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A biomechanical examination of calculating spinal compression: Towards a new polynomial approach.

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A BIOMECHANICAL EXAMINATION OF CALCULATING SPINAL COMPRESSION: TOWARDS A NEW POLYNOMIAL APPROACH

by

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ABSTRACT

A BIOMECHANICAL EXAMINATION OF CALCULATING SPINAL COMPRESSION: TOWARDS A NEW PLOYNOMIAL APPROACH

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The purpose of this study was to develop a regression equation that, incorporating the potential energy of the load in the hands, was capable of improved predictions of spinal compression forces. A stepwise polynomial equation was developed from EMG profiles of 15 muscles, and its spinal joint loading predictions at L4/L5 were compared to current methods of calculating spinal compression. Generally, muscle activation levels were shown to increase with increased loading height, implicating that the Central Nervous System responds to changes in spinal stability. The inclusion of potential energy into the calculation of spinal disc compression at L4/L5 improved estimates of the compressive forces acting on the spine. This is the first model to incorporate potential energy into a predictive model for lumbar spine compression without the use of electromyography. Thus, it is concluded that potential energy plays a vital role in dictating the recruitment patterns of the trunk.

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LIST OF ABBREVIATIONS

2-D: 2-Dimensional

3-D: 3-Dimensional

3DSSPP: Michigan 3-Dimensional Static Strength Prediction Program

Ag/AgCl: Silver/Silver Chloride

ALL: Anterior Longitudinal Ligament

ANOVA: Analysis of Variance

APs: Action Potentials

CMRR: Common Mode Rejection Ratio

CNS: Central Nervous System

CofM: Centre of Mass

CSA: Cross Sectional Area

DMH: Distributed Moment Histogram

DoF: Degrees of Freedom

EMG: Electromyography

EMGAO: EMG Assisted Optimization

EO: External Oblique

ES: Erector Spinae

IO: Internal Oblique

LBD: Low Back Disorder

LBP: Low Back Pain

LD: Latissimus Dorsi

LES: Lumbar Erector Spinae

LF: Ligamentum Flavum

LSM: Linked Segment Model

MULT: Multifidus

MUs: Motor Units

MVE: Maximum Voluntary Exertion

N: Newtons

NIOSH H: NIOSH Horizontal value

NIOSH V: NIOSH Horizontal value

NIOSH: National Institute for Occupational Safety and Health

Nm: Newton-metres

OPT: Optimization

PE: Potential Energy

PLL: Posterior Longitudinal Ligament

QL: Quadratus Lumborum

RA: Rectus Abdominus

RDELT: Right Deltoid (anterior portion)

RMSe: Root Mean Square Error

SEM: Single Equivalent Model

sEMG: surface EMG

SI: Stability Index

TA: Trunk Angle

TES: Thoracic Erector Spinae

WSIB: The Workplace Safety and Insurance Board

Chapter 1: INTRODUCTION

1.1 Background

Low back pain (LBP) is one of the most prevalent medical problems in western societies. Low back pain is multifactorial in origin and it is for this reason that it is not well understood. In Canada, back injuries account for over 25% of all lost time claims, the largest single claims category in most workers' compensation jurisdictions. Low-back pain, which is often seen initially in primary care practice, is estimated to be the most costly ailment in working-age adults. Disability resulting from LBP is the most common chronic health problem in adults under the age of 45 years and is second only to arthritis in those aged 45–65. Of the more than 90% of workers who return to work within 6 months of their injury, 20%–44% will experience recurrences resulting in further time off work and 15%–20% of patients will continue to experience back pain for at least 1 year from the initial onset (Canadian Task Force on Preventive Health Care, 2003)

Workplace research from Ontario found that 20.0% (18,665) of lost time claims involved the lower back, including the lumbar, sacral, coccygeal regions (WSIB, 2003). In Canada, the morbidity costs from long-term disability due to musculoskeletal diseases are staggering. In the year 1999, the cost associated with musculoskeletal diseases was $12.6 billion and, of that, the back and spine accounted for 37.3 % (Canada Department of National Health and Welfare, 1998).

The incidence of LBP in the workplace highlights the importance of an effort to reduce the work-related risks associated with the injury. Biomechanical factors, such as spinal compression and shear forces, along with the postural demands of a task, are often deemed as risk factors for LBP. The injury may start with mechanical failure of the tissues, or of the system as a whole, i.e. buckling of the spine. Spine stability is very important in injury mechanisms as transient buckling events are believed to result in injury and tissue damage.

In 1985, the National Institute for Occupational Safety and Health (NIOSH) assembled an informal committee of experts, who reviewed the literature on the recommended criteria for defining lifting capacity in order to develop an equation that could be used to evaluate lifting tasks (Waters, Putz-Anderson, Garg, & Fine, 1993). The now renowned NIOSH equation is widely used by ergonomists in an effort to reduce lifting injuries in the workplace.

It is well documented that postural muscles act in response to spinal loading in order to stabilize the spine. When the spine is stable under a given load, then small neuromuscular or vertebral movement errors are automatically corrected without tissue damage (Granata, Slota, & Wilson, 2004). These complex responses are often not well predicted by existing models. Cocontraction of agonist and antagonist muscles may be the most important factor in stabilizing the spine (Brown, Haumann, and Potvin, 2003). The literature suggests that antagonistic cocontraction can reduce the risk of low back injury by increasing spinal stability. In fact, the maximum load the spine could endure was found to increase by 36-64% as a result of antagonistic cocontraction (Granata and Marras, 2000). Therefore it is reasonable to deduce that, when investigating LBP risks, antagonistic cocontraction must be considered.

In an attempt to further understand the etiology of back pain, several researchers have developed biomechanical models of the spine in order to predict compression including; Morris, Lucas, and Bresler 1961; Bean, Chaffin, and Schultz, 1988; McGill, 1992; Raschke, Martin & Chaffin, 1996; Cholewicki and McGill, 1994; McGill, Norman, and Cholewicki, 1996; Stokes, and Gardner-Morse, 2001; and Brown and Potvin, 2005. These models can be used to replicate manual material handling tasks, and the results subsequently are used to guide the design of such tasks to reduce various low back stresses (Chaffin, 2005). In the most recent literature, electromyographic analyses of the trunk muscles have been incorporated into the model in order to estimate the force being generated by each muscle. These complex models are labour- and time-intensive, and the extensive measurements are not easily performed, therefore the existing biomechanical models used to predict compression are less than practical. Hence, it is feasible that a regression equation based on experimental data would be a useful ergonomic tool that is easy to use and widely applicable. It is undeniable that factors other than load moments may play a role in determining how muscles are utilized. Select postures may limit the ability of the neuromuscular system to maintain stability (Granata and Wilson, 2001). A muscle's contribution to lumbar spine stability depends on the combination of direction and load level of isometric exertions. This was indicated in Cholewicki and van Vliet (2002) by a significant 2nd level interaction between the effects of exertion direction, load level, and the musculature involved on the Stability Index (SI).

Methods that are currently available for calculating compression force include single equivalent models (Morris et al, 1961; Chaffin 1975), EMG-based models (McGill, 1992; Granata and Marras, 1993; Cholewicki et al, 1995; Cholewicki and McGill, 1994), optimization techniques (Bean et al, 1988; Raschke et al, 1996; Granata and Wilson, 2001; Stokes, and Gardner-Morse, 2001; Brown and Potvin, 2005), and polynomial equations (McGill, et al, 1996). All the aforementioned methods of modeling the spine based their estimates of muscle forces on moments, with no consideration of destabilizing potential energy (PE). There are only two models that have used stability as a criterion in their optimization scheme. Stokes & Gardner-Morse (2001) showed that maximizing stability actually decreased the correlation between the experimental and predicted compression values. The authors stated that maximum stability would require very high levels of activation, consequently increasing the compression of the spine. Brown and Potvin (2005) used a more realistically moderate level of stability, as a constraint in their optimization model, and found that the inclusion of stability improved the accuracy of predictions of trunk muscle coactivity and subsequent spine compressive loading. These results illustrate that it is feasible to consider a delicate interplay between compression and stability. Therefore the goal of the current study is to use stability in a simple equation to convert moments and load potential energy into predictions of compression force.

1.2 Statement of the Purpose

The purpose of this study is to develop a regression equation that accurately predicts the joint compression at the L4/L5 level through the inclusion of potential energy in the model. Utilizing electromyography, and biomechanical models, joint compression will be estimated for 1-handed static vertical loading tasks at 3 load-heights (10 cm, 40 cm, and 70 cm above L4/L5), 3 asymmetrical postures (0°, 45° and 90° from the left of midline), and at 2 reach distances (25 cm and 50 cm from the acromion), with one of 3 separate loads (1.36 kg, 2.72 kg, or 5.44 kg). Additionally, subjects will be asked to perform exertions that result in moments about all 3 orthogonal axes, to simulate muscle activation patterns that would occur during an asymmetrical loading task.

1.3 Hypotheses

1. It is predicted that the muscle activation of the trunk will show a positive relationship with the potential energy of the hand load, even while the moment demand remains relatively constant. As the hand load increases in vertical location, while the horizontal moment arm stays constant, the spinal system will increase in stiffness to protect from injury through increased activation levels. Thus, a high positive correlation will be found between the potential energy of the load and the myoelectric activity recorded in this study.
2. A regression model, which includes the potential energy of the system in the calculation of spinal compression, will result in lower errors than previous biomechanical models. Root-mean-square errors (RMSe) and correlations will be calculated between predicted forces from the models and the actual forces obtained through experimentation.

Chapter 2: LITERATURE REVIEW

2.1 Anatomy of the Trunk

The anatomical model in this study will be the 52 muscle fascicle model that was used by Brown and Potvin (2005). The muscles that are included in the model are those that cross the L4/L5 joint. In this way, the model accounts for only the muscles that contribute to the moments created about this joint. In order to accurately represent the lines of actions of the muscles, nodal points are included in the model. Furthermore, all spinal rotations will be assumed to occur about the L4/L5 disc, which will be considered to be fixed. The current section is a comprehensive anatomy review compiled from Gray (2000), Tortora (2005) and Behnke (2001).

2.1.1 Passive Tissues

2.1.1.1 Bone

The vertebral column is a flexible column, formed of a series of bones called vertebrae. There are thirty-three vertebrae, and they are grouped according to the regions they occupy; seven in the cervical region, twelve in the thoracic, five in the lumbar, five in the sacral, and four in the coccygeal (Gray 2000). The lumbar region is highlighted in Figure 1. The vertebrae of the sacral and coccygeal regions are termed false or fixed vertebrae, they are united with one another in the adult to form two bones—the sacrum, and the coccyx (Gray, 2000; Tortora 2005). The entire structure serves to support the upper body and protect the spinal cord from trauma.

Figure 1. The human spine, highlighting the lumbar region. (Behnke 2001)

A typical vertebra consists of two parts: an anterior segment, the body, and a posterior part, the vertebral or neural arch. These two portions enclose the vertebral foramen, which houses the spinal cord. The vertebral arch consists of a two short pedicles that unite with a pair of laminae, and is the origin of seven processes: four articular, two transverse, and one spinous. The transverse and spinous processes are anchors for muscles, and the four articular processes (2 superior, 2 inferior) form joints with adjacent vertebrae (Gray, 2000).

2.1.1.2 Intervertebral Joints

Intervertebral discs are pads of cartilage that distribute compressive loading evenly on the vertebral bodies. The bony surfaces of the spine are connected by broad flattened disks of fibrocartilage, and are surrounded by a joint capsule of dense irregular connective tissue. Intervertebral discs are composed of fibrous tissue and fibrocartilage, forming the annulus fibrosus (Gray, 2000). The central region of a disc is a soft, pulpy, highly elastic substance called the nucleus pulposus which is constrained by the tough concentric lamellae of the annulus fibrosus. The fluid nature of the nucleus ensures that compressive loading applied to a disc generates a tensile hoop stress in the annulus (Adams, and Dolan, 2005).

The joints between adjacent vertebral bodies are amphiarthroses, or slightly movable articulations. More specifically, they are symphysis joints. Discs are protected from flexion, extension and shear by the facet joints (Adams and Dolan, 2005). The nucleus pulposus are especially well-developed in the lumbar region as they serve primarily as shock absorbers. Under pressure, the highly elastic nucleus pulposus becomes flatter and broader and pushes the more resistant fibrous laminae outward in all directions (Gray, 2000). The center of rotation of the joint for sagittal bending is within the disc. When analyzing the mechanics of the lumbar spine it is often assumed that extensor musculature of the trunk has a lever arm distance between 5 cm (van Dieen, and de Looze 1999; Chaffin, 2005) and 7.5 cm (McGill and Norman, 1987; Dolan and Adams, 1993) posterior to the center of the vertebral body.

2.1.1.3 Ligaments

Ligaments are fibrous bands or sheets of connective tissue linking two or more bones, cartilages, or structures together, and are highly resistant to strain. The ligaments connecting the fifth lumbar vertebra with the sacrum are similar to those which join the movable segments of the vertebral column with each other (Gray 2000). Three of the primary ligaments in the spine include the ligamentum flavum (LF), anterior longitudinal ligament (ALL), and the posterior longitudinal ligament (PLL). The LF connects under the facet joints to create a small curtain over the posterior openings between the vertebrae. The ALL attaches to the front (anterior) of each vertebra. This ligament runs up and down the spine (vertical or longitudinal). The PLL runs up and down behind (posterior) the spine and inside the spinal canal (Tortora, 2005). The supraspinous ligament connects the apices of the spinous processes from the seventh cervical vertebra to the sacrum, while the interspinous ligaments connect adjoining spinous processes. Furthermore, an additional ligament connects the pelvis with the vertebral column on either side called the iliolumbar (Gray, 2000). Ligaments and other passive tissues of the trunk aid in the stabilization of the spine during tasks that require low levels of muscular activation (Cholewicki & McGill, 1996).

2.1.2 Muscles of the Trunk

The movements permitted in the vertebral column are: flexion, extension, lateral flexion (abduction), reduction (adduction), bi-directional rotation, and circumduction. Flexion is best described as forward bending of the joint in the sagittal plane, resulting in a decrease in the angle of the trunk relative to the pelvis. The rectus abdominis, obliques, and the psoas major are the prime movers in this action (Tortora, 2005). Extension is the opposite of flexion, or the return to the anatomical position from trunk flexion. In the lumbar region, extension is produced by the intrinsic muscles of the back, including the sacrospinalis (erector spinae), and quadratus lumborum and partially by the lower trapezius. The multifidus aids in the extension of the cervical region (Behnke, 2001).

Lateral flexion of the spine is the lateral movement along the transverse plane away from the midline of the body, decreasing the angle between the thorax and the pelvis. Reduction, also referred to as adduction, is the return to the anatomical position from lateral movement. Rotation is the rotary movement around the longitudinal axis of the spine, which rotates the thorax to one side by the contraction of the multifidus, and the abdominal muscles. Circumduction is very limited, and is merely a by-product of preceding movements (Gray 2000) as it is the result of the coupling of several smaller movements within the spinal column.

2.1.2.1 Abdominal Musculature

2.1.2.1.1 Rectus Abdominis

The rectus abdominis (RA), is a long flat muscle, which extends along the entire anterior portion of the abdomen, and is sectioned in two by the linea alba. It originates from two tendons; the larger originates on the pubic crest and the smaller tendon originates from the pubic symphysis. The muscle is inserted into the cartilages of the fifth, sixth, and seventh ribs. In terms of spinal articulation, the RA acts to flex the vertebral column.

2.1.2.1.2 External Oblique

The external oblique (EO), which is situated on the lateral and anterior parts of the abdomen, is the largest and the most superficial of the three flat muscles in this region. It arises from the external surfaces and inferior borders of the lower eight ribs, and is inserted into the anterior half of the outer lip of the iliac crest and the linea alba. The external obliques act in lateral bending, flexion, and axial rotation of the vertebral column. The external oblique aids in raising intra-abdominal pressure and, acting with the contralateral IO, abducts and rotates trunk (Gray, 2000).

2.1.2.1.3 Internal Oblique

The internal oblique (IO) is thinner than the external oblique, under which it lies. It arises from inguinal ligament, from the iliac crest, and from the lumbodorsal fascia.

From this origin, the fibers diverge and insert onto either the crest of the pubis, onto the linea alba, or onto the cartilages of the seventh, eighth, and ninth ribs (Gray, 2000). The internal oblique also aids in raising intra-abdominal pressure, and acts to abduct and rotate the trunk in conjunction with the external oblique muscles on the contralateral side (Gray, 2000).

2.1.2.2 Posterior Musculature

2.1.2.2.1 Erector Spinae (sacrospinalis)

The sacrospinalis, or erector spinae (ES), is the largest muscle mass of the back. It arises from the anterior surface of a broad and thick tendon, which is attached to sacrum, the spinous processes of the lumbar and the eleventh and twelfth thoracic vertebrae, the supraspinal ligament, the iliac crests, and the lateral crests of the sacrum. In the lumbar region, it is large and is subdivided into the three columns as it moves toward the head. These gradually diminish in size as they ascend to be inserted onto the vertebrae and ribs. It splits, in the upper lumbar region into three columns – from medial to lateral they are spinalis group, the longissimus group, and the iliocostalis group. The ES aids in lateral bending and also acts to extend the spine, with the iliocostalis thoracis acting specifically to maintain the erect position of the spine.

2.1.2.2.2 Multifidus

These muscles are located deep to the erector group. It lies against the spine and originates from the transverse process of one vertebra, and inserts on the vertebra above or often onto, the second or third vertebra above. The multifidus (MULT), originates from the sacrum, the ilium, the transverse processes of the lumbar, thoracis, and lower four cervical vertebrae, and insert onto the whole length of the spinous process of one of the vertebrae above (Gray, 2000). The multifidus is a strong stabilizing muscle of the spine that runs from the sacrum upwards supporting the lumbar, thoracic and cervical spine. The multifidus extends the vertebral column and rotates it to the opposite side (Behnke, 2001).

2.1.2.2.3 Latissimus Dorsi

The latissimus dorsi (LD) is a triangular, flat muscle, which covers the lumbar region and the lower half of the thoracic region, and is gradually narrowed until its insertion into the humerus. It arises from the spinous processes and supraspinous ligaments of all lower thoracic, lumbar and sacral vertebrae, lumbar fascia, iliac crest, the last four ribs, and the and inferior angle of the scapula. From the various origins the fibres converge to form a quadrilateral tendon that is inserted onto the intertubercular groove of the humerus (Gray, 2000). The latissimus dorsi extends and adducts the arm, as well as rotates the arm medially, and draws the arm inferiorly and posteriorly.

2.1.2.2.4 Right Deltoid

The right deltoid (RDELT) is a large, thick triangular muscle which covers the shoulder. The anterior fibres pass obliquely backward and laterally from the anterior border and upper surface of the lateral third of the clavicle to insert onto the deltoid tuberosity. During arm abduction, the lateral deltoid is contracting, while the anterior deltoid is involved in stabilizing the joint to prevent sideways motion (Gray 2000, Kinakin 2004)

2.2 EMG-Force Relationship

Muscle force is exerted by a muscle fibre following an action potential, and is provided by both the active (contractile element) and passive components (parallel and series elastic components) of the muscle. Myoelectric signals can be recorded in biomechanical studies to predict muscle tension. Measurement of this type is called electromyography, or EMG. An electromyogram is the profile of the electrical signal detected by an electrode on a muscle. The summation of several muscle fibre action potentials, or complex motor unit potential, can be recorded by placing an electrode on the skin (i.e., a surface electrode). Surface electrodes record the sum of all of the action potentials (APs) from the active motor units (MUs). Surface EMG (sEMG) can provide information related to which muscles are active, when muscles initiate and cease activity, as well as the magnitude of the electrical response of the muscles during a task.

An increase in muscle tension can arise either by an increase in the firing rate of a given motor unit, or by recruiting additional motor units (De Luca, 1993). It is well established that muscle stiffness is proportional to muscle force (Bergmark, 1989) and an increase in myoelectric activity is associated with an increase in tension (Chaffin, 1999). However, the relationship may be linear or nonlinear (Solomonow, Baratta, Shoji, and D'Ambrosia, 1990). The force producing capacity of a muscle is dependent on the size of the muscle, the muscle's length, and the muscle's velocity. Additionally, when a muscle is stretched further than its normal resting length, an additional passive force is produced. All of these factors must all be taken into account in modeling the relationship between the amplitude of an electromyographic signal and the force produced by a muscle (Adams, Bogduk, Burton, and Dolan, 2005). Models that predict muscle forces (McGill, 1992; Granata and Marras, 1993; Cholewicki & McGill, 1995) often take the general form:

$$F_m = [G \cdot NEMG \cdot f(\text{length}) \cdot f(\text{vel})] + F_p \qquad [1]$$

where F_m is the predicted force of a particular muscle: G is a gain parameter used to represent the force per unit of cross sectional area, NEMG is the normalized EMG amplitude, f(length) is a scaling function based on the length of the muscle during a particular exertion, f(vel) is a scaling function based on the speed of shortening effect and F_p is the force due to passive tension when a muscle exceeds its resting length (Chaffin, Stump, Nussbaum& Baker, 1999).

