showed a fatigue effect (p<0.05) for the peak EMG activity. And thus, the rested and fatigued EMG data for the TES are presented separately. On average,
the fatigued trials had a 4.2 %MVC larger peak EMG activity than the rested trials. Also, a significant interaction (p>0.05) was found between fatigue and pre-load for the pre-trial muscle activity of the TES. However, the differences in pre-trial muscle activity for rested and fatigued trials were small (Figure 22). However, similar to the LES, the pre-trial muscle activity of the rested TES was significantly different (p<0.0001) for the two sets of loading conditions with different pre-load masses. The trials with the 16% pre-load had a pre-trial muscle activity levels that were an average of 5.6 %MVC greater than the trials with the 4% pre-load. For the fatigued TES, the pre-trial EMG activity was 7.5 %MVC higher for the 16% pre-load compared to the 4% pre-load.
**Thoracic Erector Spinae**

Figure 22: EMG amplitudes for the TES muscle for each loading condition (n=26). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.

**Rectus Abdominus**

The EMG activity of the RA muscle did not significantly change regardless of loading condition (Figure 23). The pre-load activity level was the same for all four loading conditions. The different loading conditions did not affect the level of peak muscle activity. In fact, the peak muscle activity rarely went above 8 %MVC.

Unlike the other muscles, the RA did not exhibit any difference in peak activity for the two loading conditions with the same final load.

For the RA, the level of pre-trial muscle activation was significantly higher for the fatigued 4% pre-load condition when compared to the rested 4% pre-load condition (p<0.05). Also, the level of pre-trial muscle activation of the RA was significantly greater
for the fatigued 16% added load condition compared to the rested 16% added load condition (p<0.05). However, both of these fatigue-related differences were too small to be functionally significant.

Figure 23: EMG amplitudes for the RA muscle for each loading condition (n=52). Average value is indicated for the pre-trial values (4% or 16%) and the trials or maximum values observed in response to the added loads (12% or 24%). Standard error bars are shown.
Individual Subject Data

As stated earlier, the pooled data of all the subjects did not reveal many variables where fatigue caused a change in response to the different loading conditions. Thus, it did not appear that the fatigue resulted in global changes in most of the variables. However, subject #1 and subject #6 did have results that indicated consistent differences between fatigued and rested conditions.

Table 4 summarizes the absolute %MVC increase in peak EMG amplitude and the relative percent increase in the maximum EMG activity when comparing rested and fatigued conditions for Subject #1 and #6. Generally, compared to the rested conditions, these two subjects demonstrated greater maximum muscle activation during the fatigued conditions. Both subjects demonstrated large increases in the trunk extensor muscles (LES and TES) and smaller increases for the antagonist muscles (Table 4).

For the peak trunk flexion angle data, only subject #6 showed consistent differences between fatigue and rest for all the loading conditions. On average, the fatigue loading conditions resulted in a 21% relative increase in peak trunk flexion when compared to the rested loading conditions (Table 4).

Fatigue was not associated with any changes in the peak trunk extensor moments for any of the subjects, including subject #1 and #6.
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</tr>
<tr>
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<td>5.3 %MVC</td>
<td>16.0 %MVC</td>
</tr>
<tr>
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<td>5.1 %MVC</td>
<td>14.7 %MVC</td>
</tr>
<tr>
<td>Right RA</td>
<td>0.5 %MVC</td>
<td>0.2 %MVC</td>
</tr>
<tr>
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<td>0.3 %MVC</td>
</tr>
<tr>
<td>Trunk Angle</td>
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<td>2.4 deg.</td>
</tr>
<tr>
<td>Moment</td>
<td>1 Nm</td>
<td>-1 Nm</td>
</tr>
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</table>

**Table 4:** Absolute increase and the relative percent increase from rest to fatigue for the peak angle, moment and EMG amplitudes. The values summarized are for Subject #1 and #6, and pooled across loading conditions (n=4).