2.3 Biomechanical Models of the Spine

The etiology of LBP is not known in most patients (Panjabi, 2003). Scientific evidence implies biomechanical stability of the lumbar spine is a vital aspect when considering the risk of low back disorders (LBD) (Granata, Orishimo, and Sanford, 2001). If the spine is unstable, small postural errors can be amplified by the biomechanical forces or external moments, causing sudden undesired vertebral motion (Granata, Slota, & Bennett, 2004). Knowledge of muscle recruitment patterns, activation levels, spinal shear forces, and compression may provide insight on the low back pain issue.

Models are created to reduce the complexity and enhance the understanding of a system. Modeling is a process by which biomechanists attempt to represent reality, and when combined with scientific data, modeling becomes a powerful scientific tool (Nigg and Herzog, 1999). Biomechanical models of the spine are a safe and inexpensive way to explore complex systems and their interactions. Models for force estimation range from simple, two-dimensional ones for static tasks involving no twisting or lateral bending, to complex ones that are applicable to dynamic tasks in three-dimensional (3-D) space (Tracy, 1990). Biomechanical models that predict either the time histories of muscle forces or the forces which act on biological tissues have become the foundation in understanding not only human motion, but the mechanisms of injury as well (Cholewicki & McGill, 1995).

A limitation of biomechanical models is that their predictions cannot be directly validated in humans (Cholewicki & McGill, 1995; Potvin 1997). Additionally, biomechanical models that simplify the anatomical details of muscles and ligaments have to contend with the fact that there are a vast number of structures that can generate or resist moments about a joint (Cholewicki & McGill, 1995).

There is a considerable amount of attention in the literature on which muscle forces are used to perform a range of physical tasks (Bean et al, 1988). Biomechanical models of the torso have been developed to predict tissue loading in the spine in an attempt to understand potential injury mechanisms to the low back from everyday activities (Jorgenson, Marras, and Gupta, 2003). When an external moment must be balanced across a joint, a biomechanical model of the joint and its muscles can be used to estimate the muscle forces necessary to equilibrate that external moment (Bean et al, 1988).

Dynamic or static intervertebral joint moments can be calculated using a linked-segment modeling (LSM) approach. The LSM approach commonly requires the input of subject mass, anthropometrics, load mass and either joint coordinates or segment angles.

The internal force required to balance the external lumbar moment is then calculated based on assumptions regarding the orientation and recruitment of the extensor trunk muscles (Potvin, 1997). To date, biomechanical models to determine spinal loading include single equivalent models, EMG-based models, optimization-based models, and regression-based models.

2.3.1 Single Equivalent Models

A Single Equivalent Model (SEM) assumes that the forces are acting in one plane, and that there is a single reaction moment. This method only requires the knowledge of the hand held load and joint kinematics. The complex musculature of the system can be reduced to a "single equivalent muscle" with a set moment arm for each axis. The spine is likened to a series of rigid segments which are linked by joints and moved by the actions of muscles (Adams et al, 2005). The net moments are calculated from anthropometric data of the masses and location of the centre of mass for body segments. All of the forces acting on a given part of the spine can be summated to form a single resultant force. The resultant is then broken into components, where the component which acts perpendicular to the mid-plane of the disc is defined as the compressive force acting on the spine (Adams & Dolan, 2005). The resultant force may cause the spine to bend or twist about its centre of rotation. The torque responsible for the bending and twisting can be divided into the sagittal plane (flexion) and frontal plane (lateral flexion) and the component which causes the spine to twist about its long axis.

For a static mechanical system, the conditions of equilibrium must be fulfilled at all times. In other words, the sum of all forces and the sum of all moments acting on the system must be zero. In the three-dimensional case, 6 equations of equilibrium are available. If the equations of equilibrium are sufficient to determine the unknown forces, the system is statically determinate (Bergmark, 1989). To simplify analysis, muscles with equivalent functions are grouped together as functional units. From the mechanical perspective, any meaningful division of the load between the functional muscle units must be based on a system that satisfies all conditions of equilibrium as well as the conditions of mechanical stability.

Morris, Lucas, and Bresler (1961) created a simple static model of the lumbar spine. The model assumes two internal forces act to resist the external moment: the extensor erector spinae muscles and abdominal pressure. Upon application of the model, it appeared as though compression of the intervertebral discs was occurring when loads were lifted (Chaffin, Anderson, & Martin, 2006). It has now been widely accepted that compression occurs in the discs.

Chaffin (1975) refined the model of Morris et al. (1961) by adding a hip-sacral link, and a lumbar-thoracic link. Based on the angles and geometry of these links, Chaffin's model produced estimates of compression and the force created by the erector spinae at the L5/S1 level, estimates of abdominal muscle force, as well as an estimation of the moment at the L5/S1 level.

Although a good evaluation of the total joint compressive forces can be obtained using an SEM, the force distribution between different muscles and ligaments cannot be solved. Thus, models with increasing complexity have been developed. In cases where asymmetrical loading or twisting is present, a more sophisticated model is required. While simpler models may give a similar solution for a given loading scenario, a more complex model allows insight into the true nature of muscle recruitment patterns and hence more realism in the model predictions.

2.3.2 Optimization Models

In order to properly understand the mechanisms of injury the forces predicted by a model including active muscles, forces on facet joints, discs, ligaments and other tissues, must be as close to physiological reality as possible (Cholewicki & McGill, 1995). The number of muscles that cross the lumbar spine is greater than 180, and this is much greater than the number of degrees of freedom (6 per vertebra) in a static analysis of the lumbar spine. In the optimizing control system, no muscle is acting on its own, nor can one muscle be identified as the most important for the stability of the lumbar spine (Cholewicki & VanVliet, 2002). Therefore, there are a redundant number of possible muscle activation patterns that could be used to satisfy equilibrium (Stokes & Gardner-Morse, 2001).

The majority of existing biomechanical models have more unknown muscle forces than equations of equilibrium and the problem becomes statically indeterminate, where the equations of equilibrium are not sufficient to determine the unknown forces. For each moment demand, there is an infinite combination of muscle activation patterns that could satisfy the mechanical requirements. Therefore, additional rules and assumptions must be introduced to determine the unknown forces. The optimization approach is often used to look at this problem, and usually provides solutions that correlate with actual human performance. Although 3-D-static models have been developed they contain major simplifications concerning the way muscles provide twisting (axial rotation) moments (Tracy, 1990). This model required mathematical optimization techniques, and is what is now called an 'optimization model'.

Optimization models use mathematical techniques to determine how to distribute the forces amongst involved muscles. This technique attempts to best represent the control of the central nervous system (CNS) over muscle activation patterns, assuming the activation pattern optimizes some physiologic objective (Brown & Potvin, 2005). In order to solve the statically indeterminate system, the optimization approach presents many different possible solutions. The most accurate solution is "decided upon" by making certain assumptions about physiological systems. Common objectives often involve the minimization of some mechanical cost function such as: maximum muscle stress shear force, compression or displacement of the spine. In other words, optimization models allow muscle force estimations and moment equations to be satisfied while simultaneously optimizing some objective function.

Linear programming techniques have also been used for optimization. Using this method, the investigator can encompass many variables through the use of computers, and it also provides good insight into the nature of the solutions. With linear programming, the objective function can be minimized, subject to the constraints based on the known parameters. The theory of linear programming allows evaluation of the trade-offs between the objectives (e.g. low contraction intensity and low joint compression) (Bean et al, 1988). However, one must be aware that with linear programming, the objective function must be linear in the unknowns. For example, compression must increase linearly with muscle activation (Bean et al, 1988). Basic computational steps to estimate compression force in the spine are outlined in the models developed by several authors including models developed by Schultz and Andersson (1981), Bean et al, (1988), Raschke et al, (1996), Stokes and Garner-Morse (2001), and Brown and Potvin (2005).

Bean et al, (1988) utilized a double linear optimization method to satisfy the 6 equations of equilibrium (3 force and 3 moment equations). The model requires that the sum of all the muscle forces equilibrate the external moment. To satisfy the moment constraints the objective functions are two-fold; the first linear program finds the lowest possible maximum intensity at which feasible solutions are found and the second linear program chooses among the solutions to minimize spinal compression. Muscles were divided into two groups; those working to counter the external moment (agonists); and those adding to the external moment (antagonists). The antagonist muscle forces were set to zero and, in doing so, negate any potential for the model to predict muscle co-contraction. The muscle forces are calculated by this optimization technique, where moment arm lengths, line-of action orientations, muscle cross-sectional areas, and external moments are the known parameters, and the joint reaction forces and muscle forces are the unknown variables. In its time, this model was advantageous over other techniques due to its low computation cost, stable solutions and the ability to evaluate the trade off between spinal compression and muscle intensity. However, in all loading conditions, except pure flexion moments, the complex arrangements of muscles in the trunk unquestionably cause counterbalancing forces and moments around all 3 orthogonal axes (Delleman, Drost, & Huson, 1992).

Raschke et. al., (1996) investigated a muscle activity prediction method based on neurocortical data, called the Distributed Moment Histogram method (DMH). The DMH method assumes that the external moment is interpreted by the CNS as a distribution of force requirements, as opposed to a single moment vector to be equated. The authors attempted to predict the muscle activity of the torso by observing neural activity patterns using the DMH. This method mimics cortical activity to determine a muscle activation pattern. It evaluates a distribution of moments about the torso, and consequently recruits muscles in response to the distribution. The DMH predictions were highly correlated with the optimization estimates ($r^2 = 0.99$) and observed activity values ($r^2 = 0.91$). However, only the DMH model predicted low recruitment trends among EO and RA. The authors describe the technique outlined in this study as a more realistic simulation of muscle activation patterns than the minimum stress-compression optimization formula of Bean et al. (1988).

Granata and Wilson (2001) evaluated the influence of trunk posture on spinal stability by implementing a biomechanical model of spinal stability, and validated the results with measured trunk muscle activity. A three-dimensional, six degree-of freedom (DoF), two segment model was developed. The stability model included a 3-D inverted double-pendulum supported by 12 muscles. Three-dimensional unit-moments of each muscle were determined from the vector product of muscle-insertions and the unit-vector of muscle force. Muscle force amplitudes were estimated, for 12 muscle groups, from the six DoF constraints and the 6 stability constraints. The model required that the predicted muscle forces be physiologically feasible, and that they satisfy both equilibrium and stability. Stability requires that the second derivative of the potential energy of a system is positive (Cholewicki & McGill, 1996). A quadratic optimization, with an objective function to minimize the sum of muscle stresses, was used to solve the set of muscle recruitment that simultaneously satisfied both stability and equilibrium. Experimental conditions involved 10 healthy subjects performing static lifting tasks at forward flexion angles, and asymmetrical postures as well as all combinations of these flexion and asymmetric postures. Trunk postures were recorded with tracking sensors and EMG was collected from 4 bilateral sets of trunk muscles (RA, EO, IO, and ES). The authors compared the predicted muscle recruitment to the measured EMG to evaluate model performance. Two different methods used to calculate spinal load were then compared. The first method calculated the spinal load to satisfy only equilibrium, with no regard for stability, and the second method calculated spinal load from the optimization output that satisfied both the equilibrium and stability constraints. This allowed for a measure of the influence of posture on spinal stability. Spinal loading was increased when using the stability model in upright postures, and similar values were obtained for flexed postures. Asymmetric postures were associated with as much as a 21% increase in spinal compression when compared to sagittally-symmetric exertions. The authors concluded that, when attempting to satisfy stability requirements, greater muscle recruitment is necessary than when satisfying equilibrium requirements alone, demonstrating that the motor control system responds to stability requirements, and lifting posture can affect stability. This is the first biomechanical explanation for increased antagonistic activation, that is, antagonistic activation is necessary to achieve stability, and greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures.

Stokes and Gardner-Morse (2001) used a multi-component cost function analysis to simulate a realistic muscle activation strategy that required that several conditions must be respected. These conditions include: the stability of the spine must be maximized and the spine must be in equilibrium, subject to minimal changes in trunk posture, minimal intervertebral displacements, forces and moments, and muscle forces must be greater than zero, yet physiologically feasible. The cost function was formulated as the sum of 5 individually weighted components. These five components are thought to perform the following functions in the optimization of the spine: minimize changes of trunk posture, minimize relative displacements between vertebrae, minimize intervertebral forces and moments, avoid high muscle forces, and maximize the smallest eigenvalues (and hence spinal stability). Experiments with human subjects were simulated by calculating solutions for various external loading conditions at the T12 level. Muscle activation patterns were gathered from 14 subjects in a similar task to have comparable activation patterns to test the model against. For comparison, the muscle activations from the model were measured by the linear regression slopes of activation-versus-force generated, and normalized by the muscle's maximum force. All muscles, even at angles where they are antagonistic, increased their level of activation with increasing effort. Realistic predictions of muscle forces were not obtained by minimizing muscle stresses alone. The experimentally observed muscle activation patterns were found to be closest to those predicted when the intervertebral forces and muscle stresses were both minimized. Their findings suggest that minimizing muscle forces alone is not a feasible optimization strategy.

Brown and Potvin (2005) developed an optimization scheme using a 52-muscle fascicle model based on data from Cholewicki and McGill (1996). The number of muscle fascicles included only those that crossed the L4/L5 joint and, therefore, created a moment about that joint. Four optimization strategies were tested with the new optimization model: 1) minimizing the sum of cubed muscle forces with stability constraints 2) minimizing the sum of cubed muscle forces with no stability constraints, 3) minimizing the sum of the squared intervertebral forces at L4/L5 with stability constraints, and 4) minimizing the sum of the squared intervertebral forces at L4/L5 without stability constraints. Spinal stability was approximated and controlled to levels based on those found through *in vivo* testing. The optimization model of Brown and Potvin (2005) was found to improve estimates of spinal loading by predicting realistic levels of trunk coactivation while constraining stability levels to that observed *in vivo*. Therefore, including moderate levels of stability as a constraint in the optimization modeling was found to improve prediction of coactivation and estimations of compressive loading when compared to an EMG-based model.

Although optimization models of the spine have considered the physiological costs of muscle activation and spinal loading, these costs have generally been considered individually or as competing objectives while, in reality, several costs may be taken into account in establishing an efficient muscle activation strategy (Stokes & Gardner-Morse, 2001). A major limitation of most optimization models is that the complex musculature is drastically over simplified, and they cannot realistically determine muscle recruitment and, therefore, lack biological sensitivity (Cholewicki & McGill, 1996; Stokes & Gardner-Morse, 2001). Optimization models generally underestimate joint compression forces by 23-43% when compared with EMG-assisted models (Cholewicki & McGill, 1996), in addition to the fact that they are also insensitive to movement (Raschke, et al, 1996). Although it satisfies the three moment equilibrium constraints, optimization modeling is not sensitive to individual differences in recruitment patterns, and lifting mechanics, and thus presents same solution for every individual (Cholewicki & McGill, 1995). The solution from an optimization model does not change to suit the environment. For example, neither fatigue nor discomfort will have an effect on the model's output. The innate complexity of the human spine, and all its functioning parts, requires that models be more representative of the physiologically occurring processes, such as individual muscle activation strategies.

2.3.3 EMG-Assisted Optimization Models

Another approach to determine load-sharing is based on EMG measurements, muscle geometry, and knowledge of joint kinematics. Muscles play a very important role in the spine system due to their active control of the position of the human body in both static and dynamic conditions (Crisco & Panjabi, 1990). Performing a complex task can induce subtle changes in trunk postures, or motion dynamics, which can result in different muscle recruitment patterns being used to execute a motion and, hence, in an altered EMG-muscle force relationship (Chaffin et al, 1999). The influence of muscle activity on spinal loading is commonly approached with EMG models, since they acknowledge that the segments about each joint are moved by several different muscles.

McGill (1992) developed a 3-D model to determine spine compression loads and the resulting passive tissue loads during a lateral-bending task. In order to estimate muscle forces, the model contained extensive anatomical detail, including the co-contraction of muscles and moments produced by passive tissues. This model was found to be sensitive to inter-subject differences in muscle recruitment patterns. The intensity of the muscle contraction was estimated using EMG rather than optimization methods. Segment kinematics were tracked and the corresponding kinematics of all five lumbar vertebrae were modeled from external rib-cage and pelvis motions. The linked segment model used the dynamic load of the lifting task as input. Reaction forces and moments were then based on the hand load. Next, the reaction forces were partitioned into restorative force components including both the passive tissues and active musculature. The contributions of the passive elements were calculated first because they produce force and generate moments in response to posture and subsequent tissue strains. The remainder of the restorative moment was then assigned to musculature of the trunk. Individual muscle forces were computed with the facilitation of EMG, and based on origin and insertion points, estimated muscle lengths, orientation, and muscle lines of action. The constant gain was applied to all active muscles, and therefore relative recruitment patterns were preserved. The modulated muscle forces were applied to the skeleton and the three joint moments were then calculated and summed. A major limitation of this method is that the moments often require a gain factor, as they rarely equilibrate the reaction moments for even one axis. Furthermore, there is generally no single common gain factor that will result in static equilibrium about all three orthogonal axes.

Granata and Marras (1993) used ten channels of EMG to calculate muscle forces, assuming a linear relationship between the EMG signal and produced force. The muscles recorded were the LD, ES, RA, IO, and the EO. The muscle force was calculated using the normalized EMG, cross-sectional area (CSA) an individualized gain factor, and modulation factors that described muscle behaviour as a function of length and velocity. The model calculated the two moments about the x (frontal) and z (longitudinal) axes, respectively, so the moment about the y-axis (sagittal) axis was set to zero. Muscle forces were calculated from EMG data, cross-sectional areas, subject gain, and unit vector directions for all 10 of the modeled muscle equivalents. The model predicted muscle generated moments about the L5/S1 axis, compression, lateral shear, and anterior shear forces that were then compared to the corresponding profiles obtained in the experimental stage of the study. All modeled muscle forces were within a physiologically acceptable range. It was found that the model accounted for over 80% of the variability in 80% of the trails. Therefore, the EMG-assisted model was capable of predicting trunk extensor moment based on the model's assumptions.

In order to determine whether conclusions about the sizes and distribution of forces were the same while using the same anatomical model, Cholewicki, McGill, and Norman (1995) compared three approaches to resolve the mechanically indeterminate muscle forces, EMG-assisted, optimization (OPT) and EMG assisted optimization (EMGAO). The structural biomechanical 'model' used to estimate tissue loads consisted of two distinct models. The first model was a dynamic, three-dimensional linked segment representation of the body that used externally applied dynamic forces and individual anthropometrics as input to compute the reaction forces and moments about the three orthopedic axes corresponding to the L4/L5 joint. The second model was an anatomically detailed model of a three dimensional pelvis and ribcage with intervening lumbar vertebrae. This 50 muscle fascicle model was used to partition the three reaction moments obtained from the linked segment model into their tissue components. The contribution of passive elements was also calculated from measured angular displacements of the torso with respect to the pelvis using a calibration factor that was based on resting ligament lengths and their known stress/strain characteristics. The remaining moment was then partitioned to the muscles using the OPT, EMG and EMGAO approaches.

The OPT scheme incorporated 2 constraints; an approximate minimization of both the muscle stress and spine compression. The maximum muscle stress constraint was increased in small steps until the first solution was found. The primary consequence of minimizing muscle stress is to impose activation of the majority of agonists (Bean et al, 1988). In order to minimize joint compression, muscles with the largest moment arms are recruited first, up until their stress limit, at which point muscles with the second largest moment arms are recruited, and so on.

The EMG method based all the initial estimates of muscle forces on the normalized EMG activity from 6 sets of bilateral muscles (RA, EO, IO, LD, TES and LES), cross-sectional areas, a maximum muscle stress of 35 Ncm^{-2}, and were modulated by muscle force/length and force/velocity relationships. The EMG to force relation was calculated and a common gain was applied to all muscles to obtain the best fit between the EMG predicted and measured moments in all three axes.

The hybrid approach (EMGAO) aimed to apply the least possible gain adjustment to the individual muscle forces estimated from method 1, while concurrently satisfying the requirement to balance moments acting about the L4/L5 joint. EMG assisted estimates of muscle forces were corrected to simultaneously satisfy all three moment equilibrium constraints. Therefore, the EMGAO sought to minimize the gain adjustment and, furthermore, individual gains were not to deviate far from unity. All moments had to be solved simultaneously such that the gains and, consequently, the muscle forces, were required to be non-negative.

The experimental sEMG signals were collected during three near maximum, isometric, ramp efforts in trunk flexion, extension and left and right lateral bending in a testing apparatus. EMG data served as input to the three-dimensional lumbar spine model (McGill, 1992). sEMG signals were converted to muscle force by the following equation (2):

$$F_i = (EMG_i/EMG_{maxi})^{1/1.3} * a_i\delta_{max} + P_i \qquad [2]$$

where F_i is the ith muscle force (N), EMG_i/EMG_{maxi} is the muscle activation level expressed as a fraction of its maximum myoelectric activity, a_i is the ith muscle cross-sectional area (cm^2), δ_{max} is the maximum muscle force generated per unit of cross-sectional area (set at $35 N/cm^2$), and P_i is the passive muscle force (N).

The EMGAO approach provides the closest possible match between predicted muscle forces and their respective activation profiles while satisfying moment equilibrium constraints. Substantially different predictions of muscle forces, and consequently joint loads, resulted from the OPT and EMG methods even in simple isometric, sagittal, and frontal plane efforts. The EMGAO approach resulted in somewhat similar conclusions to the EMG approach. The OPT approach to solve the redundant set of equations was found to be far less telling than an EMGAO approach when investigating injury mechanisms and injury risk. The EMGAO method preserved the biological variability in the muscle activation patterns seen in the EMG recordings while satisfying moment equilibrium constraints, which is desirable when looking at individual differences in tissue loads.

The main advantage of the EMG approach is that it does not require the minimization of any variable, and it accounts directly for the effects of muscle length and contraction velocity on the EMG/moment relationship. The major drawback is that more often than not, all three axes cannot be equilibrated simultaneously. Additionally, the inherent variability of EMG signals may be attributable to varying recruitment strategies for individual motor units within a large muscle.

2.3.4 Regression Models

All three of the aforementioned models (Optimization, EMG based, and Single Equivalent models) can be used to determine the compression force that arises as a result of various load/posture combinations. These methods have proven to be considerably labour-intensive and time consuming, as numerous inputs are required. Furthermore these models can only make predictions for the individual who is being observed (Potvin, 1997). Regression models allow us to calculate some dependent variable from various independent variables. Regression-based models have the advantage that the input measurements are easy to obtain. The NIOSH equation is an example of a regression-based model that was developed to calculate acceptable loads during sagittal plane lifting (Waters et al, 1993). The equation itself is based on task-specific characteristics that are defined by four measurements: horizontal distance from the ankles to the load (H), vertical location of the load (V), vertical displacement of the lift (D), and frequency of lifting (F). The revised NIOSH equation also accounts for asymmetrical postures and poor coupling (Waters, et al, 1993).

McGill, Norman, and Cholewicki (1996) investigated whether muscles should be considered individually, or in an additive manner, when estimating low-back compression. To test if muscle uncoupling is reasonable, the L4/L5 compression force estimates from McGill's 1992 model were represented by a regression equation with three inputs, (compression and three moments), and compared to estimates from the uncoupled method, which used a single equivalent moment generator about each axis, and summed the compressive loads. Three male subjects performed the same seven slow dynamic tasks, and the analysis methods as in Cholewicki and McGill (1996). The regression equation was generated from the compression and moment data that were predicted for the 3-D tasks by the 90 fascicle model of Cholewicki and McGill (1996). Seventy-five moment-compression combinations were selected from subject trials.

The simplified uncoupled method was found to overpredict the compressive load by an average of 22%. The greatest disagreement between the uncoupled model output, and that of the polynomial model, occurred with large moment combinations. The best correlation was seen when single pure moments were examined. For example, a moment combination with 100 Nm of flexion, 150 Nm of lateral bend to the right, and 100 Nm of axial twist in the clockwise direction, resulted in a predicted compression difference of 2774 N between the 2 models.

This equation recognized muscle coupling, was sensitive to muscle cocontraction, and was capable of predicting compressive loads when the three axis moments were all at a maximum. However, due to the fact that individuals cannot simultaneously generate maximal moments about all three axes, the validity of these predictions was reduced.

Potvin (1997) developed regression-based models that use the NIOSH *H* and *V* (Waters et al. 1993) values, trunk angle and subject and load mass to calculate L5/S1 compression forces during symmetrical loading tasks. The criterion values were calculated with a static, two-dimensional, linked-segment biomechanical model (LSM). Models were also developed to estimate the range of trunk angles so that the worst likely posture could be determined for a given load location. The regression models were developed with 8 male and 8 female subjects performing stoop and squat lifts. The models were validated with 3 male and 3 female subjects. All 22 subjects performed movements in the sagittal plane while being videotaped. Movements were performed with both the squatting and stooping techniques of lifting. For each posture, a static, 2-D, rigid linked-segment biomechanical model was used to calculate the L5/S1 compression force and the location of the upper body's centre of mass (CofM).

Two regression-based models, designed to provide accurate predictions with easily measured variables, were developed to predict the L5/S1 compression forces obtained with the linked-segment biomechanical model. Horizontal distance from the load to the ankles (H), vertical load height (V) and trunk angle were the only posture-related independent variables used in each model. The equations in the first model were developed with a stepwise regression analysis to predict the horizontal moment arms from the L5/S1 disc to 1) the upper body CofM and 2) the hand held load. The two moment arm estimates were combined with trunk angle, body mass and load mass to calculate L5/S1 compression force using standard mechanics calculations. Three independent variables were used in these two regression equations: the NIOSH horizontal value (*H*), the NIOSH vertical value (*V*), and the trunk angle (TA). The horizontal location of the trunk segment CofM was represented by the cosine of TA. The L5/S1 joint moment was then computed with the two regression equation outputs with the additional inputs of upper body weight and load weight. The L5/S1 compression force was then calculated using a single equivalent extensor with a moment arm of 6 cm. Thus, the total compression force could be calculated by summing compression vectors of the upper body weight and the load weight and the single equivalent muscle force.

The L5/S1 compression force was estimated directly with the second model. L5/S1 compression force was predicted using a single equation developed with a stepwise regression analysis. This equation required the following six measurements: subject mass, subject gender, load mass, *H, V*, and TA. Three additional variables were also included; the cosine of the TA, the product of subject mass and cosine of TA, and the product of the load mass and *H.* In order to account for the variability in posture-related compression force the inclusion of trunk angle was necessary for both models. All trunk angle observations were placed into 10 x 10 cm bins in an effort to establish the expected range of trunk angles associated with each load location. Regression equations were developed to predict the minimum and the maximum trunk angles observed in each binned load location, with only the NIOSH *H* and *V* as independent variables.
The results indicate that the H, V and trunk angle values combine to explain most of the variance in body posture. Although the estimates from the second model were sufficiently accurate, the errors were higher than those obtained with the first model. Therefore, the first model is recommended for use. This regression-based model is comparable to, if not an improvement on, the NIOSH equations for determining safe loads based on the biomechanical criterion, while requiring the same input measurements (Potvin, 1997).

2.4 Spinal Stability

Devoid of muscles, the human spine is naturally unstable. It has been shown that the entire spine can buckle with a vertical load of 20 N, and that the lumbar spine can buckle when subjected to a load of 88 N (Crisco, Panjabi, Yamamoto, & Oxland, 1992). A spinal stability limit of 88 N is substantially less than the spinal compression experienced during most lifting tasks (Granata & Marras, 2000) and the increased load tolerance can be attributed to the work done by the musculature of the trunk. Coactivation of agonistic and antagonistic trunk muscles is necessary to stiffen the lumbar spine, and consequently increase its stability (Bergmark, 1989; Cholewicki & McGill, 1996; Cholewicki, Panjabi, & Khachatryan, 1997; Gardner-Morse et al, 1995; Stokes & Gardner-Morse, 2003).

When the trunk is perturbed, the CNS responds by eliciting a well-coordinated activation pattern in the muscles in order to stabilize the system. Failure to recruit an appropriate activation response in a timely manner will likely incur an injury due to instability of the spine (Cholewicki & van Vliet, 2002). Neuromuscular control of stability is necessary to maintain equilibrium in the presence of potential kinetic or kinematic disturbances and to reduce the risk of musculoskeletal injury (Granata & Orishimo, 2001).

Coactivation increases spinal load and, concurrently, spine stability (Cholewicki et al, 1997; Granata & Wilson, 2001; Brown and Potvin 2005) and levels of trunk muscle coactivation rise with increasing stability challenges (Gardner-Morse, Stokes, and Laible, 1995). Antagonistic co-contraction is necessary to achieve stability; furthermore, greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures. Twisted and flexed postures are associated with increased spine compression relative to normal standing (Schultz, Andersson, Haderspeck, Ortengren, Nordin, and Bjork, 1982).

A more complex loading situation (e.g. standing with a twist and 30 degrees of flexion) can cause as much as 1800 N of compression.

While the additional stiffness produced by trunk coactivation acts to increase the stability of the trunk, there exists an additional consequence, specifically an increased compressive load, which tends to destabilize the spine (Gardner-Morse & Stokes, 1998; Granata & Marras, 1995). Cholewicki and McGill (1996) proposed a hypothetical relationship between task demand and injury risk (Figure 2). The premise of this relationship is that there is some optimal level of cocontraction that maintains a sufficient level of stability while keeping the risk of injury low.

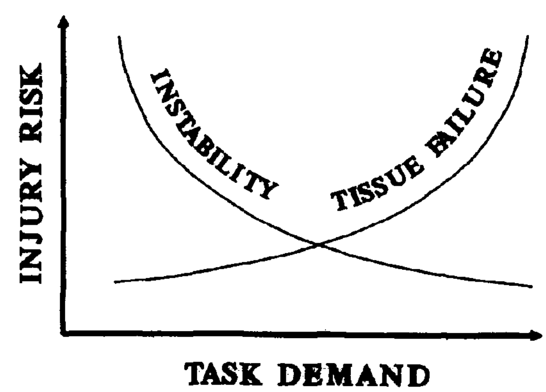

Figure 2. Hypothetical model for injury risk to the spine (Cholewicki and McGill, 1996).

Cholewicki and Van Vliet (2002) state that a prerequisite for spine stability is that all trunk muscles must be active at some level at a given instant, which includes antagonists. All trunk muscles are assumed to be involved in stabilizing the spine, and it has been observed that no single muscle group contributes more than 30% to overall stability and individual muscle contributions depend on recruitment patterns, and the direction and magnitude of load.

2.5 Role of Muscles in Spinal Stability

The stabilizing system of the spine was conceptualized by Panjabi (2003) to consist of three subsystems: spinal column (intrinsic stability), spinal muscles (dynamic stability), and neural control unit (evaluating and determining the requirements for stability and coordinating the muscle response). Under normal conditions, the three subsystems work in harmony and provide the needed mechanical stability. In anticipation of a predictable disturbance the CNS controls spinal stability by initiating a preparatory motion of the spine to `dampen' the forces instead of making the trunk rigid in a "worst-case scenario" strategy (Hodges, 1999).

Stability of a column is a discrete function of both its length and its bending stiffness (Crisco et al, 1992). The spine naturally optimizes an increase in stiffness from its cervical portion to lumbar portion, in that the lower portion has stiffer joints, and the buckling load is increased for the same average stiffness (Crisco & Panjabi, 1992). Spine stability is determined by the activation levels of all trunk muscles, ligamentous stiffness of the inter-vertebral joints, spine posture, and the magnitude, direction and end-conditions of external trunk loads (Cholewicki & VanVliet, 2002). In fact, practically every muscle of the trunk works together to create the optimal stiffness needed to ensure adequate stability in all degrees of freedom of the spine (McGill, Grenier, Kavcic, & Cholewicki, 2003). Stability, or instability, should be thought of as a state of the entire spine system, which is determined by the activation of all trunk muscles, passive joint properties, spine posture, and loading conditions (Cholewicki & VanVliet, 2002).

Cholewicki and VanVliet (2002) compared the relative contribution of 10 major trunk muscle groups to the stability of the lumbar spine by removing each muscle systematically from the biomechanical model developed for quantifying spine stability. No one trunk muscle was identified as contributing the most to spine stability under all different loading conditions. While the relative contribution of a given muscle to spine stability depends on loading magnitude and direction, the contribution of different trunk muscle groups to lumbar spine stability depends on the combination of direction and load level of isometric exertions (Cholewicki & Van Vliet, 2002).

More recently, results of Granata, Slota, and Wilson (2004) showed that when stability constraints are included in their model, significant changes in load predictions were observed. Their results suggest that spinal stability criteria must be considered in combination with spinal loads when evaluating the biomechanical risk of low-back injury. Additionally, Brown and Potvin (2005) found that the inclusion of moderate levels of stability as a constraint in optimization modeling allowed for the prediction of trunk muscle coactivity and improved predictions of spine compressive loading.

2.6 Mechanical Stability

Leonardo da Vinci was the first to investigate the mechanical stability of the spine, depicting the muscles of the neck as ropes (Crisco & Panjabi, 1990). The phenomenon of instability was first defined by Euler, a Swiss mathematician, in 1744. A column carrying a concentric compressive load is said to be stable, in the Euler sense, if the column returns to its original vertical position after it is perturbed (Crisco et al, 1992).

Biomechanically speaking, stability pertains to a mechanical structure that can become unstable when a 'critical point' is reached (McGill et al, 2003). Stability is defined as the ability of a system to return to equilibrium after a small perturbation (Gardner-Morse & Stokes, 2001). Mechanical stability of the lumbar spine must be maintained at all times to prevent its buckling (Cholewicki & McGill, 1996). The overall mechanical stability of the spinal column is provided by the spinal column itself and the precisely coordinated surrounding musculature (Panjabi, 2003).

Static equilibrium in mechanical systems requires that the sum of the forces and moments is equal to zero. In the spinal system, static equilibrium is achieved when the first partial derivative of the potential energy (PE) of the system is equal to zero (Bergmark, 1989). Crisco and Panjabi (1990) applied the principles of PE to the investigation of the stability of the spine, as influenced by gross musculature, and showed that coactivation of the back muscles is necessary to laterally stabilize the spine.

Mechanical overloading has the potential to cause injuries which, in turn, have a considerable effect on the mechanical behaviour of the spine. Biomechanical studies under controlled laboratory conditions, have provided some insight into the role of spinal column tissues (disc, ligaments and facets) in providing spinal stability (Panjabi, 2003). The spinal stabilizing system functions by altering the muscle activation pattern in response to the ligamentous tissue mechano-receptor signals (Panjabi, 2003). Also, increased activation in rest postures may be aimed at enhancing stability (van Dieen, Selen, & Cholewicki, 2003).

2.7 Stability in Spine Modeling

Cholewicki and McGill (1996) developed a method to analyze the overall stability of the multi-degree-of freedom *in vivo* lumbar spine under a wide variety of dynamic, 3-D loads and postures. This method built on previous work and incorporated anatomical detail for the entire length of the spine, and included a measure of stability termed the Stability Index (SI). The external forces were obtained from a quasi-dynamic, three-dimensional rigid linked-segment model. Moment generating forces arising from upper body segment masses and external forces were summed to calculate moments about each intervertebral level. The forces generated by the passive tissues were then calculated, and any remaining moment was then partitioned among the 90 muscle fascicles using EMG.

The spine was mathematically perturbed in all directions as part of a stability analysis. If the spine can return to its original position, it is considered to be mechanically stable. The potential energy of the system was calculated as the work done by the external load subtracted from the elastic energy stored in the muscles, tendons, intervertebral discs, ligaments, and passive tissues. The stability analysis was performed using the instantaneous spine geometry, external load, instantaneous muscle stiffness and their respective forces as input. An 18 x 18 Hessian matrix of the partial derivatives of the system's energy was compiled, diagonalized, and its determinant was calculated. The criterion for stability requires that the second derivative of the model's potential energy be positive, therefore any value less than or equal to zero would indicate instability. Therefore, the eighteenth root of the Hessian's determinant was selected as a relative measure of stability index (SI).

The SI was highest with increased moment demand or joint compression, and was the lowest when the demand was negligible. When considering asymmetrical tasks, the highest level of stability was attained when moment arm was the shortest, and the least amount of stability when the moment arm was long. The local stabilizing system was found to be inadequate for stabilization of the spine during light tasks. Also, the contribution of the passive elements to counteract the external moments acting on the spine were negligible, yet are thought to be very important during light tasks (Cholewicki & McGill, 1996). Furthermore, it was assumed that each intervertebral joint contributes a set amount of the total rotation angle between S1 and T12 in all three directions: flexion/extension, lateral bending, and axial twist. Translations between intervertebral joints were constrained, and the centres of joint rotation were constant. The diminutive spine stabilizing potential of small transversospinalis and rotatores were simulated with an increased passive joint stiffness.

Stokes and Gardner-Morse (2001) attempted to introduce trunk stability into the cost function of an optimization model. The inclusion of stability into the model did not improve the predictions, however the authors suggest that stability is not maximized, but rather set to some target level. The authors assumed that forcing antagonistic muscle activation, by introducing the trunk stability into the cost function, would improve the model. Since we do not know for certain whether the CNS sets specific parameters to be optimized, or if they are casually achieved, 'forcing' antagonistic muscle activation is potentially not the best approach to the inclusion of stability into models.

Cholewicki, Simons, and Radebold (2000) evaluated how various trunk load magnitudes and directions affect lumbar spine stability. Lumbar spine stability was shown to increase significantly in all directions with added horizontal load as estimated by the biomechanical model. When loaded vertically, the only significant change was predicted for flexion motion. It was concluded that lumbar spine stability increased with increased trunk load magnitude to the extent that this load brought about an increase in trunk muscle activation.

Chapter 3: METHODOLOGY

3.1 Study Design

The study aimed to develop a polynomial equation that uses moment and potential energy to predict joint compression forces at the L4/L5 level. The compression output from three existing biomechanical models was also calculated for comparison during 48 loading tasks (36 vertical, & 12 triaxial loading tasks). The selected biomechanical models of the spine provided an estimation of joint compression and shear forces, by distributing the moment generated about L4/L5 amongst the musculature of the trunk. Electromyographic (EMG) data were collected from 15 muscles during static load handling tasks. In addition, various types of biomechanical models of the spine were utilized to develop a simple equation that has more biological fidelity than previous statistical models used to predict lumbar joint loading during asymmetrical tasks. Subjects performed various one-handed static loading tasks, in a laboratory setting using their right hand. Subjects were all right handed. Subjects were asked to hold low-risk loads at set moment arms, shoulder angles and heights. A total of 36 three-dimensional vertical loading conditions were examined: each load was held at: 3 heights x 3 shoulder angles x 4 set moment conditions (n=36 vertical loading conditions). Additionally, for the sake of formulating regression equations, exertions causing moments about all three axes of the spine were performed during an additional 2 sets of triaxial loading trials. This subset of exertions was performed at 3 heights, 1 shoulder angle (0°), and 2 reach distances, for a total of 2 different triaxial moments (n=12). Three repetitions of each loading trial were performed. A summary of the experimental conditions can be seen below in Figure 3.

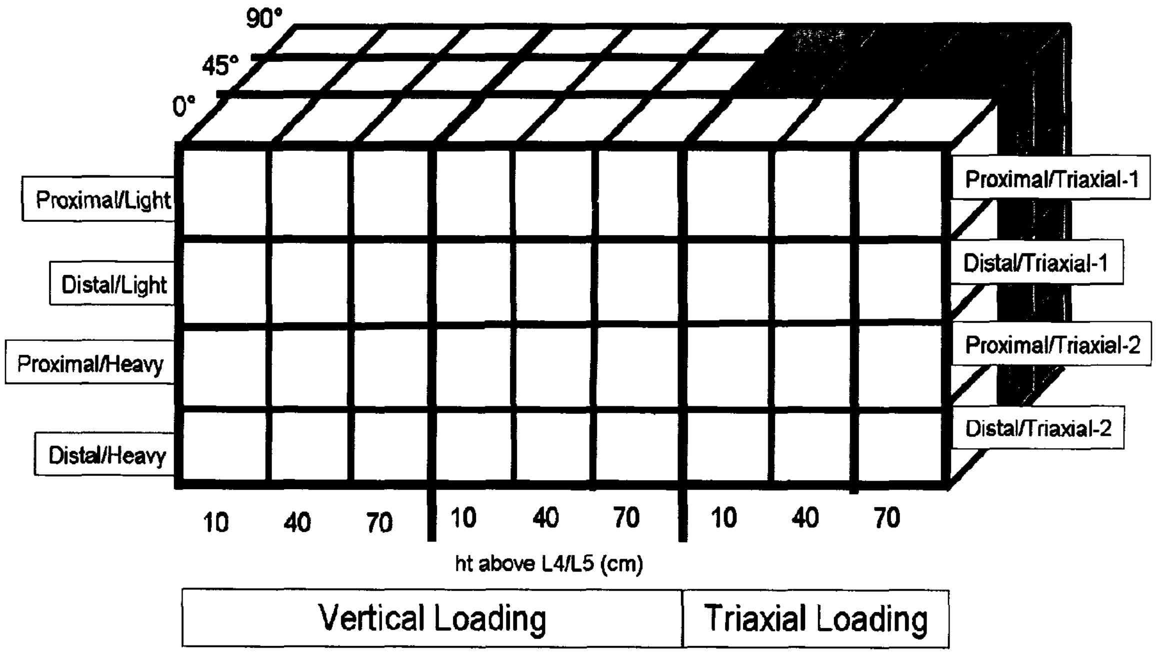

Figure 3. A summary of the experimental conditions. The four models will estimate compression at L4/L5 based on the various moment conditions and the height of the load. Moment conditions are defined by a horizontal reach distance (cm) and the angle of shoulder rotation from the midline (°). The load heights are 100, 135 and 170 cm. This block shows all the conditions during the loading trial. Note that Triaxial conditions are only performed at 0°, with one set mass.