**Reproducibility**

Table 6 and 7 show the average within subject coefficients of variation as a percentage of the mean (%CV) for the trunk extensor moment, trunk flexion angle and trial duration variables; and the standard deviation for the muscle EMG results. These data are used to represent the reproducibility or reliability of each variable for each condition. The standard deviation value was used to demonstrate reproducibility for EMG data because of the low mean values observed, and the subsequently high %CV values even when standard deviations were very low. The data presented is the within subject variability across the ten trials for each condition.
<table>
<thead>
<tr>
<th></th>
<th>Rest</th>
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<th></th>
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<tr>
<td></td>
<td>4+12 Pre-Trial</td>
<td>Maximum</td>
<td>4+24 Pre-Trial</td>
<td>Maximum</td>
<td>16+12 Pre-Trial</td>
<td>Maximum</td>
<td>16+24 Pre-Trial</td>
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<tr>
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</tr>
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<td></td>
<td></td>
<td></td>
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<tr>
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<td>7.5</td>
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<td></td>
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<td></td>
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<tr>
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<td>1.7</td>
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<td>2.3</td>
</tr>
<tr>
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<td>3.9</td>
<td>1.7</td>
<td>5.0</td>
<td>1.7</td>
<td>3.9</td>
<td>1.9</td>
</tr>
<tr>
<td>Right TES</td>
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<td>5.3</td>
<td>2.0</td>
<td>6.5</td>
<td>2.2</td>
<td>5.5</td>
<td>2.1</td>
</tr>
<tr>
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<td>5.7</td>
<td>2.7</td>
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<tr>
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<td>1.0</td>
<td>0.4</td>
<td>0.8</td>
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</table>

**Table 5:** Reproducibility results for the rested trials. Values are averaged within each condition/time combination and then averaged across subjects. %CV reported for trunk extensor moment, trunk flexion angle, and trial duration. The standard deviation reported for muscle EMG data. Each value represents the average %CV across 13 subjects.

<table>
<thead>
<tr>
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<th></th>
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<td>4+24 Pre-Trial</td>
<td>Maximum</td>
<td>16+12 Pre-Trial</td>
<td>Maximum</td>
<td>16+24 Pre-Trial</td>
</tr>
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<td>8.2</td>
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<tr>
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</tr>
<tr>
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<td>0.6</td>
</tr>
<tr>
<td>Right IO</td>
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</tr>
<tr>
<td>Right LES</td>
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<tr>
<td>Left LES</td>
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<tr>
<td>Left TES</td>
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<td>8.0</td>
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<td>Right RA</td>
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<td>0.7</td>
<td>0.5</td>
<td>0.6</td>
<td>0.4</td>
</tr>
</tbody>
</table>

**Table 6:** Reproducibility results for the fatigued trials. Values are averaged within each condition/time combination and then averaged across subjects. %CV reported for trunk extensor moment, trunk flexion angle, and trial duration. The standard deviation reported for muscle EMG data (n=13).
Chapter V

Discussion

This study was conducted to investigate the effects of fatigue on the response of the trunk to a variety of conditions of initial static loads and step inputed loads causing rapid sagittal trunk flexion. This study was modeled after previous research (Krajcarski et al., 1999) except that the current study was repeated after a prolonged task resulting in acute fatigue of the trunk extensors. Therefore, two sets of data were collected, one during rested conditions and the other after the trunk extensor muscles had been fatigued. The variables measured in the study were the pre-trial EMG levels, peak EMG levels, peak extensor moments, peak trunk flexion angle, trial duration, time to peak moment and times to peak EMG level. The EMG data were collected from 10 surface EMG channels of the bilateral external oblique, internal oblique, lumbar erector spinae, thoracic erector spinae, and rectus abdominus muscles.

The within subject variability was generally low for the mechanical and EMG variables, indicating that the data collected from the ten trials was reliable for each loading condition.

A comparison between the findings of Krajcarski et al. (1999) and the rested values in the present study revealed similar effects of pre-load and added load on peak trunk flexion and extensor moments. The EMG activity of the trunk extensor muscles (TES and LES) had similar responses to the loading conditions, although the pre-trial LES and TES EMG amplitudes from Krajcarski et al. (1999) were about 6 to 13 %MVC larger than was found in the current study. Also, the peak muscle activity of the TES was about 10 %MVC lower and the IO and EO had larger EMG amplitudes during the pre-trial and trial in the Krajcarski et al. (1999) study. In fact, in the current study, the peak EMG activity rarely got above 10 %MVC. However, both studies have found
similar responses to the loading conditions, since the EO and IO remained relatively unchanged regardless of the pre-load magnitude. In addition, the two studies found that the RA EMG activity remained unchanged regardless of the loading condition.

This main objective of this study was to determine what, if any, were the effects of fatigue on sudden loading of the spine. The findings provide evidence that there were some changes in the response of the spine to sudden loading during fatigue conditions. However, the only main effect of fatigue was for the TES response to the different loading conditions. There was also a significant interaction between fatigue level and pre-load for the maximum trunk flexion variable, with the fatigued 4% pre-load trials having greater values than the rested 4% pre-load trials. The expected increase in LES and trunk flexor (co-contraction) EMG activity during fatigue was not observed.

Under the current protocol, fatigue seemed to have few global effects on the subject pool. The fatigue related increase in peak flexion angle was representative of a decrease in spine stability. In fact, the effect of fatigue on peak flexion angle was greater for the lower pre-load. This finding is a result of fatigue decreasing the level of spine stiffness associated with the 4% pre-load. The faster degradation in MnPF for the LES, compared to the TES, may be responsible for the increase in TES EMG amplitudes during fatigue. To maintain the same level of spine stability in the rested condition, the TES may have an increased role in maintaining joint moment and stability when the trunk extensors are fatigued.

A review of the individual subject data revealed that two subjects did demonstrate increases in agonist and antagonist muscle activity for the fatigued loading conditions, when compared to the rested conditions, and one subject was observed to have increased peak flexion values as a result of fatigue. Overall, the data seem to indicate that fatigue affects individuals in different ways, with respect to spine mechanics. It may be hypothesized that some subjects were more susceptible to the
effects of fatigue. This may help explain why some, but not all, individuals are injured while manual material handling over prolonged periods.

**Hypotheses**

1. **Anticipation of Added Loads**

   It was hypothesized that the pre-trial EMG amplitudes trial would be the same for a given pre-load magnitude, regardless of the mass of the added load. If the subjects were not anticipating the load perturbation, the EMG amplitudes should not vary for a given pre-load mass. All of the muscles, except for the I.ES, showed no difference in the pre-trial EMG levels of the loading conditions with the same pre-load magnitude. The I.ES had a significantly greater pre-trial EMG activity during the 24% added load trials, but the increase was only 1.1 %MVC. This small difference in %MVC should not be considered functionally significant since it would have very little effect on the pre-trial stability of the spine. Therefore, the subjects did not appear to anticipate the addition of the added load since the pre-trial EMG amplitudes were not functionally significant for a pre-load magnitude regardless of the added load. The hypothesis was accepted.

2. **Effect of Fatigue of Pre-Trial EMG Activation**

   It was hypothesized that pre-trial EMG amplitudes would increase for the fatigued trials, when compared to the rested trials with the same static pre-load magnitudes. This assumption was based on studies that have found that fatigued muscles have a reduced capacity to generate force (Parnianpour et al., 1987). The present study found that none of the observed muscles had a significant increase in pre-trial EMG amplitudes greater than 0.3 %MVC for the fatigued conditions in comparison to the rested conditions. Therefore, it must be concluded that fatigue had no impact on pre-trial EMG amplitudes, and the hypothesis is rejected.
3. **Effect of Fatigue on Maximum Trunk angular displacement**

Prior to the study, it was hypothesized that the slower muscle contractile speed, longer reaction times to perturbations and weakening that are associated with fatigued muscles (Bigland-Ritchie & Woods, 1984; Parnianpour et al., 1987) would result in larger trunk displacements for the fatigued trials when compared to the rested trials. However, fatigue was found to have no effect on the amount of maximum angular displacements in response to various added load conditions. The only observed effect of fatigue on peak trunk flexion was the increased flexion for the fatigued 4% pre-load trials in comparison to the rested 4% pre-load trials. The hypothesis was rejected, although fatigue was found to result in an increase in peak flexion for the 4% pre-load conditions.