3.1.1 Subjects

Data were collected from 16 healthy subjects (eight males; eight females). Outliers were identified and removed using the Grubb Critical Z Criterion ($p<0.05$), before calculating the final means and standard deviations. Using this method, one subject was determined to be an outlier and, thus, removed from all subsequent analyses. The mean (standard deviation) anthropometric data for all subjects (n=16) were as follows; age 24.3 (sd=1.53) years, height 178.1 (8.36) cm, mass 76.6 (12.6) kg. Subjects were screened for any prior back problems, or any known postural or fatigue related disorders prior to advancement into the experimental trials . All subjects were asked to provide informed consent preceding any participation in the study, by reading and signing the informed consent document (Appendix A) as well as consent form for audio/video taping (Appendix B). The details of this study were reviewed and approved by the University of Windsor's Research Ethics Board.

3.1.2 Study Tasks

For the static vertical loading tasks, the external load was applied at every combination of two reach distances (25 and 50 cm) and three transverse shoulder angles of rotation (0°, 45° and 90° rotation from the sagittal plane). The subjects were required to hold the load at three different heights: 1) 10 cm above L4/L5 height (waist height), 2) 40 cm above L4/L5 (chest height), and 70 cm above L4/L5 (head height). Subjects performed trials for 4 vertical loading moment conditions (Proximal/Light, Distal/Light, Proximal/Heavy, Distal/Heavy) (n=18 for each moment condition) and 2 sets of triaxial loading trials.

All 48 testing conditions were presented in a randomized order. Each trial was held for 3 seconds, and repeated three times. Two minutes rest was provided between trials and data collection took place during one session.

3.1.3 Independent Variables

This study incorporated independent manipulation of the 1) load height, and 2) shoulder angle, and 3) moment condition. Two reach distances and three angles of shoulder rotation, in the transverse plane, were combined in order to represent six three-dimensional hand locations within the reach envelope for each of the three height levels. Refer to Figure 3 for a summary of the study design.

3.1.4 Dependent Variables

The dependent variables were the EMG amplitudes measured for the individual muscles and the subsequent total L4/L5 compression forces that occurred as a result (based on an EMG-driven model).

3.2 Data Acquisition

3.2.1 Electromyography

Surface EMG (sEMG) was collected from the bilateral lumbar erector spinae (LES), thoracic erector spinae (TES), multifidus (MULT), latissimus dorsi (LD), rectus abdominus (RA), internal oblique (IO), and external oblique (EO), and also from the right anterior deltoid (RDELT). Bipolar Ag/AgCl surface electrodes (Medi-trace disposable electrodes, The Ludlow Company, Chicopee, MA) were used to monitor the muscle activation levels. Four channels were dedicated to posterior musculature (LES, TES, MULT, LD) three channels were dedicated to the abdominal musculature (RA, IO, EO) and the last channel was used for the RDELT.

sEMG data were collected with LabView software (National Instruments, Austin, TX) using a PC compatible computer and converted by a 12-bit A/D card (National Instruments, Austin TX). All sEMG signals were processed through a differential amplifier (gain = 1000 to 5000, input impedance = 10 GΩs, 10-1000 Hz, CMRR = 115 dB at 60 Hz, Bortec, Octopus AMT-8, Calgary, Canada). sEMG signals were digitally sampled at 2048 Hz. EMG data were then bandpass filtered (20-1000 Hz), full wave rectified, and then low-pass filtered using a 2^{nd}order Butterworth filter with a frequency cutoff of 2 Hz.

3.3 Experimental Protocol

3.3.1 Laboratory Familiarization

Upon arrival in the laboratory, the protocol was explained to the subjects and they were familiarized with the instrumentation. During this time, subjects were asked to complete consent forms and a background questionnaire covering variables such as height, age, weight, and health limitations that may have affected their inclusion in the study.

Next, bipolar Ag/AgCl surface electrodes were attached over the belly of fifteen muscles. The area for electrode placement was cleansed and treated to reduce impedance. Locations of electrode placement were as follows (Cholewicki & McGill, 1996; Brown & Potvin, 2005): LES (3 cm lateral to L3 spinous process, TES (5 cm lateral to T9 spinous process), MULT (2 cm lateral to L4-L5 spinous process), LD (lateral to T9 over the muscle belly), RA (3 cm lateral to the umbilicus), IO (approximately midway between the anterior superior iliac spine and symphysis pubis, above the inguinal ligament), and EO (approximately 15 cm lateral to the umbilicus). QL activity was estimated from the LES electrode location. The intra-electrode distance was 2.5 cm.

3.3.2 Experimental Conditions

Prior to the testing sessions, maximum voluntary efforts (MVEs) were collected from each of the 14 muscle sites for the purpose of normalization of experimental EMG data. A series of maximal contractions were performed against an externally applied resistance through a range of motion. For posterior muscle MVEs, subjects performed a series of back extensions while lying prone with their hips and legs secured to a table.

For abdominal muscle MVEs, subjects performed a series of isometric trunk flexion efforts to the right and left and in the sagittal plane.

The loading trials were then presented to the subjects in a randomized manner. Subjects performed one-handed static lifting tasks with their dominant hand at all combinations of: 3 load heights (10, 40, and 70 cm above L4/L5), 3 angles of transverse shoulder rotation (0°, 45° and 90° from the midline), for each of the vertical loading moment conditions (n=36).

Within the first set of loading trials at the three loading heights, subjects held 2 set load magnitudes at 2 set distances from the right acromion; 2.7 kg at 0.25 m and 1.4 kg at 0.50 m. This loading combination created the Proximal/Light and Distal/Light moment conditions, respectively. Throughout all conditions in this set of loading trials the external moment imposed on the body by the hand load was 6.7 Nm. Subjects performed both the Proximal/Light and Distal/Light moment combinations at three angles of shoulder rotation (0°, 45, and 90°) (Figure 4) and the three loading heights.

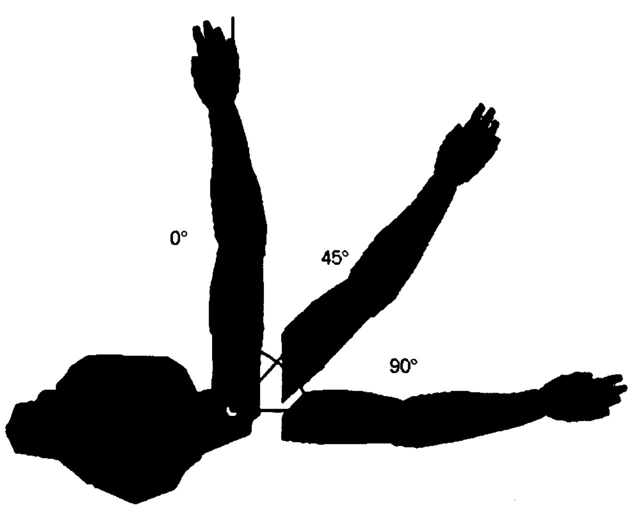

Figure 4. 3 angles in the transverse plane with respect to the sagittal plane through the shoulder (0°, 45°, 90°).

Within the second set of loading trials at the three loading heights, subjects held masses of 5.4 kg at 0.25 m and 2.7 kg at 0.50 m. This loading combination created the Proximal/Heavy and Distal/Heavy moment conditions, respectively. Throughout all conditions in this set of loading trials the external moment imposed on the body by the hand load was 13.3 Nm. Subjects performed both the Proximal/Heavy and Distal/Heavy moment combinations at three angles of shoulder rotation (0°, 45, and 90°) and the three loading heights.. A summary of the parameters, including moment arms and hand loads, for the vertical loading moment conditions can be seen in Table 1

Table 1. A summary of vertical loading moment conditions

Moment Condition	Moment Arm (cm)	Hand Load (kg)
Proximal/Light	25	2.7
Distal/Light	50	1.4
Proximal/Heavy	25	5.4
Distal/Heavy	50	2.7

Additionally, two sets of triaxial hand loading trials were achieved by utilizing 2 different pulley systems. The first set of triaxial loading trials, Triaxial-1, required the subject to perform a hand effort downwards, forward, rightward. The second set, Triaxial-2, required the subject to perform a hand effort upward, backward, and leftward. The resisted load was 44.5 N for both triaxial conditions. The Triaxial-1, and Triaxial-2 conditions were performed at each height and reach combination with the shoulder at 0° (n=6 for each triaxial loading set).

Each subject was set up so that the L/4 L5 joint was in the same vertical position. Subjects were asked to stand with their hips pressed to a foam-padded saddle that was set at a constant vertical height. Platforms were used under the feet, as required. Next, the acromion was located and used as the origin for all hand locations. A 3D grid of 18 hand locations was created with a series of light strings attached to an overhead surface. A series of Styrofoam balls were hung from the strings. For each particular tasks, subjects held the appropriate load in their hands and reached that hand to the appropriate Styrofoam ball.

3.4 Data Analysis

3.4.1 Estimation of Compression Values

Lumbar compression force was calculated for each of the 48 conditions with 4 biomechanical approaches: 1) Cholewicki and McGill (1994) EMG/Optimization hybrid model, 2) McGill, Norman and Cholewicki (1996) 3rd order polynomial equation (3) Michigan 3-Dimensional Static Strength Prediction Program (3DSSPP), and (4) the polynomial regression equation developed in this study.

3DSSPP software was used to determine the reaction moments and forces about L4/L5 by inputting external hand forces and force vectors for desired anthropometry. This was achieved by manipulating the Michigan 3-D mannequin into the body postures adopted by the subjects in order to replicate joint kinematics. The reaction moments were then used as input to other models, and the compressive forces were compared between spine models. Figure 5 illustrates how the modeled compression forces were obtained, and subsequently compared.

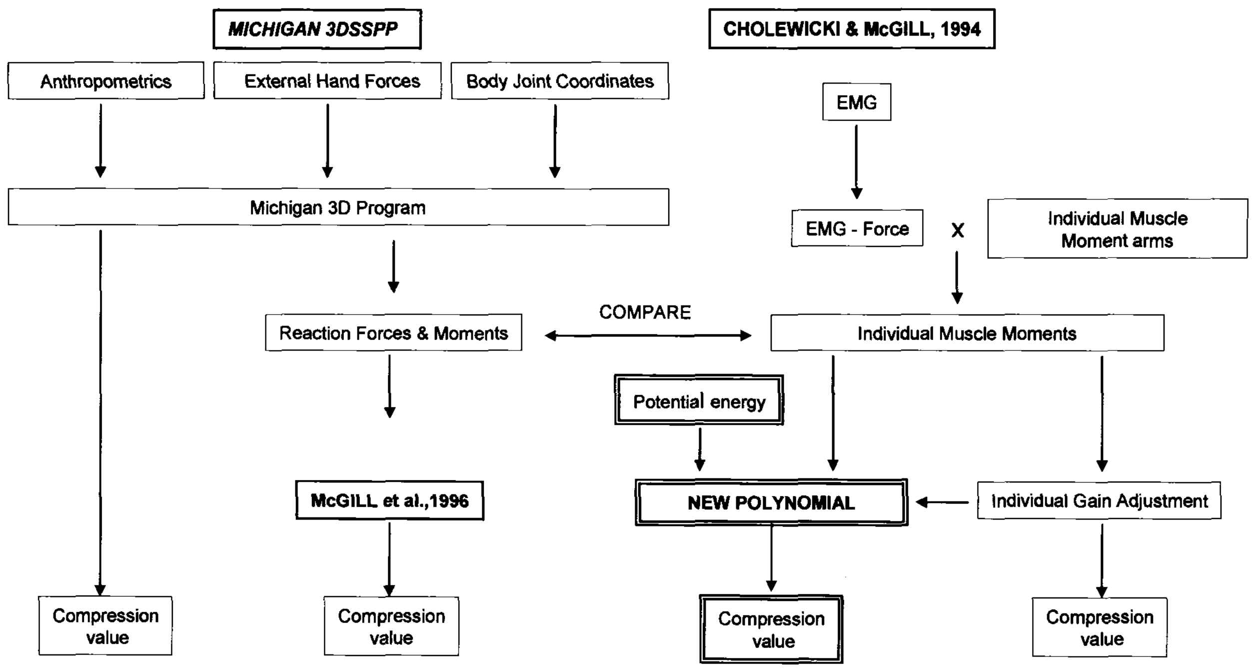

Figure 5. Outline of the Study. A flowchart cascading from external hand forces to the calculation of L4/L5 compression. Compression values will be output from Michigan 3DSSP, Cholewicki & McGill, (1994), McGill et al. (1996), and eventually the polynomial developed from this study.

The actual EMG data were averaged and used to represent activation patterns during individual trials. In the case of Cholewicki and McGill (1994), normalized EMG and muscle length data were used to calculate individual muscle moment contributions. The net moments about all three orthogonal axes were then balanced simultaneously using an EMG-optimization hybrid technique to match the moment calculated by the linked segment model by assigning individual gains to the force profile of each muscle. Specifically, in order to balance the net moments, muscle gains were set at a level that resulted in the least amount of adjustment possible within a physiological range. In this way, the moments are equilibrated while the adjustments made to the muscle forces are kept to a minimum. The total compression calculated was then compared to that of the polynomial of McGill et al. (1996), 3DSSPP, and the regression-based equation from the present study.

3.4.2 Statistics

A 3-way Repeated Measures ANOVA was performed to determine the main effects and interactions of the independent variables on EMG amplitude. The independent variables were: 1) load height, 2) shoulder angle, and 3) moment condition. The dependent variable was the normalized EMG amplitudes for each monitored muscle. Tukey's Post-hoc analyses were run to interpret any significant main effects or interactions that were found.

3.5 Development of Equations

There are a number of biomechanical models that are capable of predicting L4/L5 spinal disc compression. However, the numerous assumptions and limitations, as well as highly involved measurements, make some of these models impractical for use in ergonomics in the field. The final goal of this study was to develop a regression based tool that would be used to predict the compression at the L4/L5 level, given the three joint moments and, most innovatively, the potential energy of the hand held load. This equation uses values that are easily measured in order to accurately estimate L4/L5 compression during asymmetrical, isometric loading tasks. The forward stepwise regression was performed using inputs of the height of the subject, the height of the load above the L4/L5 level, body mass, the flexion moment acting at the L4/L5 level, the lateral moment acting at the L4/L5 level, the axial moment acting at the L4/L5 level, as well as squared and cubed values of each of the aforementioned moments.

The three moments at the low-back, and the load potential energy from the time-histories of all subject-trials, were input as variables in StatView (SAS Institute Inc., 1992-1998) software. The output from the model of Cholewicki et al, (1995) has been selected as the criterion, as it is based on EMG recorded from the trunk muscles and was sensitive to between and within subject differences for each trial of each condition. A step-wise regression was used to develop a polynomial equation to estimate the compression forces from the EMGAO model of Cholewicki et al. This equation was developed using variables from the entire subject sample (15 subjects). The compression values generated by the new polynomial were compared against those predicted by the common compression estimation techniques described above to evaluate the performance of this model, using the EMGAO as criterion for the best possible solution. The predicted values from this equation were then compared to the common compression estimation techniques described above. All comparisons included correlations, and RMS differences between the biomechanical methods.

Chapter 4: RESULTS

4.1 Independent Variables

Fifteen repeated measures ANOVAs were conducted to determine the effect of the various independent variables on muscle activity for each muscle examined. The independent variables were height, moment condition, and shoulder angle. The average EMG amplitudes were calculated as a percentage of MVE for each condition.

There were main effects for load height, shoulder angle, and moment condition. Main effects were subsequently incorporated into 3 two-way interactions including moment condition x height, shoulder angle x height, and moment condition x shoulder angle, as well as three-way interactions of height x shoulder angle x moment condition. Alpha level was set at 0.05 for all comparisons. Significant interaction and main effects are presented in Table 2. Although there were significant main effects for both height and moment; both independent variables were involved in higher level interactions and, therefore, post-hocs were not performed. Post-hoc analyses are summarized in Table 3 and Table 4 for the significant interaction effects.

Table 2. Significant interaction and main effects ($p < 0.05$) presented for muscle activation levels. Blank cells indicate that no significant differences occurred. Shaded cells indicate the significant main effects and interactions for which post-hoc analyses were performed. Moment (Mo) Shoulder Angle (Ang) Loading Height (Ht). Note that no post hocs were performed on the Rectus Abdominis due to its low activity level.

		Main Effects			Interaction Effects			
		Angle	Moment	Height	Mo x Ht	Mo x Ang	Ang x Ht	H x A x M
Muscle	Side							
External Oblique	Right	0.0009	0.0001	0.0002	0.0001	0.0344	0.0223	
	Left	0.0001	0.0001	0.0001	0.0001	0.0001	0.0042	
Internal Oblique	Right	0.0001	0.0001	0.0002	0.0001	0.0001	0.0001	
	Left	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001	
Rectus Abdominis	Right		0.0001	0.0001	0.0001			
	Left	0.0001	0.0051	0.0160	0.0034	0.0001		
Lumber Erector Spinae	Right	0.0001	0.0001	0.0006	0.0001	0.0001		0.0006
	Left	0.0001	0.0001	0.0001	0.0001	0.0001	0.0006	0.0075
Thoracic Erector Spinae	Right		0.0001	0.0001	0.0001			0.0001
	Left	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001	0.0077
Multifidus	Right	0.0001	0.0001	0.0001	0.0006	0.0001	0.0018	0.0043
	Left	0.0001	0.0001		0.0001	0.0001	0.0001	
Latissimus Dorsi	Right	0.0008	0.0001	0.0001	0.0001		0.0001	
	Left	0.0067	0.0001	0.0001	0.0001			
Deltoid	Right	0.0001	0.0001	0.0001	0.0001			

Table 3. Summary of post-hoc results for significant main effects, and 2-way interactions ($p<0.05$).Moment conditions are identified as follows: Proximal/Light (PL), Distal/Light (DL), Proximal/Heavy (PH), Distal/Heavy (DH).

Muscle	Side	Angle	Moment	Height	Effect: Moment * Height				Effect: Moment * Angle				Effect: Angle * Height		
					PL	DL	PH	DH	PL	DL	PH	DH	0°	45°	90°
Eternal Oblique	Right						H>(M=L)	H>L		0°>90°	(0°=45°)>90°	0°>45°>90°	H>(M=L)	H>(M=L)	H>(M=L)
	Left				H>L		H>M>L	H>(M=L)	90°>(45°=0°)	90°>(45°=0°)	90°>45°>0°	90>(45°=0°)		H>L	H>(M=L)
Internal Oblique	Right						H>(M=L)	H>(M=L)		0°>(45°=90°)	0°>(45°=90°)	0°>(45°=90°)	H>(M=L)	H>M>L	H>M>L
	Left						H>M>L	H>L	90°>45°>0°	90°>45°>0°	90°>45°>0°	90°>45°>0°	H>L	H>(M=L)	H>M>L
Rectus Abdominis	Right														
	Left														
Lumber Erector Spinae	Right														
	Left														
Thoracic Erector Spinae	Right														
	Left														
Multifidus	Right														
	Left					L>H	H>L		(0°-45°)>90°	(0°-45°)>90°	(0°-45°)>90°	(0°-45°)>90°	(L=M)>H		
Latissimus Dorsi	Right				H>(M=L)	H>L	H>M>L	H>L					H>M>L	H>M>L	(H=M)>L
	Left	(0°=45°)>90°			H>(M=L)	H>(M=L)	H>M>L	H>(M=L)							
Deltoid	Right	(0°=45°)>90			H>M>L	H>(M=L)	H>M>L	H>M>L							

Table 4. Summary of post-hoc results for significant 3-way interaction (p<0.05). Moment conditions are identified as follows: Proximal/Light (PL), Distal/Light (DL), Prosimal/Heavy (PH), Distal/Heavy (DH).

Muscle	Height * 0° * Moment				Height * 45° * Moment				Height * 90° * Moment			
	PL	DL	PH	DH	PL	DL	PH	DH	PL	DL	PH	DH
REO												
LEO												
RIO												
LIO												
RRA												
LRA												
RLES			H>M>L	H>(M=L)			H>(M=L)	H>L			H>L	
LLES		L>H	H>(M=L)				H>M>L	H>(M=L)			H>(M=L)	
RTES	H>M>L	H>(M=L)	H>M>L	H>M>L	H>M>L	H>(M=L)	H>M>L	H>M>L	H>(M=L)	H>(M=L)	H>M>L	H>(M=L)
LTES	H>M>L	M>(H=L)	H>M>L	(H=M)>L	H>M>L	(H=M)>L	H>M>L	(H=M)>L	H>M>L	(H=M)>L	H>M>L	(H=M)>L
RMULT		H>M	H>(M=L)	H>(M=L)			H>L	H>(M=L)				
LMULT												
RLD												
LLD												
RDELT												

4.2 Main effects

4.2.1 Height on Muscle Activation Levels

Increased load height resulted in a generalized increase in myoelectric activity in 14 of the 15 monitored muscles, with the exception of the left multifidus when the average EMG amplitude was pooled across angle and moment conditions (Figure 6). Relative to the condition where the load was held 10 cm above the L4/L5 level (low-height), the muscles on the right side of the body increased approximately 148% at 40 cm (mid-height), and 212% at 70 cm (high-height). The height effect was also significant for the muscles on the left side of the body; however, the EMG amplitude was not as high. The average EMG amplitude of the left musculature showed increases in activation with increased height. When pooled across angle and moment conditions, the muscles on the left side of the body demonstrated increases of approximately 114% at 40 cm, and 142% at 70 cm above the L4/L5 joint.