4. **Effect of Fatigue on Peak Extensor Moments**

It was hypothesized that the fatigue trials would have greater acceleration of the trunk in response to the added loads and an increased need for deceleration, which would result in increased peak extensor moments to arrest the forward motion. This was based on the finding that the reaction time to a sudden load is longer for a fatigued trunk than a rested trunk (Wilder et al., 1996). It was theorized that the increase response time to a sudden load would result in the need for greater trunk decelerations. This was not found to be the case in this study since there were no differences in peak moment values for the fatigued and rested trials. Therefore, the hypothesis was rejected.

5. **Effect of Fatigue on Peak Muscle Activity**

The fatigued trials were hypothesized to have greater peak levels of EMG amplitudes when compared to the rested conditions. Fatigue has been shown to require increased activation to maintain a constant force (Maton, 1981). Also, the antagonist muscle activity was expected to increase because previous studies have demonstrated an increase in abdominal co-contraction during fatigue (O'Brien & Potvin, 1996; Potvin & O'Brien, 1998). However, only the TES, showed increased peak muscle activity in
response to fatigue and there was no increase in co-contraction observed as a result of fatigue. Therefore, the hypothesis was rejected for all muscles except the TES.

6. Effect of Fatigue on Time to Peak Angle, Time to Peak Moment and Time to Peak EMG Activity

Prior to data collection, it was thought that fatigue would increase the time to peak for each of the measured variables because the reaction time to a sudden load is longer for a fatigued trunk (Wilder et al., 1996) and the force generating capacity decreases with fatigue (Parnianpour et al., 1987). However, the data for the time to peak for all the measured variables did not indicate any differences between the fatigued and rested conditions and the hypothesis was rejected.

7. Effect of Pre-Load on Maximum Trunk angular displacement

It was hypothesized that the maximum trunk angular displacement would be greater for the 4% pre-load conditions than for the 16% pre-load conditions because the trials with the lower pre-load would have lower spine musculature stiffness and decreased spine stability (Cholewicki & McGill, 1995). The maximum trunk angular displacement data reveals that the trials with the 4% pre-load had larger maximum angular displacements than the 16% pre-load. Therefore, the hypothesis was accepted.

8. Effect of Added Load on Maximum Trunk angular displacement

It was hypothesized that the trials with the 24% added load would have larger peak trunk displacement than the trials with the 12% added load. The larger added load magnitude should have resulted in larger inertia and initial trunk flexion accelerations that would ultimately result in larger trunk flexion angles before the rotation could be arrested. The 24% added load trials did, indeed, have greater maximum angular displacement than the 12% added load trials and this hypothesis was accepted.
9. **Comparison of the [4+24] and [16+12] Loading Conditions**

It was hypothesized that the [4+24] condition would result in greater trunk angular displacement, peak moments, peak muscle activation and trial duration than the [16+12] condition, even though the final total load was equal in both conditions. According to Cholewicki and McGill (1995), the lower (4%) pre-load condition should result in lower trunk extensor muscle stiffness and spine compression causing the trunk to be less stable when compared to the 16% pre-load condition. The decrease in pre-trial stability should result in the larger responses to added loads. The results from this study indicate that there were greater peak trunk flexion, peak moments, and peak muscle activation for the LES, TES, EO, and IO with the 4% pre-load. However, the peak EMG amplitude for the RA and the trial duration showed no differences between the two loading conditions. Therefore, the hypothesis accepted for all variables but the peak RA and trial duration.

**Limitations and Assumptions**

The limitations and assumptions of the experimental protocol and analysis should be addressed when attempting to understand the results gathered from this study. First, the length of the fatiguing session was determined using a psychophysical rating criteria and the tester's visual observation. Subjects were encouraged to maintain the static contraction until the subject reached a fatigue rating of 9 out of a maximum of 10 (exhaustion). This rating of 9 was intended to be the point just prior to a fatigue level where the subject would not be able to maintain body position while resisting the forces caused by the static load. If the subject did not appear to be fatigued (no muscular tremor or visual signs of discomfort) but gave a rating of 9, the tester would extend the fatiguing session until characteristics associated with fatigue and discomfort were observed. The duration of the fatiguing session was based on both the subject's ability to rate their level of fatigue and the tester's ability to judge the subject's level of fatigue.
through qualitative visual observation. This may have resulted in the varying levels of
decrease observed in the MnPF of the trunk extensor muscles.

Secondly, the EMG data that were collected during the dynamic motion were not
corrected for changes in length and velocity. The trunk movement during the trials
resulted in changes in muscle length and contraction velocity that serve to modulate
force (Winter, 1990). For this reason, the interpretations of the EMG data from the trial
phases were based on activation and not force. However, it was assumed that the
corrections would not lead to changes in the final interpretation of the data.

Thirdly, this study used a back harness to apply the force causing the forward
flexion moment. Several other studies have used this method to apply the forward
flexion moments (Cresswell et al., 1994; Wilder et al., 1996; Thomas et al., 1998;
Krajcarski et al., 1999). Another option to produce forward flexion moments is to use a
hand-held load that is positioned away from the body (Lavender et al., 1989; Lavender et
al., 1993; Mannion et al., 2000). The difference between the two methods is that the
hand-held loads will primarily place compressive forces on the spine whereas the back
harness method will place external shear forces on the spine. It is hypothesized that the
neuromuscular response to the shear forces would results in an increase in EMG activity
levels, especially in those muscles (LES) responsible for counteracting the effects of
shear forces on the spine. However, the use of the back harness method allowed for the
application of larger flexion moments to the spine than would be generated with a hand-
held loading system because the moment arm of the load is generally larger for the back
harness method and the strength capacity of the handgrip was not a possible limiting
factor. Also, the protocol of this study dictated that the subject was fatigued in the trunk
extensor muscles at the completion of the fatiguing sessions and throughout the fatigued
loading trials. If the study was to be completed using a hand-held loading method, it is
possible that the length of the holding time and the mass of the load would cause
muscles in the upper limbs to fatigue before the trunk extensor muscles were fatigued. Also, the study used the variable of trunk flexion angle to approximate spine stability. The compressive pre-loads associated with hand-held loads would increase lumbar stiffness and increase spine stability (Edwards et al., 1987; Janevic et al., 1991) such that the hand-held loading conditions would reduce the level of muscle activation that would be needed to maintain stability during the trials, and reduce the responses of the EMG signals from the muscles that were monitored.

Only muscles that were accessible with surface electrodes were considered for analysis. It has been noted that the quadratus lumborum, an intrinsic muscle, assists the erector spinae muscle groups in extension and is an important stabilizer of the lumbar region of the spine (McGill et al., 1996). However, due to the technology available to the experimenter, intrinsic muscles were not included as muscles that were analyzed. The lack of significant difference in the monitored muscles, in response to fatigue, could possibly be due to unmonitored changes in the more intrinsic muscles.

Finally, the lack of added load effects on the trial duration data could be due to the magnitudes of added load values used in this study. Perhaps an added load effect would be found if the range of added loads were increased to have an absolute difference that was greater than 12 % of IEMmax. However, concerns for the safety of the subjects, especially under fatigued conditions, prohibited the use of larger loads.

**Main Findings**

The findings presented in this study are in general agreement with the findings from past studies regarding the rested trunk's response to sudden loading. However, the results concerning the effects of fatigue were not what were expected when compared with previous studies regarding the behaviour of fatigued agonist muscles and the counterpart antagonist muscles.
The findings of the study will be presented with the fatigued and rested loading conditions pooled together due to the small number of measured variables having been significantly affected by the fatigued conditions. The variables that showed a fatigue effect will be discussed at the end of this section. Additionally, the bilateral data from the muscles were also pooled. A comparison of the left and right sides of each muscle revealed no differences for each of the measured variables, as would be expected considering the trunk movement was within the sagittal plane.

This study was conducted in an attempt to expand the knowledge of what is responsible for the maintenance of spine stability. The concept of stability can be explained with a mechanical and a clinical definition. However, the stability of the spine cannot be simplified using the mechanical definition that implies that the spine is either stable or unstable. The spine has the capacity to adapt its responses to the different conditions that may cause instability. This concept is in accordance with the definition of clinical stability as the spine’s capacity to limit potentially injury-causing displacements during normal physiological loading (White and Panjabi 1978).