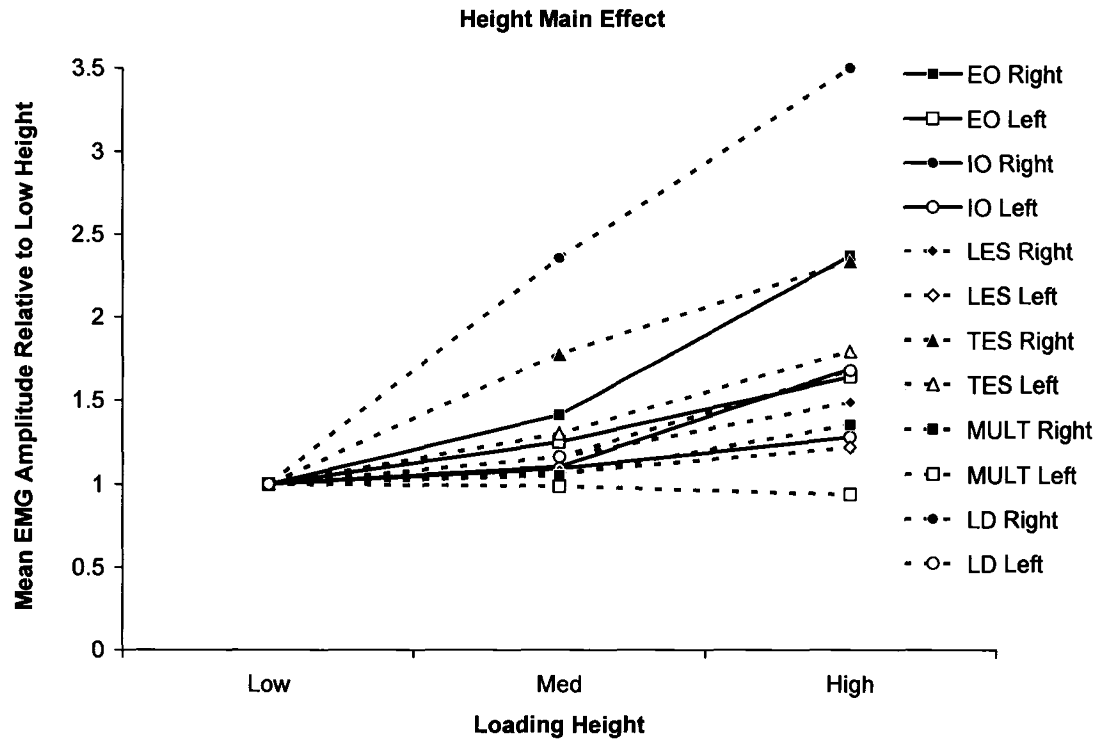

Figure 6. The average EMG amplitude pooled across angle and moment conditions, relative to the low condition. All muscles examined demonstrated increased myoelectric activity in response to increased load heights, with the exception of the left Multifidus (n=60).

4.2.2 Angle Effect on Muscle Activation Levels

Muscle activation was significantly affected by shoulder angle for all muscles, excluding the RRA and RTES. The LLD and the RDELT were the only muscle groups that did not show a significant interaction that included angle and, as such, post hoc analyses of the main effect of angle is presented for these two muscles ($p<0.05$, Figure 7).

Increasing the shoulder angle from 0° to 45° resulted in a significant increase in EMG amplitude for the LLD with muscle activation being 25% higher at 45° relative to 0° ($p<0.05$). Note that the LLD activation at 90° was not statistically different from either of the other shoulder angle conditions.

Conversely, there was no significant difference between the 0° and 45° conditions for the RDELT, however muscle activation was 24% and 19% higher at 0° and 45° respectively, relative to 90° (p<0.05).

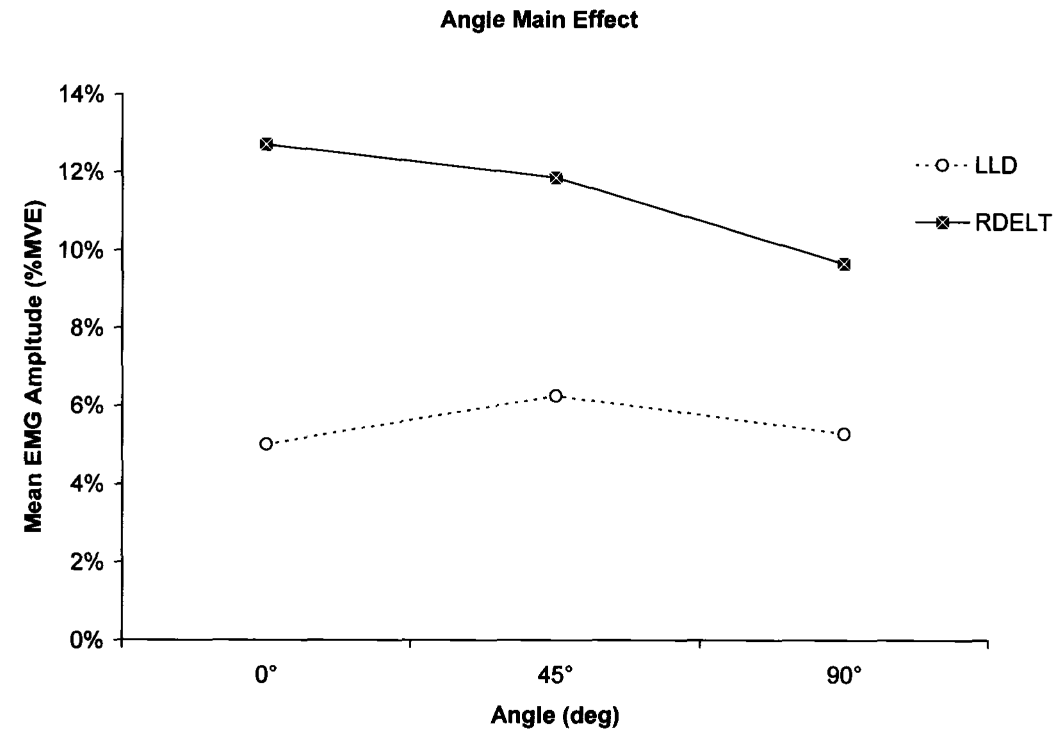

Figure 7. Main effect of angle, expressed as a percentage of MVE levels. Data were collapsed across moment and height conditions (n=60).

4.3 Interaction Effects

4.3.1 Moment x Angle Effect on Muscle Activation Levels

Within each moment condition, the EMG activity of the REO and RIO was observed to decrease as the angle of the shoulder increased (became more lateral). For these two muscles, the proximal light condition revealed no significant differences, but significant differences did exist within each of the other 3 moment conditions (Table 3, p<0.05). When comparing the activation trends of the RIO to the REO, the RIO was found to have more marked decrease in activation from 0 to 45 degrees, than from 45 to 90 degrees. The REO displayed a more uniform decrease as shoulder angle was increased. Figure 8 contains data describing the interaction between moment and shoulder angle conditions on the EMG activity.

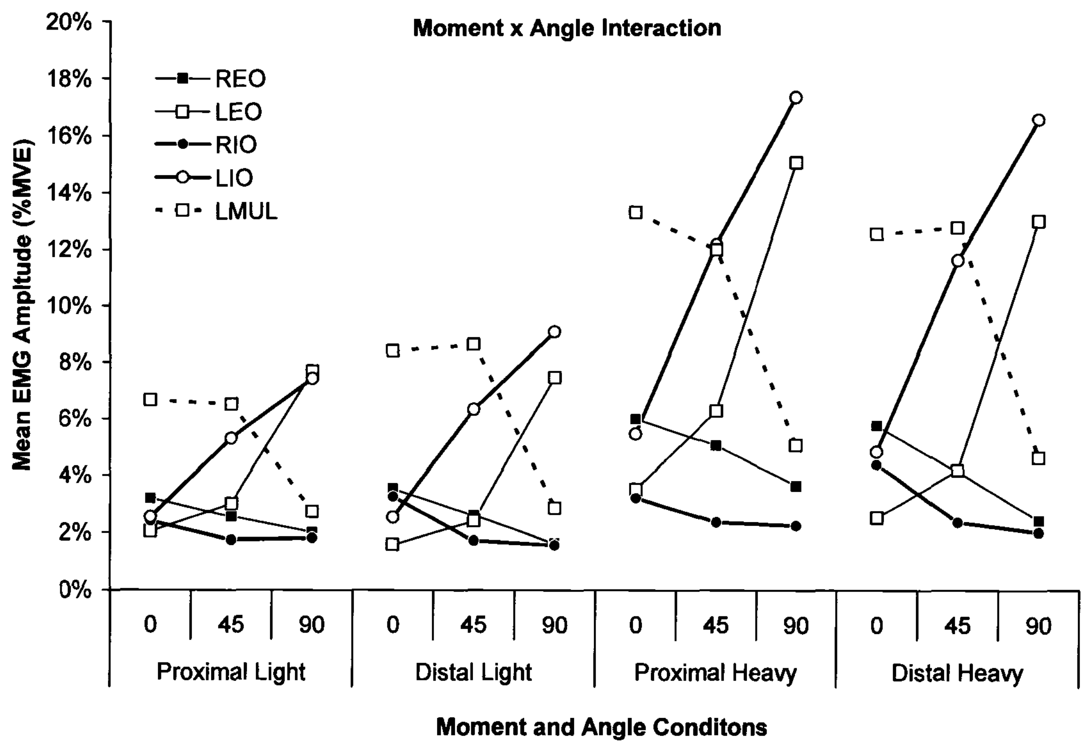

Figure 8. Mean EMG amplitude levels for all moment conditions, comparing shoulder angles: 0, 45 and 90 degrees of external rotation. Data were collapsed across height conditions.

The opposite trend was observed for the left abdominal musculature. EMG amplitudes were observed to increase with increasing shoulder angle. The LIO was subject to more consistent increases with corresponding increases in shoulder angle, while the LEO activation stayed relatively constant from 0 to 45 degrees, but increased significantly from 45 to 90 degrees. While the LIO demonstrated an average increase of 30% MVE from 45 to 90 degrees, when averaged across all moment trials, the average increases in activation of the LEO from 45 to 90 degrees was 64% MVE.

The LMUL activation levels were relatively constant from 0 to 45 degrees within each moment condition, but there was a large EMG decrease when the shoulder was rotated to 90 degrees (163%, when averaged across all moment conditions).

The effect of these interactions was amplified for each muscle as the moment conditions increased from Proximal/Light to Distal/Heavy. Muscle activation levels in the proximal and distal light conditions, including all muscles, ranged from 1.6% to 9.1% of MVE, while the range of recorded muscle activation in the proximal and distal heavy conditions ranged from 2.0% to 17.3% of MVE.

4.3.2 Angle x Height Interaction Effect on Muscle Activation Levels

Muscle activation levels were significantly affected by a 2-way shoulder angle x height interaction for the REO, LEO, RIO, LIO and RLD (Figure 9). Although analyses revealed a significant interaction effect for the RLES, LLES, RTES, LTES, and RMUL, these muscles were also found to be incorporated into a significant 3 way interaction, and they will be discussed later. At each angle, there was a general increase in muscle activity from Low (10 cm) to Mid (40 cm) and Mid to High (70 cm) load heights. All muscles showed a significant increase with increasing height, except the LEO in the 0 degree condition, and the LMUL in the 45 and 90 degree conditions. The average EMG amplitude in the right abdominals was found to decrease with increasing shoulder angle, however increases in activation were still seen when the loading height was increased (Figure 9). Within each angle condition, the REO and RIO were observed to have relatively constant activation during the Low and Mid heights, and a more pronounced increase at the High height. The LEO and LIO showed steady increases in muscle activity associated with increasing load height, within the angle conditions. This effect was amplified as shoulder angle was increased.

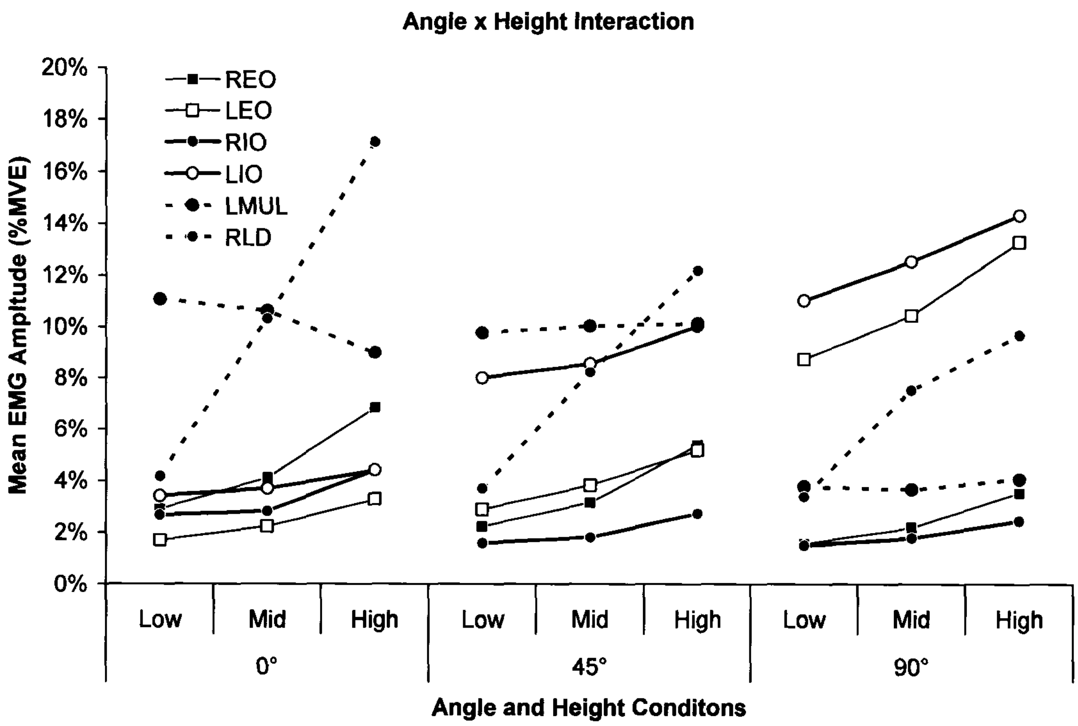

Figure 9. Mean EMG amplitude levels for each angle condition, comparing three load heights: low, mid and high-height relative to L4/L5. Data were collapsed across shoulder moment conditions (n=60).

The RLD showed the most dramatic interaction effect of shoulder angle and load height. While the RLD activity level was generally the same in the low condition, across all angles, it was observed to have especially large increases in activation within the 0 degree condition. The extent of this increase diminished with increasing angle.

4.3.3 Moment x Height Effect on Muscle Activation Levels

There was a significant Moment x Height interaction effect on muscle activation level for all muscle groups ($p<0.05$). Within each moment condition, all muscle groups demonstrated the highest level of muscle activation in the High height condition, and the least amount of activity at the Low height, with the exception of the LMULT in the Distal Light condition, where the greatest level of muscle activation occurred at the Low height. The RTES, LTES, RLES, LLES, and RMUL, with their 3-way interactions, were once again exempted from this section, and the remaining 8 muscles will be discussed here.

Changes in Height significantly affected the activity pattern of all trunk muscles across all Moment conditions, except for the REO, RIO, and LIO in both of the Light conditions, the LEO in the Distal Light condition only, and the LMUL in the Proximal Light and Distal Heavy conditions. While these muscles maintained the general activation trends, the interactions were not identified as significant.

The High height condition resulted in greater EMG amplitude than the Mid and Low height conditions for 16 of the 23 significant comparisons identified. The Mid height condition muscle activity was not significantly different from either of the other height conditions in 6 of the 23 comparisons; however there was still a significant difference between the High and the Low height for all comparisons. Lastly, the only significant decrease in muscle activation with increasing height was the LMUL in the Distal Light condition (Figure 10).

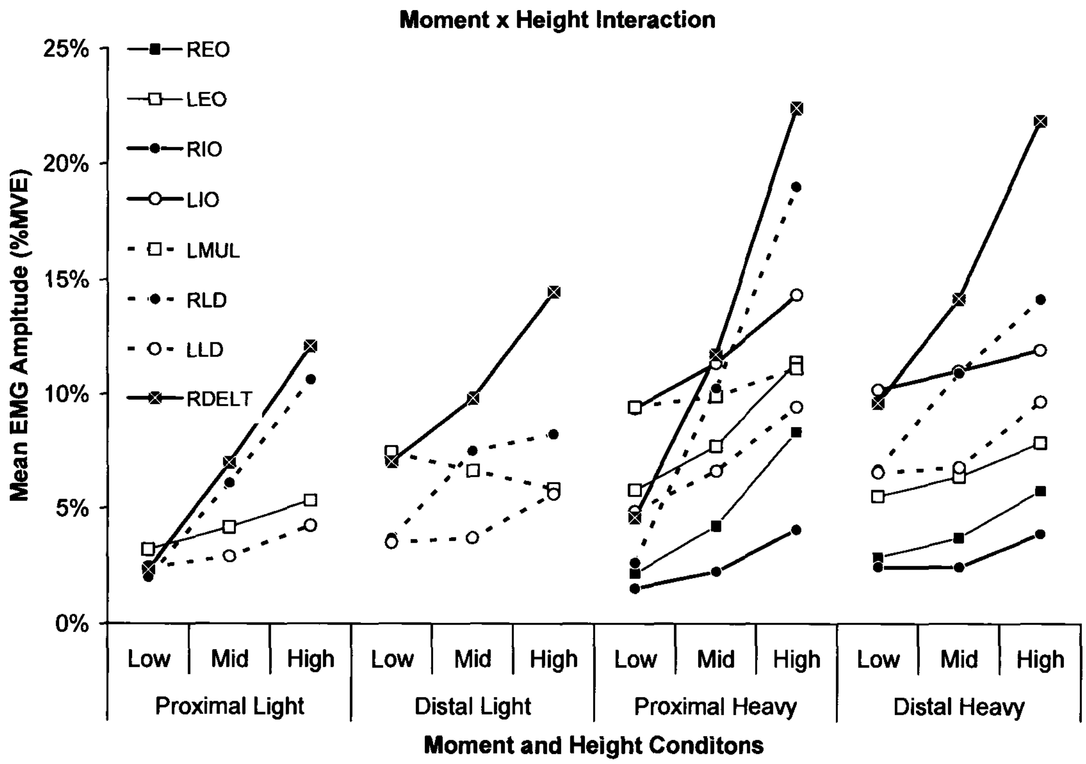

Figure 10. Mean EMG amplitude levels for each moment condition, comparing three load heights: low, mid and high-height relative to L4/L5. Data were collapsed across shoulder angle conditions (n=60).

Across all Moment conditions, the RDELT was the muscle most affected by Height (Figure 10). Increasing Height caused significant increases in the activity of the RDELT, especially in the proximal heavy condition. The RDELT activation showed increases of 520%, and 488%, relative to the Low height, in the Proximal Light and Proximal Heavy conditions, respectively. Interestingly, in the both the Light and Heavy Distal conditions, the lowest recorded activation was higher than that of both Proximal conditions. The EMG amplitude however, increased only 206% and 228% in the Distal Light and Distal Heavy conditions, respectively.

The RLD also exhibited marked increases in activity. Increases in the loading height during the Proximal Light conditions resulted in a 535% increase in muscle activation when comparing the High height to the Low height condition. Again, while the lowest recorded activation was greater in the distal conditions, the EMG amplitude increased 223% and 211% in the distal light and distal heavy conditions, respectively.

4.3.4 Moment x Height x Angle Effect on Muscle Activation Levels

There were five significant 3-way, Moment x Height x Angle interaction effects (RLES, LLES, RTES LTES and RMUL EMG data). The following sections will report the post-hoc results, organized by muscle. Standard deviations are provided in Appendix C.

4.3.4.1 Right Lumbar Erector Spinae

EMG amplitude for the RLES was observed to decrease with increased shoulder angle. While the RLES EMG remained relatively constant at all three heights for the Proximal and Distal Light conditions, the RLES was recruited most heavily at 0 degrees; this interaction effect was most prominent in the Proximal Heavy condition. While the 45 and 90 degree conditions maintained relatively low activation levels, the activity level of the RLES was greater at 0 degrees. At the High height, EMG amplitude was 9% MVE, while it was activated at only 5% MVE at 45 degrees, and 3% at 90 degrees. (Figure 11).

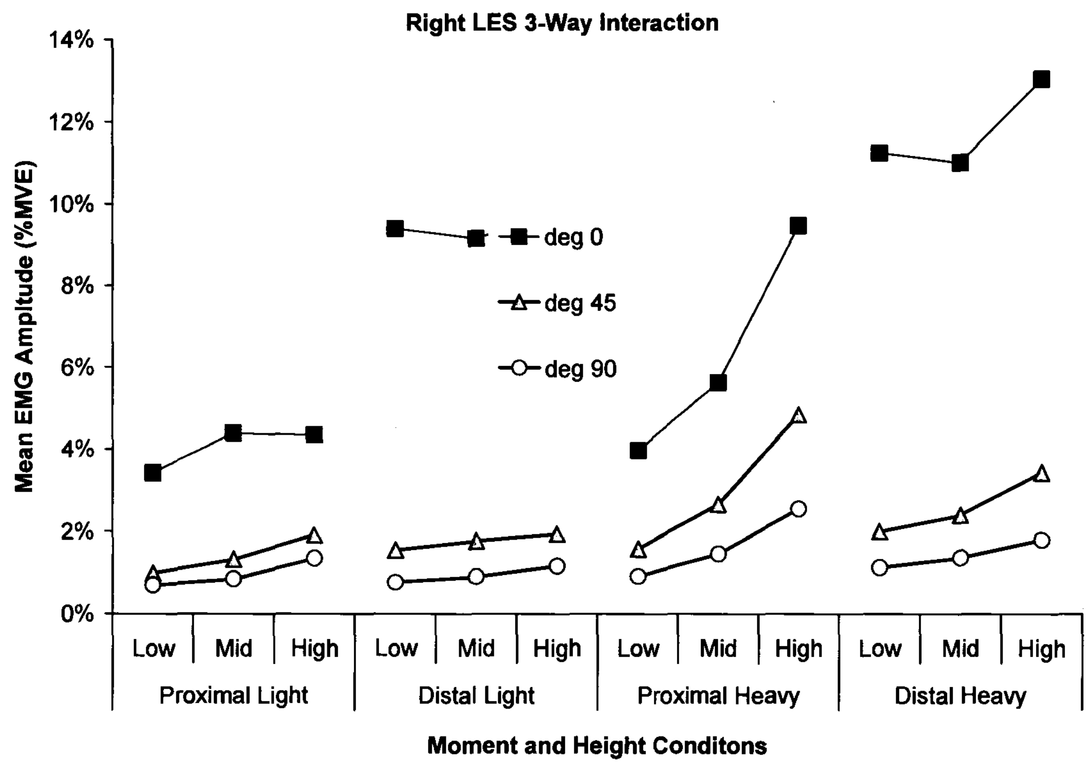

Figure 11. 3-way interaction for the RLES EMG amplitude. Mean EMG amplitude levels for each moment and angle condition, comparing three loasing heights: low, mid and high-height relative to L4/L5. Data were collapsed within shoulder angle conditions (n=60).

4.3.4.2 Left Lumbar Erector Spinae

Increasing the height of the load did not change the activation level of the LLES significantly for the Proximal Light conditions at any Angle. In the Distal Light conditions, however, there was a significant decrease in muscle activation as the height of the load increased at an angle of 0 deg (Figure 12). In the Distal Heavy condition, the activation remained approximately the same across all Heights for the 0 and 90 degree conditions, while the activation in the 45 degree condition increased significantly when the load height was increased to High. During the Proximal Heavy condition, all 3 Angle conditions showed a significant Height effect, however, contrary to the RLES activation, levels were most similar in the 0 and 45 degree conditions for the LLES. The greatest recorded activity in LLES (21%) was recorded at the High height, in the Proximal Heavy condition, while the shoulder was laterally rotated 45 degrees.

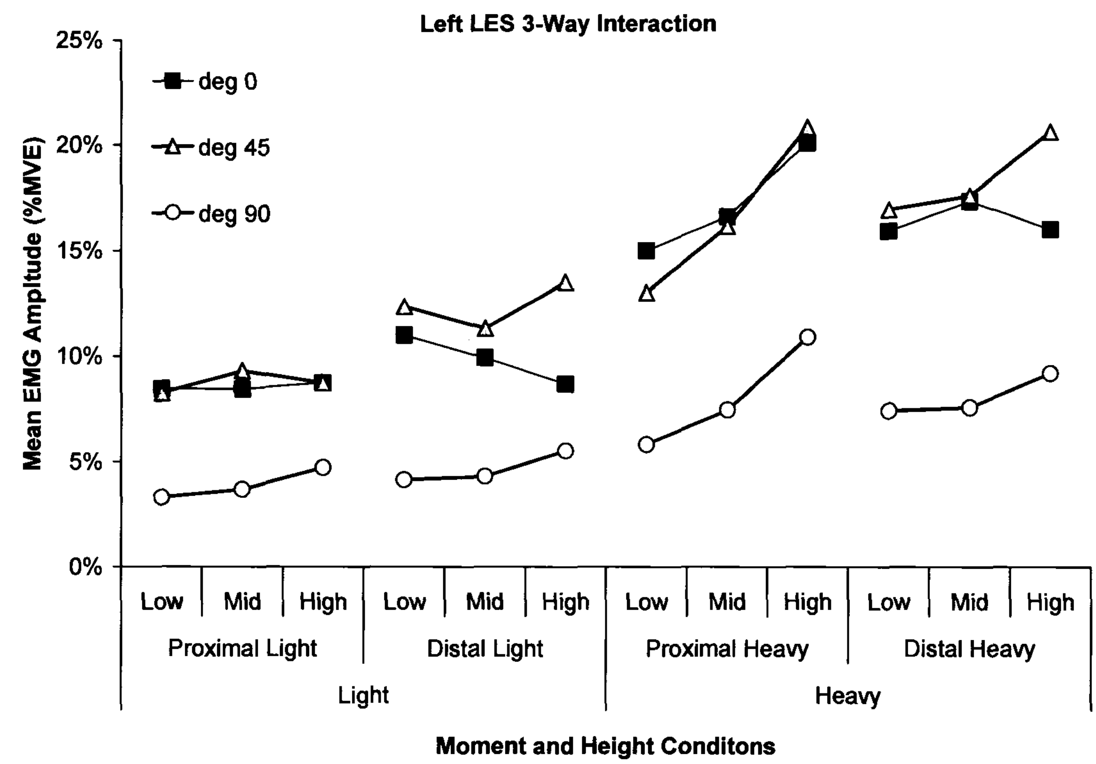

Figure 12. 3-way interaction for the LLES EMG amplitude. Mean EMG amplitude levels for each moment and angle condition, comparing three loasing heights: low, mid and high-height relative to L4/L5. Data were collapsed within shoulder angle conditions (n=60).

4.3.4.3 Right Thoracic Erector Spinae

The RTES was highly activated, second only to the LTES. A significant Height effect was observed in the RTES for each Moment condition, within each Angle condition examined (Figure 13, $p<0.05$). While the greatest activity was observed in the 0 degree conditions across all Moment conditions, the 45 degree conditions generated slightly less muscle activity, and the activation of the RTES in the 90 degree conditions was noticeably lower. The highest recorded activity of the RTES (32%) occurred at the High height, in the Proximal/Heavy condition, with the shoulder at 0 degrees. The Proximal and Distal Light conditions generated similar activation trends with increased height, which is not surprising since the external moment was the same. The same was true for the Proximal and Distal Heavy conditions.

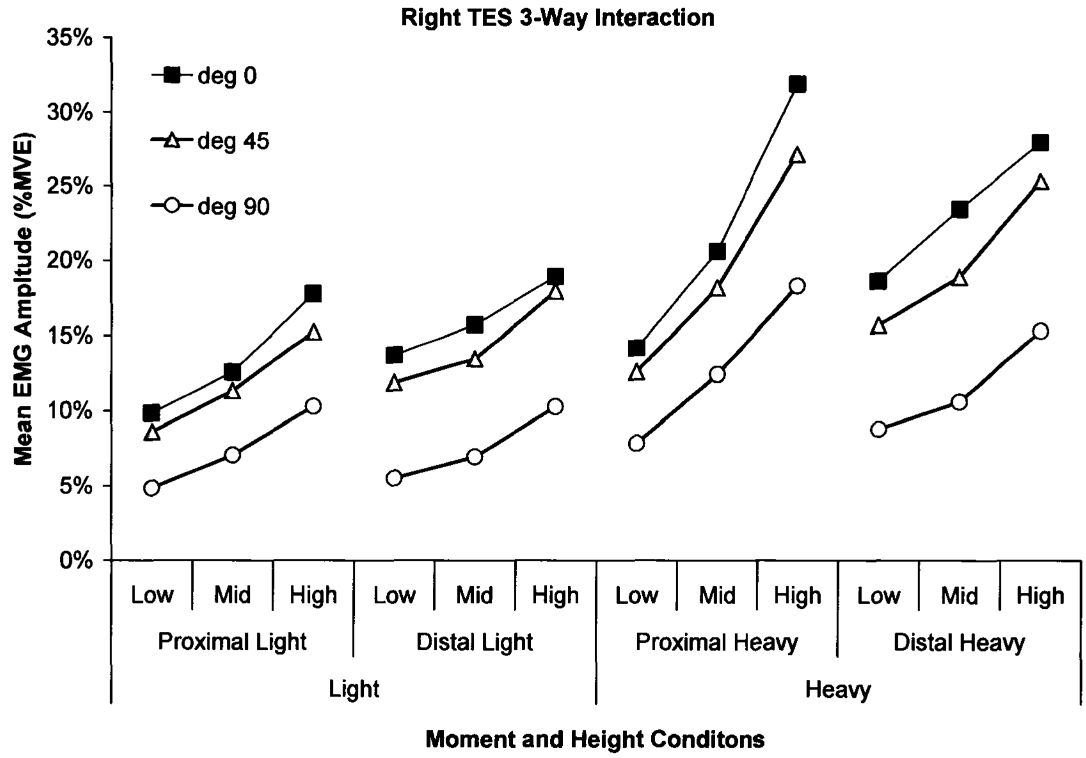

Figure 13. 3-way interaction for the RTES EMG amplitude. Mean EMG amplitude levels for each moment and angle condition, comparing three loasing heights: low, mid and high-height relative to L4/L5. Data were collapsed within shoulder angle conditions (n=60).

4.3.4.4 Left Thoracic Erector Spinae

A significant Height effect was observed in the LTES for each moment condition, within each angle condition examined (Figure 14). The LTES was the most highly activated muscle, with its maximum activation recorded at 47% MVE in the Proximal Heavy, High height condition, while the shoulder was at 0 degrees. While the greatest activity was generally observed in the 0 degree conditions across all Moment conditions, the 45 and 90 degree conditions generated only slightly less muscle activity. The Proximal Light and Heavy conditions generated similar activation trends with increased

Height, although the external moment was doubled. The same was true for the Distal Light and Heavy conditions. Thus, changing the moment arm seemed to affect the activation more so than increasing the external load.

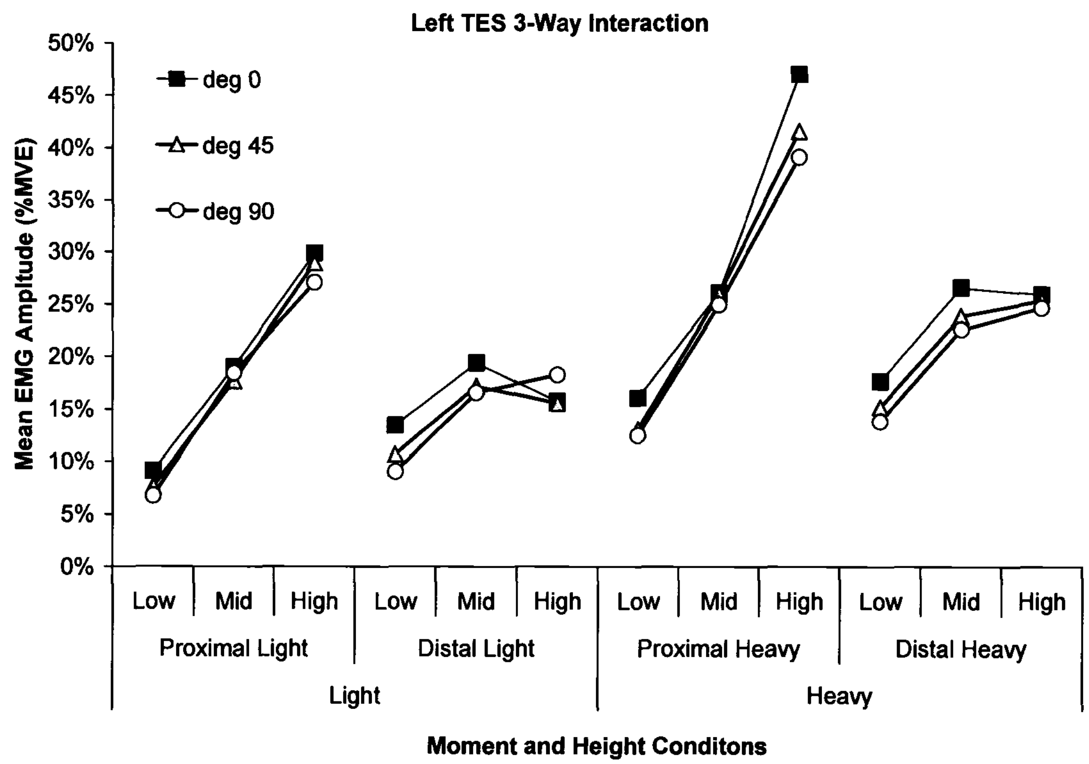

Figure 14. 3-way interaction for the LTES EMG amplitude. Mean EMG amplitude levels for each moment and angle condition, comparing three loasing heights: low, mid and high-height relative to L4/L5. Data were collapsed within shoulder angle conditions (n=60).

4.3.4.5 Right Multifidus

The EMG amplitude of the RMUL decreased significantly with increased shoulder angle across all Moment conditions (Figure 15). The greatest activation of the RMUL (15%) occurred in the Distal Light condition, at the High height, while the shoulder was at 0 degrees. The activation for the RMUL was consistently higher while the shoulder was at 0 compared to 45 and 90 degrees. This disparity is most prominent when comparing the 3 shoulder angles in the both the Distal Light and Distal Heavy conditions. In the Distal Light conditions, the activation level decreased from 12% MVE at 0 degrees, to 3% and 2% MVE at 45 and 90 degrees, respectively. In the Distal Heavy conditions, the same comparison reveals 15% of MVE at 0 degrees, 4% at 45 degrees, and 2% at 90 degrees.

No significant activation changes were detected during the 90 degree conditions as a result of increased Height. The RMUL also did not respond to height changes for the Proximal and Distal Light conditions while the shoulder was rotated 45 degrees, nor did its activation level change significantly while the shoulder was at 0 degrees, in the Proximal Light condition.

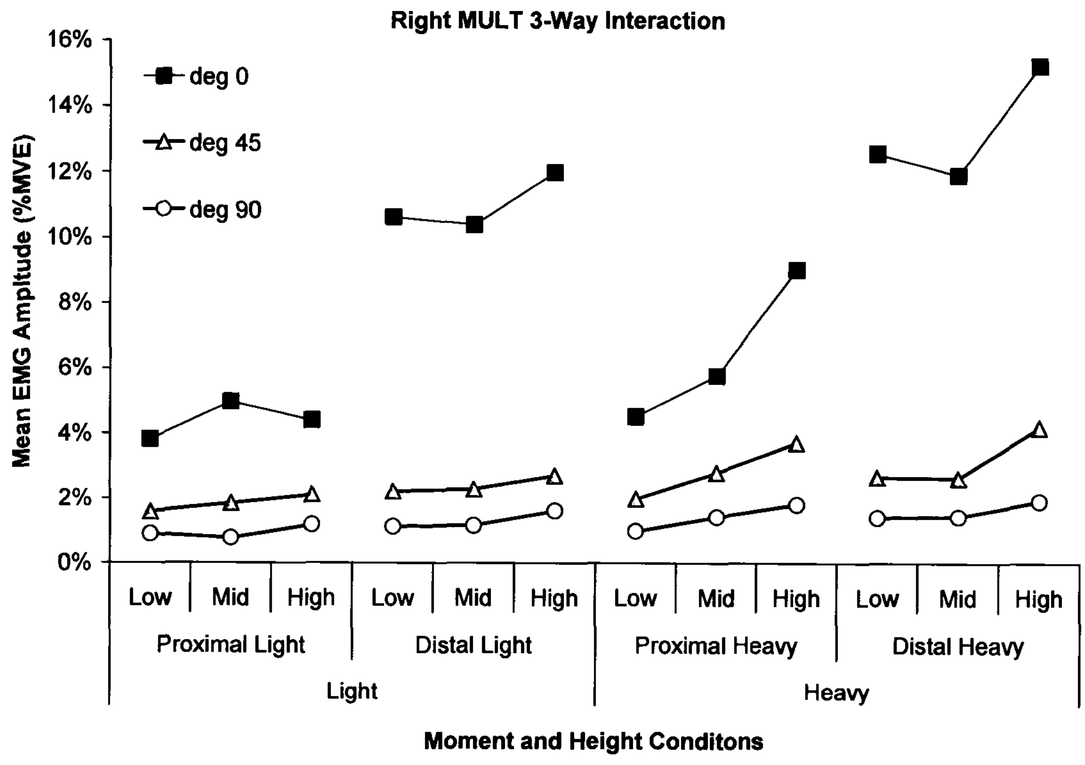

Figure 15. 3-way interaction for the RMULT EMG amplitude. Mean EMG amplitude levels for each moment and angle condition, comparing three loasing heights: low, mid and high-height relative to L4/L5. Data were collapsed within shoulder angle conditions (n=60).

4.4 Regression Model

The forward stepwise regression identified the following variables as being critical in the prediction of spinal joint loading under the current experimental settings: the height of the subject, the flexion moment acting on the subject at L4/L5, the lateral moment acting on the subject at L4/L5, and the axial moment acting on the subject at L4/L5, as well as squared and cubed values of some of these variables. The equation, resulting from the stepwise regression to predict L4/L5 compression, is as follows (n=696):

$$\text{Compression (N)} = 3326.64 + 3.81 * Ht_L + (-16.68 * BH) + 11.81 * Mo_{FLEX} + 0.68 * Mo_{FLEX}^2 + 0.56 * Mo_{LAT}^2 + (-0.06 * Mo_{AX}^3) \quad [3]$$

Where: Ht_L = height, above L4/L5, of the external load (cm)

BH = Height of the subject (cm)

Mo_{FLEX} = external flexion moment acting on the subject (Nm)

Mo_{FLEX}^2 = external flexion moment acting on the subject, squared (Nm)

Mo_{LAT}^2 = external lateral moment acting on the subject, squared (Nm)

Mo_{AX}^3 = external axial moment acting on the subject, cubed (Nm)

4.4.1 Regression Model Performance

Comparisons were made with the EMG data pooled across all subjects, and averaged across all isometric trials (n=48). The compression outputs from the third-order polynomial (McGill, Norman and Cholewicki, 1996), the Michigan 3-Dimensional Static Strength Prediction Program (3DSSPP), and the regression based polynomial were compared to the EMGAO approach (Cholewicki and McGill, 1994). The EMG-optimization model was considered to be the criterion to which each of the other three model's performance was measured.

Using the experimentally derived EMG profile of each muscle, the EMG amplitude was averaged for each loading condition, for all subjects, across all trials (n=48) and utilized as input for the EMGAO model. The relationship between the compression values derived from the EMGAO model and each of the selected models for comparison can be seen in Figure 16.

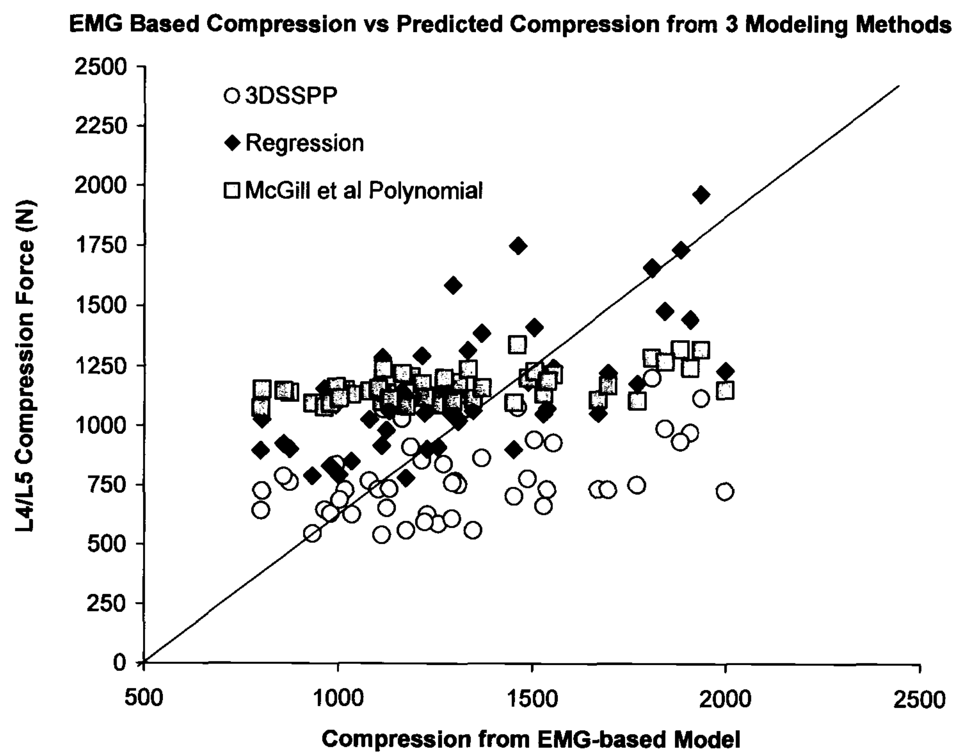

Figure 16. Comparisons were made between the compression values calculated by each of the 4 models; 3DSSPP, the polynomial of McGill et al. (1996), the regression equation developed in this study, and lastly, the EMG-optimization model (Cholewicki & McGill, 1994), which is displayed on the x axis. N=48 for all.