Similar to Chiang (1997) and Krajcarski et al. (1999), trunk angular displacement was used as an analog of spinal stability. The 4% pre-load conditions had larger trunk angular displacements than the 16% pre-load conditions. This supports the concept that the spine was less stiff and less stable during the lower pre-load condition (Rack & Westbury, 1974; Bergmark, 1989, Cholewicki & McGill, 1995).

It is interesting to note that the magnitude of trunk angular displacements were more dependent on the mass of the added load than the mass of the pre-load. The difference in trunk angular displacement is greater when comparing loading conditions with the same pre-load as opposed to the comparison of the loading conditions with the identical added load. However, the amount of trunk flexion decreases as the mass of the pre-load increases for a given added load. This magnifies the importance of
avoiding high sudden load magnitudes and low pre-loads when concerned about maintaining spine stability.

The [4+24] and [16+12] loading conditions both had a final load of 28 %EMmax. The [16+12] condition had a peak trunk angle that was 49% smaller than the [4+24] peak trunk angle, supporting the notion that higher pre-load and pre-activation results in higher trunk stiffness and spine stability. The maximum extensor moment and maximum EO, IO, LES and TES EMG amplitudes were observed to be higher for the [4+24] condition than the [16+12] condition. This may indicate that the [4+24] condition has lower spine stiffness before the trial and that the higher added load mass results in higher trunk accelerations. Therefore, the trunk musculature in the [4+24] condition must generate larger EMG activity to stop trunk movement and arrest forward flexion when compared to the [16+12] condition. Also, the spine acts like an underdamped system and, in response to the added load, the trunk musculature will overcompensate in an attempt to regain stability (Lavender et al., 1993). Since the [4+24] condition has a larger added load than the [16+12] condition, this overcompensation will result in larger peak moments.

According to Parnianpour et al. (1988), fatigue of the trunk musculature affects the coordination and control of the neuromuscular system. The decrease in muscle coordination, as a result of fatigue, would lead to a decrease in spine stability (Parnianpour et al., 1988). The neuromuscular system's ability to maintain spine stiffness should decrease due to the diminishing force generating capacity of a fatigued muscle. Theoretically, the instability resulting from fatigue should impact the peak trunk angle since that is the variable that is used to indicate the destabilizing effect of the perturbation. However, there was no functionally significant change detected in peak trunk angle when comparing rested and fatigued conditions. The only functionally relevant change in EMG data that was attributed to the effects of fatigue was the
increase in peak EMG activity of the TES during the fatigued conditions when compared to the rested trials. A possible explanation for the change in TES peak EMG activity could be an increased dependence on the TES to protect against instability during fatigue conditions. It was observed during the fatiguing session that the LES musculature fatigued more rapidly than the TES. Therefore, the TES may have had a greater remaining force generating capacity to protect the spine when the LES were fatigued. Also, the TES may take a more active role in the maintenance of spine stability during fatigue because it has a greater maximum moment potential than the LES due to its larger cross-sectional area (Cholewicki & McGill, 1996). Crisco and Panjabi (1991) determined the larger and more superficial muscles spanning a number of intervertebral joints were more efficient at stabilizing the spine. The TES is more superficial and larger than the LES (Cholewicki & McGill, 1996), which would lead one to believe that it is a more efficient spine stabilizer (Figure 5).

**Additional Findings**

The basis of the current study was a previous study by Krajcarski et al. (1999), evaluating the response of the spine to sudden loading causing rapid flexion. In addition to the experimental protocol that was used by Krajcarski et al. (1999), the effect of fatigue was introduced into the protocol and dependent variables, such as trial duration and time to peak data for all measured variables, were added.

A comparison with the results of Krajcarski et al. (1999) revealed similar effects of pre-load and added load on peak trunk flexion and extensor moments under rested conditions. The magnitudes of the extensor moments were higher in Krajcarski et al. (1999) because of larger maximum isometric extensor moments for that study. Krajcarski et al. (1999) found the average IEMmax was 305.7 ± 58.8 Nm, and the present study found the average IEMmax was 267.5 ± 42.1 Nm, an average of 12%
lower. The EMG activity of the trunk extensor muscles (TES and LES) had similar responses to the loading conditions, although the pre-trial LES and TES EMG amplitudes from Krajcarski et al. (1999) were higher than those observed in the present study. The pre-trial LES EMG amplitudes were about 13 %MVC larger for each loading condition Krajcarski et al. (1999). Krajcarski et al. (1999) had pre-trial TES EMG amplitudes that were approximately 5 %MVC greater, and peak TES EMG amplitudes that were at least 8 %MVC lower, for each loading condition. In the Krajcarski et al. (1999) study, the IO and EO had larger EMG amplitudes during the pre-trial and trial. In fact, in the current study the peak EMG activity rarely got above 10 %MVC. The pre-trial IO EMG amplitudes were at least 10 %MVC greater in Krajcarski et al. (1999).

However, both studies have found similar responses to the loading conditions, since the EO and IO remained relatively unchanged due to the pre-load magnitude. In addition, the two studies found that the RA EMG activity remained unchanged regardless of the loading condition. However, the level of Krajcarski el al.’s RA EMG amplitudes were 7 %MVC higher for each loading condition.

With regards to the time to peak (TTPK) findings, the higher pre-load resulted in greater TTPK values than the lower pre-load for the extensor moment and the five muscles of the trunk. Interestingly, the trials with the larger pre-load had increased initial stiffness based on the smaller maximum trunk angle displacement when compared with the lower pre-load trials. It was thought that the trials with greater initial spine stiffness should have the shorter times to peak muscle activation because of the increased spinal compression and the increased tension within the muscles. It would be expected that the trials with the greater TTPKs would be those with the 4% pre-load. However, a possible explanation for the current results may be that the trunk acceleration caused by an added load is smaller for the 16% pre-load trials because of the larger initial spine stiffness. The greater acceleration of the trunk would pose an increased risk to the spine
such that the need to regain stability would become much more of a priority. The increased accelerations may cause the proprioceptive organs within the muscles and tendons to react more vigorously, resulting in a stronger stretch reflex response and shorter TTPKs for the muscles during the 4% pre-load trials.

**Case Studies**

The amount that fatigue affected the subjects varied throughout the entire list of measured variables. However, subject #1 and subject #6 did have results that seem to give a strong indication that their spine mechanics were affected by muscular fatigue.

Upon a review of the literature on the effects of fatigue on the trunk musculature, one would expect to find increased agonist and antagonist muscle activity (Parnianpour et al., 1988; Potvin & O'Brien, 1998). While this was not the case in the grouped averages, subjects #1 and #6 did demonstrate consistent increases in peak EMG activity in response to fatigue. It was theorized prior to the study that fatigue would elicit higher erector spinae peak EMG activity due to the weakening of the fatigued muscle. Therefore, it was expected that the fatigued response to an added load would be a higher percentage of maximum activation for the LES and TES. Also, it was thought that the antagonist abdominal muscles would play a more prominent role in maintaining spine stability during fatigue conditions.

Subject #6 was the only subject that showed a consistent increase in peak trunk angle in response to fatigue. It is possible that this subjects had an increase in peak trunk flexion due to fatigue since the coordination, strength and motor control of the muscle would have been impaired (Parnianpour et al., 1988; Wilder et al., 1996).

**Implications of the Research Findings**

The grouped data that were collected did not correlate well with the hypothesized results. It was thought that fatigue would cause global changes for the measured
dependant variables. However, only the TES peak EMG amplitudes showed consistent increases due to fatigue. The amount of spine stability did not seem to have been affected by fatigue except for the increased peak flexion angle during the fatigued 4% pre-load conditions compared to the rested 4% pre-load conditions.

However, the analysis of the data for each individual subject reveals that fatigue had a greater affect on some subjects and less for others. Two subjects showed consistent increases in peak EMG amplitudes during fatigue, and one of the two had consistently larger peak flexion angles as a result of fatigue.