Overall, the current regression model was found to provide the best fit of predicted compression values with the actual data points (r^2=0.43) and also the lowest RMS error (291 N). The relationships between the actual compression (EMGAO) and compression predicted by each of the individual models, including r^2 and RMS error values, are shown in Figures 17, 18, 19.

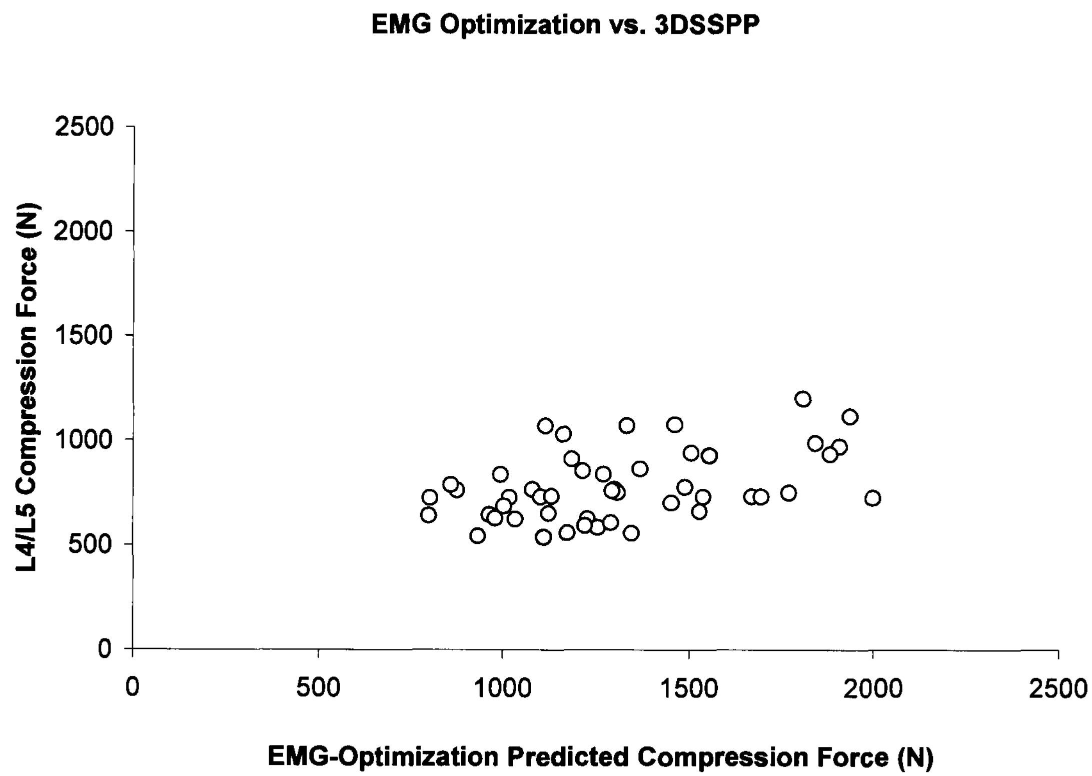

Figure 17. Modeled L4/L5 compression force predictions from 3DSSPP compared to experimentally derived compression forces calculated with the EMG-optimization technique (Cholewicki & McGill, 1994),averaged across all trials (n=48). (R^2= 0.20, RMS error = 600 N).

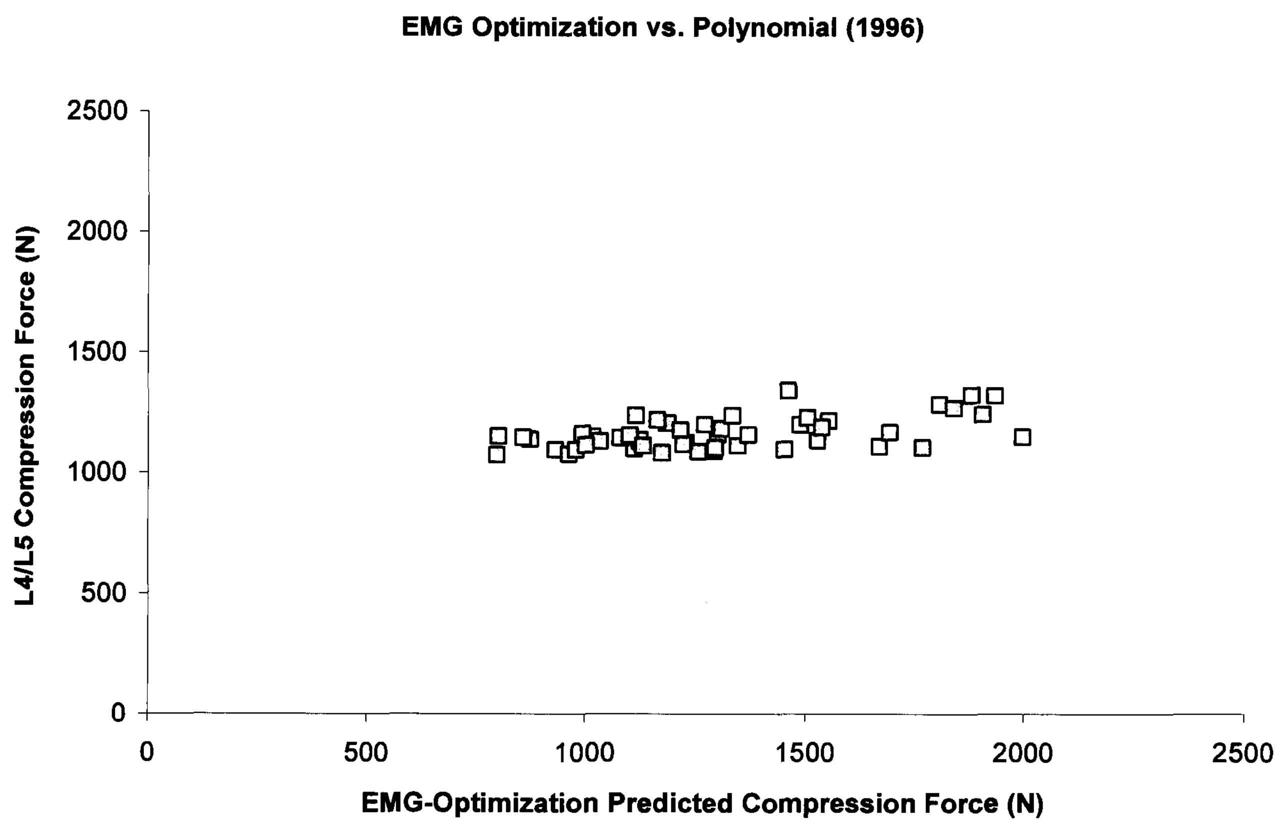

Figure 18. Modeled L5/S1 compression force predictions from the polynomial equation of McGill et. al. (1996) compared to experimentally derived compression forces calculated with the EMG-optimization technique (Cholewicki & McGill, 1994), averaged across all trials (n=48). (R^2= 0.29, RMS error = 321 N)

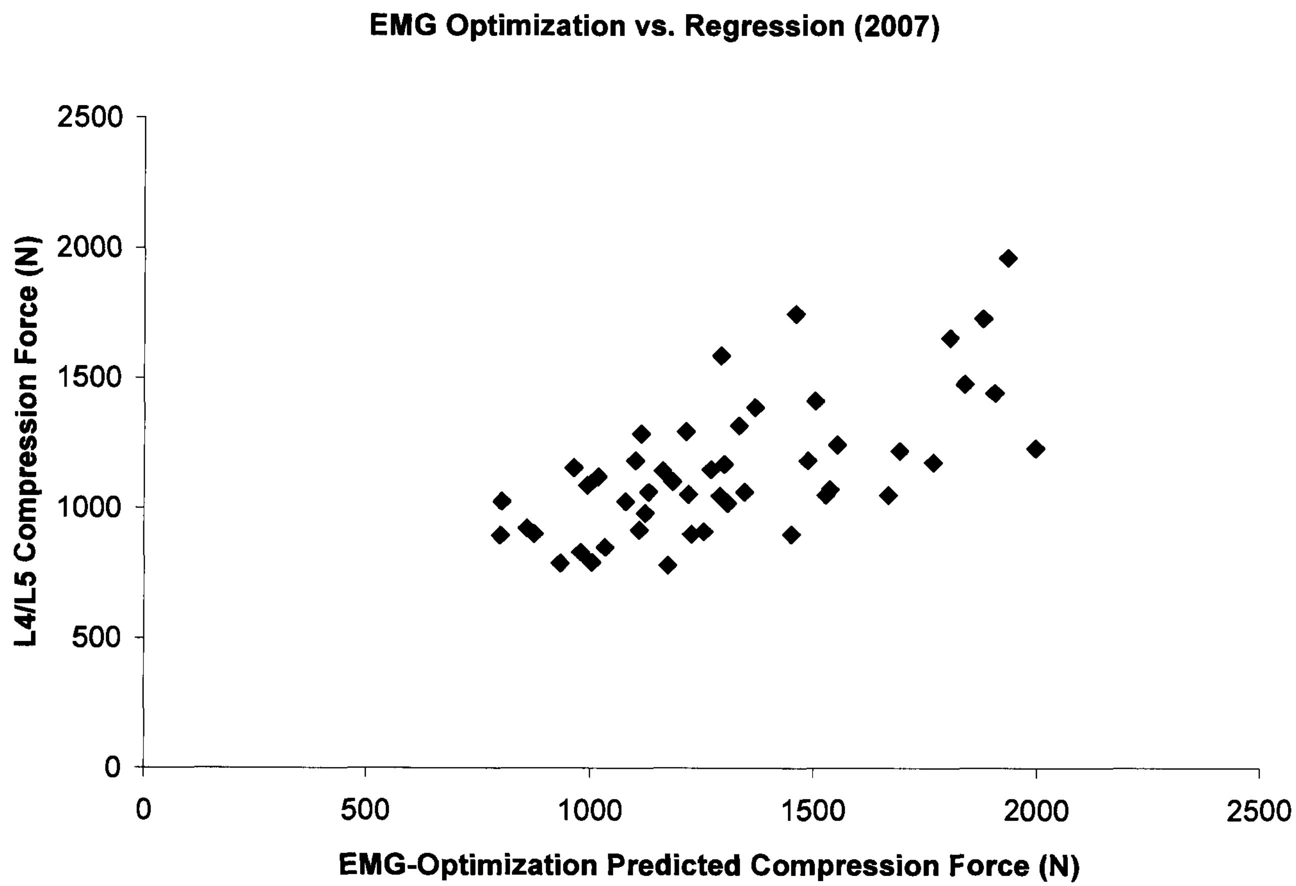

Figure 19. Modeled L5/S1 compression force predictions from the regression equation from the current study compared to experimentally derived compression forces calculated with the EMG-optimization technique (Cholewicki & McGill, 1994) averaged across all trials (n=48). (R^2= 0.43, RMS error = 291 N)

4.4.2 Regression Model versus Polynomial Equation

Further analysis of the regression model and the polynomial was performed as they provided the highest correlation with the EMGAO outputs. The averaged EMG amplitude, for all 15 muscles, was calculated across subjects for every loading condition, (n=696) and utilized as input for the EMGAO model. The EMGAO compression values were then compared to the values predicted by the current model, as well as the 3rd order McGill polynomial (Figure 20).

The compression predictions of the current regression equation were shown to have a stronger correlations with the actual compression values, and lower RMS errors ($r^2 = 0.58$, RMSe = 14 N) than the McGill polynomial equation ($r^2 = 0.30$, RMSe = 680 N) (Figure 20).

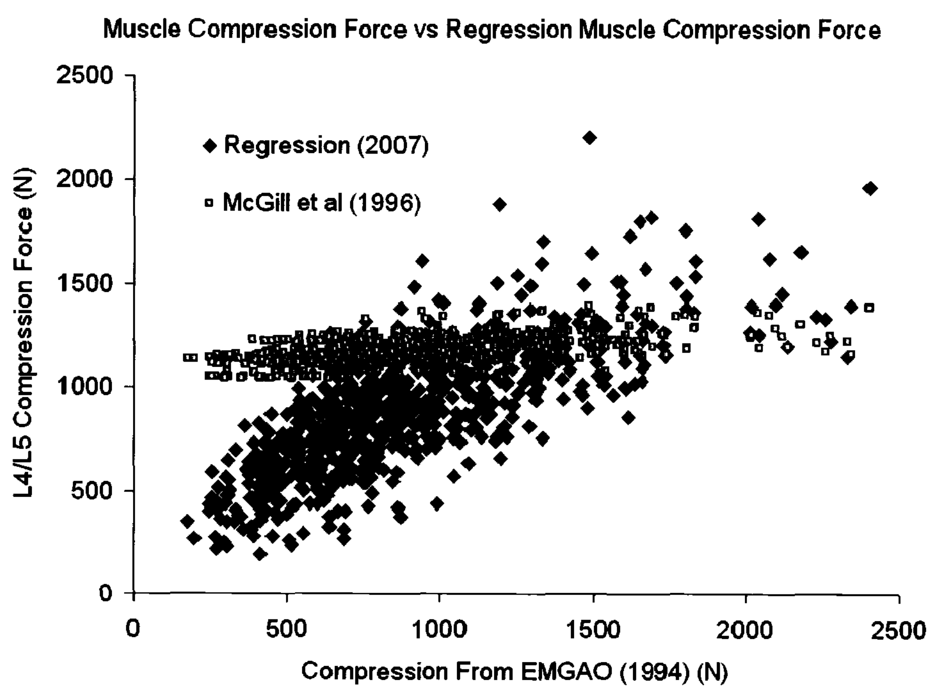

Figure 20. Relationship between muscle compression forces predicted by the EMGAO method and the muscle compression force predicted by the Regression equation from the current study (2007) ($p<0.0001$, $r^2= 0.58$) and the Polynomial equation of McGIll et al. (1996) ($p<0.0001$, $r^2= 0.30$). N=696 for both.

Chapter 5: DISCUSSION

The current regression based model is similar to that of McGill et al. (1996), which used triaxial joint moments to estimate internal muscle compression. The current model also has the additional input of external load height, or load potential energy, to its polynomial regression equation. The current equation was found to improve predictions of compressive loads at the L4/L5 level. This appears to indicate that some incorporation of load potential energy improves the prediction of internal muscle compressive forces, thus demonstrating that the height of the load may be a critical variable in determining low-back compression. This has not been accounted for in any previous method to predict compression force without EMG.

Trunk muscle EMG data were collected to allow for the criterion EMG-based estimates of lumbar joint loading. The following will discuss the electromyographic findings, as well as performance of the current polynomial regression equation for predicting force, as compared to three other, previously published, models commonly used for predicting spinal compression.

5.1 Hypothesis Revisited

1. *It is predicted that the muscle activation of the trunk will show a positive relationship with the potential energy of the hand load, while the moment demand is relatively constant. As the hand load increases in vertical location, while the horizontal moment arm stays constant, the spinal system will increase in stiffness to protect from injury through increased activation levels. A high positive correlation*

will thus be found between the potential energy of the load and the myoelectric activity recorded in this study.

Increases in the height of the external load were associated with increases in trunk muscle activation for every muscle except the LMULT.

2. *A regression model which includes the potential energy of the system in the calculation of spinal compression will show to improve upon compression predictions from current biomechanical models. Root-mean-square errors (RMSe) and correlations will be calculated between predicted forces from the models and the actual forces obtained through experimentation.*

   The inclusion of potential energy as a critical variable in the calculation of spinal compression was successful in improving L4/L5 compression predictions, when compared to existing models which do not account for the height of the external load. This indicates that it is crucial to consider the potential energy of a system when calculating spinal load as increases in trunk muscle forces clearly indicate that the CNS shows a significant response to increased loading height. These findings will be discussed further in the following sections.

5.2 Limitations of the Regression Model

It should be emphasized that the regression model presented in this study is intended only to provide proof, in principle, that the inclusion of height in modeling spinal loads can improve predictions of compression.

Like all other models that attempt to predict biological forces, such as joint compression, the current study is subject to the limitation of anatomical modeling assumptions and simplifications. The regression equation developed in this study was derived from experimental EMG during loading tasks that were static in nature, and the dynamic nature of most industrial tasks was not considered. As in any study using surface EMG electrodes, there existed a potential for crosstalk, which would result in inaccurate EMG profiles. In addition, the asymmetric hand loads utilized were relatively light and unilateral; therefore the treatments within this study do not approach the complex loading scenarios seen in the workplace. The loading scenarios may have demanded increased stability of the right shoulder, thus driving up the activation of the RLD. The study was performed with a limited number of subjects and postures, and therefore, the regression model should be tested under more complex loading conditions. In other words, the regression equation should be tested with greater load magnitudes and for tasks that are dynamic in nature. Hence, this model is not intended to be used in field studies or by practitioners. For the scope of this thesis, all three models were evaluated under the same set of averaged isometric conditions and are, therefore, subject to the same limitations. Although the current study utilized relatively light loads, and the tasks were static in nature, clear evidence has been provided that the height of the external load affects coactivation, and ultimately, compression

5.3 Height

Biomechanical models are the most common and widely accepted method for estimating spinal loading in the workplace. However, existing models do not consider loading height when calculating compression. The potential energy of the system, as a whole, readily influences the stability requirements. Granata and Orishimo (2001) indicated that muscle forces must increase with height, despite the fact that external trunk flexion moments were the same at each height. Therefore, changing the height at which one holds an external load alters the requirements to maintain spinal stability as indicated by an increase in coactivation. The present study demonstrated increases in muscle activity, with increasing height, in every muscle but the LMULT (Figure 6).

5.4 Electromyographic Findings

5.4.1 Abdominal Musculature

5.4.1.1 Rectus Abdominus

The activation of the RA was found to be functionally minimal within the scope of this thesis. The activation levels recorded were, on average, 1-2 % MVE. These findings are also consistent with previous research. In a series of single-equivalent models to estimate compression and shear forces on the lumbosacral disc during a lifting task, van Dieen and DeLooze (1999) always set the weight factor of the greatest RA activity to 0. Such behaviour has also been documented by Brown, Haumann, and Potvin (2003).

5.4.1.2 Obliques

In a study investigating the potential stabilizing role of individual muscles the obliques were found to have a critical role in maintaining stability (Kavcic et al 2004). The current study supports this finding. High height conditions (with their increased stability requirements) always demanded the highest muscle activation levels, from each of the external and internal obliques, in every moment condition, and every angle condition, when compared to the Low height conditions. The height effect was further amplified as the loading conditions progressed from Proximal Light to Distal Heavy. The right obliques were shown to have decreased activation as the load became more lateral (increasing shoulder angle), while the left obliques were shown to increase in activation with increased shoulder angle, as indicated by a 2-way interaction between shoulder angle and loading height (Figure 9).

The current EMG data suggests that the external obliques have a greater impact on stability than the internal obliques, which is contrary to the findings of Kavcic et al. (2004). The significant interaction between shoulder angle and loading height (Figure 9) indicates that at 90° the REO and RIO activation levels both increase with increasing loading height; however the response of the REO was greater. Therefore, during right lateral bend (90°), the REO and RIO act to stabilize the spine in the lateral bend axis, while REO have a greater influence in stabilizing the spine. This finding agrees with EMG data from previous research (Brown 2003).

As the external load was rotated to the right (at 90°), the increases in activation were much more extreme for the left abdominals (LEO, LIO). This was expected because an increase in the lateral moment will require a greater increase in the agonistic left obliques' activation level. These findings are consistent with those of Bean et al. (1988). The authors found that, when the held weight is shifted as little as 30° to the right of the mid-sagittal plane, the LIO and the LEO muscles increased in activity. In the current study, at 0°, the bilateral obliques had approximately the same activation level. As the weight was rotated 90° from mid-sagittal plane, both left oblique muscles became substantially more active while the right oblique activity decreased. These data are also supported by the model of Stokes and Gardner-Morse (2001), which predicted the RIO muscle activation to be greater at 0° than 90°, and the LEO to be more than the REO at 90°.

The interaction between moment condition and shoulder angle (Figure 8) indicates that, while the muscle activation levels of the LEO and LIO increase with increased angle, this effect is amplified as the moments increase from Proximal/Light to Distal/Heavy. The LIO was activated to a greater extent than the LEO, independent of shoulder angle and moment condition. However, the change in activation was greater for the LIO from 0° to 45°, yet from 45° to 90°, greater activation changes were seen in the LEO. This finding shows that, at 90°, the LEO is activated to a higher level as it is a true agonist in these conditions. Although the LEO has the potential to create very similar maximum lateral torque and flexion torque as the LIO, its slightly lower activation level is indicative that it is a secondary agonist during left lateral bending.

The moment x height interaction (Figure 10) shows that, within the Proximal Heavy and Distal Heavy conditions, where neither the moment arm, nor the hand load increased, the EMG amplitude of all oblique muscles increased. Interestingly, the REO was shown to have higher activation than the RIO. In a lateral bending task, the right abdominal muscles increase activation to oppose the left lateral bend moment, thereby supporting the neutral spine posture, but also, they potentially protect against an instantaneous instability at the level of a single lumbar joint resulting from an excessive rotation in lateral bend (Kavcic et al, 2004). Therefore, it can be said that the REO is counteracting the "excessive rotation" generated from lateral bending, in conjunction with the LIO to maintain spinal stability.