From the findings of this study, it might be concluded that perhaps the effects of fatigue on spine loading should not be evaluated based on an average group response. It seems possible that certain individuals have physiological and anatomical characteristics that make them predisposed to injury when fatigued. Therefore, the general concensus that fatigue increases the chance of injury is still plausible. But, not everybody will have an increased risk to injury, since some individuals display no changes in response to fatigue. Future research should focus on the identification of factors that identify whether an individual will be adversely affected by fatigue.
Chapter VI

Conclusions

The purpose of this thesis was to investigate the effects of fatigue on the response of the trunk to loading conditions causing rapid flexion. The study used trunk flexion as a measurement to represent spine stability. The other mechanical variables that were measured included: extensor moment, trial duration, time to peak moment, and the time to peak EMG amplitude for each of the muscles. In addition, to evaluate the muscular response to the loading conditions during rested and fatigued conditions, ten trunk muscles were monitored for EMG activity.

The rested results gathered from this study support the findings of Krajcarski et al. (1999). Both studies showed similar hierarchies of response to the different loading conditions for the measured variables.

The grouped findings, across subjects, indicated some average changes, as a result of fatigue, in the response of the trunk musculature to the different loading conditions. The TES showed increased activity during fatigued trials relative to the rested trials. The trunk also demonstrated higher trunk rotations (lower levels of spine stability) during the fatigued trials with the 4% pre-load when compared to the rested 4% pre-load conditions.

The data were also analyzed for each subject, and it was revealed that two subjects responded to the fatigue conditions by increasing the activation levels of the antagonist and agonist muscles in response to the sudden loading. One of the subjects demonstrated greater peak trunk flexion for each loading condition during the fatigued conditions when compared to the rested trials.
Recommendations for Future Research

To increase our understanding of the effects of fatigue on the sudden loading of the spine several topics should be addressed:

First, the range of the pre-loads and added loads would be increased so that the effects of load condition of spine stability can be magnified. Perhaps the difference in magnitude between the pre-loads was not sufficient, as indicated by negligible effect pre-load magnitudes had on the trunk flexors.

Secondly, the use of indwelling electrodes would have eliminated many of the limitations associated with surface electrodes and allowed for the investigation of intrinsic muscles such as the quadratus lumborum, which act in trunk extension and help stabilize the lumbar region of the spine (McGill et al., 1996).

Lastly, more research should be conducted to determine if certain individuals have physical characteristics that cause them to have a higher chance of injury due to fatigue. And if this is the case, research identifying those risk factors or characteristics should be conducted to help screen those individuals predisposed to fatigue related injury and keep them away from certain jobs with prolonged and/or repetitive loading.
References


Appendix A

Information and Consent Form
Information and Consent Form

Ergonomics and Biomechanics Lab
School of Human Kinetics
University of Windsor

Study Title: Response of the Spine to Sudden Loading in the Sagittal Plane during Fatigued Conditions

Conducted by: Ben Parcero (Master's student) and Dr. J. R. Potvin (supervisor)

I agree to participate in a study that is designed to add new knowledge concerning the response of the musculoskeletal system to sudden loads during fatigued conditions. I am eligible for participation in this study since I have never experienced an injury to my lower back. The investigator has explained the procedures and the necessary time commitment to me. Before the sudden loading trials begin, I will be asked to perform maximal isometric exertions of the back extensor and abdominal muscles. These maximal isometric exertions are strength tests that have commonly been used in isometric strength evaluations. The dynamic sudden loading 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 surface electrodes to bilaterally collect EMG from five muscle sites: lumbar erector spinae, thoracic, erector spinae, external oblique, internal oblique, and rectus abdominis. I understand that will be asked to maintain upright posture while resisting forward flexion moments. The testing session will consist of sudden loading trials while in a rested condition and while I am in a fatigued condition. For the trials that require that I am fatigued, I will be asked to maintain an upright posture while resisting a flexion moment that is 50% of my maximum extensor moment for two minutes prior to the start of the trials. I will be exposed to four loading conditions during the rested and fatigued trials. These conditions will be a combination of 4, 12, 16, and 24% of the maximum isometric extensor moment (IEM_{MAX}). Conditions consist of a preload (4 or 16% IEM_{MAX}) and an added load (12 or 24% IEM_{MAX}). Each loading condition is achieved by placing an added load on a preload (e.g. 4 + 12 = 16%). For each loading condition, there will be 12 sudden loading trials. I understand that with each sudden loading trial there is always a risk that I may experience discomfort or injury. I have been instructed to terminate the session 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.

Subject Name:                            Subject Signature:
Date:                                   Witness Signature:

86
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