5.5 Posterior Musculature

5.5.1 Erector Spinae

5.5.1.1 Lumbar Erector Spinae

In general, the LLES was more active than the RLES, which is not surprising as the system is being asymmetrically loaded on the right side, always generating at least some left lateral bend, in which case the LLES is a primary agonist. The LES activation trends reported in this study agree with Bean et al. (1988) who found that increasing the angle of the hand held weight from 0° to 30° to the right, decreased RES activity. Significant interaction effects indicate differing activity trends in the bilateral LES (Figure 11, Figure 12). The RLES activation noticeably decreased with increasing shoulder angle, and the angle effect was most pronounced in distal conditions (Figure 11). The LLES however, was found to have similar activation patterns in both 0 and 45 degree conditions, and when the load was held at 0.5 m, the LLES was more active during 45° conditions (Figure 12).

When data were pooled across angle conditions, and within moment conditions, the observed RLES always increased activation with height increases, independent of moment or shoulder angle. This increased activity was most noticeable in the Proximal/Heavy conditions for the RLES, at 0° (Figure 11).

An interesting finding was observed with the LLES at 0°. In the Distal/Light condition, it was actually found to decrease with increasing height (Figure 12). Also at 0° and 90° in the Proximal/Heavy condition, the LLES activity showed no significant increase in activation with increased loading height. It may be that beyond critical moment arms, and above critical heights, an increase in the activation of the LLES would actually cause an extensor and a left lateral bend moment, which would in turn require increased coactivation, and ultimately would increase the loading of the L4/L5 disc.

5.5.1.2. Thoracic Erector Spinae

There was a pronounced height effect within all moment conditions, for the RTES. For the LTES, however, the height effect was more pronounced in the proximal conditions (Figure 13 & Figure 14). The angle effect observed for the RTES was consistent across all moment conditions, but was not as dramatic in the LTES. There did not appear to be a large effect for reach, as in the light and heavy conditions, the activation levels averaged across angles, and between heights, were relatively the same. In general, increasing the external load generated larger increases in muscle activation than increasing the moment arm for the bilateral TES.

Using the model of Cholewicki and McGill (1996), Brown and Potvin (2007) have indicated that the LES and TES may contribute approximately 62% of the maximum rotational stiffening potential of the spine about the flexion/extension axis of the L4/L5 joint. Previous research has identified the ES musculature as the most active group during dynamic lifting tasks (Gallagher, Marras, Davis, and Kovacs, 2002). While the TES has large moment generating potential at the L4/L5 joint, recent research has suggested that these muscles also provide the majority of the stiffness to the lumbar spine (Brown & Potvin, 2007), and have a direction-dependent stabilizing role (Kavcic et al, 2004).

The current findings show that, when the external load is rotated to the right, the RES activation level drops significantly. Bean et al. (1988) reported that at 60° of rotation from the mid-sagittal plane, the LES works with the left obliques to balance the external moment, and the RES is not needed. While the RLES and RTES both exhibited lower EMG amplitudes with increasing distances from the mid-sagittal plane (Figures Figure 11, Figure 12, & Figure 13), they remained active to stabilize the spine. Furthermore, within each moment condition, when the height of the load increased, the RES muscles were observed to increase. This shows that when left lateral bend internal moments are required, the RES muscles act as stabilizers. The fact that the experimental RTES activity levels are significantly higher than the RLES signifies that the RTES may be a better stabilizing muscle. This is likely due to the fact that the TES muscles have a longer effective moment arm than the LES muscles; hence they are better equipped biomechanically to generate forces (Brown & Potvin, 2007).

Brown and Potvin (2005) developed an optimization model which incorporates stability as a constraint when calculating spinal compression. Experimental data highlighted the TES as both the highest agonist and antagonist force producing extensor muscle, yet the model underestimated the bilateral activation of both the LES and TES. Current EMG data are consistent with muscle force predictions of Brown and Potvin (2007) which identified the TES as a stabilizer at multiple spinal levels, as well as a large moment generator. The direction-dependent stabilizing role suggested by Kavcic et al. (2004) is a suitable explanation for the high activation levels recorded in the LTES, as the asymmetrical task demands of the present study stipulate that the LTES stabilizes the spine regardless of whether it is acting as an agonist, or antagonist.

The simplified muscle equivalents in the model of Bean et al. (1988) do not take into consideration the level of the ES. The EMG data suggests that, while the angle has a clear effect on the LES, the same is not necessarily true for the TES. Generally, the bilateral TES muscles were more activated than their lumbar counterpart, which is consistent with previous research (Brown & Potvin, 2005). The activation of the LTES stayed relatively high during all 3 angle conditions as it is a primary agonist in left lateral bend conditions. Interestingly in the RTES, when the weight was rotated from 0° to 45°, the decrease in activation was not as remarkable as what was observed for the RLES.

Overall, the bilateral LES EMG amplitude was affected more by the moments that were required to be equilibrated, comprehending shoulder angle, moment arms and load weights. Comparatively speaking, EMG amplitude for the TES was more heavily influenced by the height of the external load, than by increasing the shoulder angle, moment arm, or the mass being held in the hand. This indicates that the increases in myoelectric activity associated with increased height are, in fact, a stabilizing mechanism because the effective moment arm has not increased. The present data suggests that, under the current circumstances, the LES is acting more as a moment generator, while the TES acts as a spine stabilizer.

5.5.2. Multifidus

Of the multifidi, higher EMG amplitudes were recorded from the left side, which agrees with previous research using similar loading conditions (Brown & Potvin, 2005). In all loading conditions in the present study, the LMULT was always an agonist, while the RMULT would be considered antagonistic, and a stabilizer in the 45 ° and 90° conditions.

A significant interaction was observed between height, shoulder angle, and moment condition for the RMULT. The RMULT activation level decreased with increasing shoulder angle, especially when going from 0° to 45°. The angle effect is most noticeable in the distal conditions, where the activation at 0° was much higher than the other two angles in the distal moment conditions. The height effect was most noticeable at 0° in the Proximal/Heavy condition. The RMULT was most active during the 0° conditions, when the RMULT countered the flexion moment. Since the multifidus has a small lateral moment arm, it comes as no surprise that, while the shoulder angle was at 90 degrees, height did not have a significant effect on the muscle's activity, which was activated at a low level. Activation during right lateral bending, therefore, is acting to stabilize the spine by providing resistance to the right lateral bend moment.

Cholewicki & McGill (1996) indicated that instances of instability occur when subjects were in a neutral posture, and the EMG amplitude of the MULT and the LES were negligible. There was an interaction effect between angle and moment which indicated that, when averaged across moment conditions, the LMULT was most active when the load was held at 0° or 45° of shoulder rotation, yet dropped significantly when the arm was rotated to 90°. Interestingly, the interaction between height and moment indicated the LMULT was more active in the low conditions than the high conditions. Brown and Potvin (2005) found that the inclusion of stability constraints into an optimization model predicted MULT forces to decrease. It is possible that the multifidus has the potential to destabilize the spine at higher heights, and explains why the activity levels are low. Kavcic et al. (2004) demonstrate that the multifidus is an efficient translator of its generated force to spinal stiffness and stability. Based on this finding, it may be that this propensity to be efficient, allows for the MULT to be recruited at the low levels recorded in the present study.

5.5.3. Latissimus Dorsi

Although the LD has large effective moment arms, and a large force producing capacity, its activation levels are poorly predicted in optimization models. There was a main effect of angle on the LLD, showing the greatest activity level during 45° conditions, when the effective moment arm was the largest. The LD was always activated to a higher level at the High height. EMG profiles from this study show that the RLD is more active then the LLD during asymmetrical loading tasks, with the load in the right hand. This increase in activation is a reaction to the dramatic increases in activation of the RDELT, as the LD stabilizes the shoulder joint. This finding is supported by McGill (1992) and Ng (2003).

A significant interaction between load height and shoulder angle for the RLD revealed that the height effect became less prominent as shoulder angle increased. This can be seen in Error! Reference source not found.., where it is clear that, while the RLD is more active at High heights within every angle condition, it is activated to a lesser degree at 90°. This finding is not surprising, since the RLD acts as an antagonist to the RDELT, which was also shown to decrease with increasing shoulder angle, and will be discussed later.

Kavcic et al. (2004) reported that the LD is inefficient at translating its generated force into joint stiffness and stability. In muscle modeling, nodal points are used to recreate the curvilinear lines of actions of muscles. These nodes can serve to increase the effective moment arm of a muscle, and thereby affect its stiffening capacity of a muscle (Brown & Potvin, 2007). The stabilizing capacity of a muscle is a function of its origin and insertion points, nodal locations, length, moment arm and force (Potvin & Brown, 2005). The LD crosses 2 joints, thus it has nodal points which invariably change the line of action. This, in part, may explain why the RLD is so much more active at 0°. The ideal stabilizing muscle would have a short length, and a long moment arm (Brown & Potvin, 2007). The RLD originates on the lower level of the spine, and crosses the shoulder joint to insert on the humerus. Thus, when the load is held at 0°, the RLD is subject to non-linear lines of action, as it must "bend" around the shoulder joint. This geometric orientation, or change in direction of the RLD fibers increases its effective moment arm, and improves its ability to stiffen the joint, and therefore may be preferentially recruited for the purposes of enhanced joint stability when the load is being supported close to the mid-sagittal plane. As the shoulder angle is increased to 90°, the RLD is no longer required to change direction, hence the moment arm is shorter, thereby rendering the RLD less capable of stabilizing the joint; and lower activation levels are observed.

A significant interaction between load height and moment condition shows that, when EMG data were pooled across shoulder angles, the height effect was most pronounced in the Proximal Heavy conditions for the bilateral LD (Figure 10). The bilateral LD was subject to greater increases in activation when the load was increased than when the moment arm was increased. In the distal conditions, there was a greater increase in activation level for the RLD when going from low to high height, while the LLD was observed to increase more when going from the mid to high height. Again, this can be attributed to the RLD reacting to satisfy the equilibrium constraints of the shoulder in conjunction with the increase RDELT activation.

5.5.4 Deltoid

A significant main effect of angle revealed that the RDELT activation decreased with increasing shoulder angle (Figure 7). This is primarily due to the fact that the EMG recordings were taken from the anterior portion of the RDELT. The anterior fibres of the RDELT act to draw the arm forwards, and do not generate a substantial lateral moment, which would be required to stabilize the shoulder at 90 degrees.

RDELT EMG data revealed a significant moment x height interaction (Figure 10) and was activated to the highest level at the High height. The height effect was amplified in the Proximal/Heavy and Distal/Heavy conditions indicating, once again, that despite the fact that the external moment was relatively constant, the contractile forces increased with increasing height. The deltoid is the primary antagonist of the LD in the stabilization of the shoulder joint and, therefore, the large increase in RDELT activation to raise the external load to the highest heights was associated with large increases in RLD activity. The RDELT and RLD then, work together to stabilize the shoulder joint.

5.6 Model Performance

Previous ergonomic methods utilized for predicting spinal compression use estimations of muscle forces that are based solely on moments. Research has shown that controlling the moment arm, the load magnitude or the anatomical model will have an effect on the predicted compression (Cholewicki et al, 2000; Cholewicki & VanVliet, 2002). Theoretically, these models would not show an increase in predicted muscle forces if the effective moment arm remained constant, and the load height increased. Research has shown that the motor control system responds to stability requirements in addition to moment equilibrium requirements (Cholewicki & McGill, 1996; Gardner-

Morse et al, 1995). Two existing optimization models use stability as a criterion when predicting compression (Stokes & Gardner-Morse, 2001; Potvin & Brown, 2005), however, these models do not isolate load height as a variable which may affect the prediction of compressive forces in the lumbar spine. Despite the limitations of this study, the results agree with previous research which states that the CNS responds to changes in the biomechanical spinal stability of the trunk (Granata & Orishimo, 2001). By independently influencing the potential energy of the system and trunk moment, this study shows that the inclusion of load height may improve predictions of spinal load predictions.

The experimental trials were designed so that the external moment would be similar for two conditions, at two separate moment arms. This allowed the potential energy of the system to be altered while the load moment was controlled. In this manner, the external moment acting on the subject would not change. Thus, prospective muscle activation pattern changes would implicate a modified stability scenario to which the motor control system was responding.

The current regression-based model was found to correlate very well with the measured myoelectric activity. This relationship was established by comparing the output from the various predictive methods with the compression predicted by the EMGAO method (Cholewicki & McGill, 1994). Comparisons were made between the compression values calculated by each of the four methods; 3DSSPP, the polynomial of McGill et al. (1996), the regression equation developed in this study, and lastly, the EMG-optimization model (Cholewicki & McGill, 1994).

The model that produced predicted compression values with the closest agreement to the experimental data, was the regression-based polynomial model of the present study (RMS error = 291N, n=48). The third order polynomial of McGill et al. (1996) was found to have an RMS error of 321 N, and the posture prediction software 3DSSPP had an even larger RMS error of 600N for compression predictions.

The regression-based polynomial derived from this study and the third order polynomial method (McGill et al, 1996) had average RMS differences with the EMGAO method of 14.5 N and 680.4 N, respectively. Across all trials (n=696), when compared to the EMGAO method, the polynomial equation presented in this study displayed a strong positive correlation (r^2 of 0.58), while the third order polynomial of McGill et al. (1996) displayed a weaker positive correlation (r^2 of 0.30). The RMS errors were found to be the lowest at Low heights and highest at the High heights for the prediction of compressive forces for all methods. More specifically, in terms of compression, the RMS error was 188.3 N with the regression equation as compared to 279.0 N with the 3rd order polynomial. This is not surprising as lower heights have a lower stability demand than high heights. Therefore, there less antagonistic activity and, ultimately, lower absolute errors. As can be seen in Figure 20., the compression predictions from the polynomial were relatively constant, while the regression equation outputs were observed to better fit the actual compression data. This finding shows that the polynomial equation of McGill et al. (1996) is not sensitive to low loads, which is intuitively expected as the intercept value is 1067.6 N. The polynomial was based on slow dynamic movements and, as such, the intended usage of this model is for dynamic moments.

The main finding of this study was that as the load height increased, while the externally applied moment stayed relatively constant, trunk muscle activation increased as well, with the exception of the LMULT. The most reasonable explanation for this phenomenon is that increased load heights actually decreased the stability of the spinal system and, as such, the CNS responds to the destabilizing effect by increasing the activation of the trunk musculature. These findings agree with those of Granata and Orishimo (2001), who found that increasing the load height resulted in increases in the activation level of both agonist, and antagonist muscles of the trunk.

5.7 Implications for Stability Modeling

Current biomechanical models of the spine do not account for the inherent variability that exists between individuals. Even the best possible optimization model could not predict the outliers seen in Figure 16. The only way a model would be able to predict such levels would be if actual EMG signals were recorded for each trial. However, this is obviously not practical in an industrial setting.

With the current model, average RMS errors were shown to be lower than those of other popular models used to predict spinal compression forces in industry. This was because of the novel inclusion of loading height as a variable in the calculation. The significant increases in antagonist and antagonistic trunk muscle activity observed when load height was increased and the external moment remained relatively unchanged. This indicates that the CNS is responsive to factors other than moment arm and load magnitude when generating the optimal activation pattern in order to satisfy the 6 equations of equilibrium in isometric conditions. Therefore, incorporating loading height into the predictive measures of spinal loading promotes a more realistic representation of actual spine compression.

During the more complex loading scenarios, such as the triaxial moment conditions of this study, the role of the muscles acting to stabilize the spine becomes increasingly obscure. It is possible that the role of any given muscle is twofold; 1) generate a restorative moment (agonist), and 2) stabilize the joint. In the course of this study, particular muscles were shown to be affected more than others by the increases in loading height. The RLD, RTES and REO have been identified to generally act antagonists/stabilizers and they were found to be the most sensitive to increases in height from Low to Mid. Further, when the load height increased from Mid to High, the greatest relative increases in activation were observed in the REO, RIO an RLD, which also were identified to primarily act as stabilizers. When applying relatively constant external moment, and varying the height of the load, measured EMG activity in the monitored trunk muscles was observed to increase with the height of the external load. This was true regardless of whether the muscles were acting as agonists or antagonists. These results demonstrate that the CNS responds to modified stability levels of the spine.

5.8 Conclusions

The purpose of this thesis was, simply, to test the hypothesis that increasing the height of the external load would decrease the stability of the system, and influences the recruitment patterns in trunk muscles, such that incorporating loading height into predictive models of spine compression would enhance a model's realism and accuracy.

Spinal stability is partially dependent on the height of the external load, as increases in loading height renders the system less stable and, subsequently, necessitates increased muscle forces to maintain some minimum level of stability for safety. The analyses reported here give insights into the way the CNS responds to stability disruptions, specifically loading height. The effect of height demonstrates that the neuromuscular control of stability is facilitated by increased activation of the trunk muscles when the loading height is increased. The strongest argument for the current theory is presented by the significant increases in antagonistic muscle activation, and estimated force, when the load height was increased. Statistically, significant increases in muscle activation were demonstrated when the load was elevated, while the trunk moment was held constant in agonist and antagonist muscles alike.

In the course of this study, particular muscles were demonstrated to be affected more than others by the loading task parameters which affected the stability of the system. A given muscle's activation response was found to be sensitive to increases in shoulder angle, moment arm and/or load magnitude, which clearly demonstrates the complex nature of recruitment patterns in the trunk. The abdominal musculature was particularly responsive to increases in loading height. This was also true for the muscles that stabilize the shoulder, as stability of the system is not limited to the spine. The MULT demonstrated an interesting non-response to the increased height, as it demonstrated very little response to the height effect. The LES and the TES were the most active of all the muscles. These muscles showed significant responses in activation to increased loading height and, hence, decreases in stability, reaffirming their critical role in stabilizing the spine. The current model, by incorporating the loading height, significantly improves the prediction of compressive forces acting on the spine. This is true for both isometric vertical loading tasks, and isometric triaxial tasks.

The current study has demonstrated in principle that, when predicting spinal compression, despite limitations, the inclusion of the external hand load height (or potential energy) as an input variable significantly improves the prediction of spinal compressive forces.

5.9 Recommendations for Future Research

Further work is necessary in the field of spinal stability; specifically in the modeling of stability of the spinal system. Models should be developed to encompass the entire spine as a system and include the effects of intra-abdominal pressure, the thoracolumbar fascia, and the transversus abdominis. This will provide a more realistic representation of compression in the human lumbar spine. These models should ultimately be validated with in vivo measurements of compression, rather than comparing the output to that of similar models.

Additional research is needed to test trunk muscles under systematically varied levels of stability, in which external moments are applied to more than one axis simultaneously. In the current study, this was done but with only 12 of 48 sets of loading conditions. The current model should be tested under more complex loading conditions using loads that are more characteristic of those seen in the industrial setting, and also loading scenarios that are dynamic in nature. Although it was intended that the loads incorporated into the current experimental design did not produce compression loads in the range of the NIOSH limit (Waters et al, 1993), a great deal of insight into neuromotor recruitment of muscle activity, spinal stability and lifting biomechanics could be gained through the investigation of this model with a range of heavier loads.

Further investigation is warranted to explore the feasibility of incorporating the effect of the potential energy of a load into existing ergonomic analyses software programs used for calculating joint compression forces. This could potentially be achieved with a similar regression-based equation that would be used as a supplemental tool to the software.

All of the aforementioned advances in spinal modeling would unquestionably provide invaluable insight into the nature of spine instability and buckling, and discern those loading parameters that are most likely to cause tissue damage in the low back.

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APPENDIX A

CONSENT TO PARTICIPATE IN RESEARCH

Title of Study: A Biomechanical Examination of Calculating Spinal Compression: Towards a New Polynomial Approach

You are asked to participate in a research study conducted by Christy Calder and Dr. Jim Potvin, from the Department of Kinesiology at the University of Windsor. The results of which, will contribute to a masters thesis. This research is sponsored by the The Centre of Research Expertise for the Prevention of Musculoskeletal Disorder (CRE-MSD).

If you have any questions or concerns about the research, please feel to contact Christy Calder: *253-3000 ext. 2468* or Dr. Jim Potvin: *253-3000 ext. 2461* .

PURPOSE OF THE STUDY

The purpose of this study is to develop a regression equation that is able to predict spinal disc compression at the L4/L5 level. In evaluation of the current biomechanical models used for this purpose, it has been determined that theses models have limited application due to their deficiencies. This study will use biomechanical methods to evaluate disc compression during various postures. Utilizing electromyography, and biomechanical models, joint compression will be estimated for 1-handed static vertical loading tasks

PROCEDURES

If you volunteer to participate in this study, we would ask you to do the following things:

Subjects will perform maximum voluntary exertions (MVEs) of their back and abdominal musculature prior to data collection. This will include performing back extensions and abdonmial crunches against an external resistance.

Participants will stand in front of a pelvic brace in order to maintain their position relative to their hand location. Subjects will then be asked to hold 2 separate low-risk loads in their dominant hand at 3 load-heights, 3 shoulder angles, and at 2 reach distances. Additionally, subjects will be asked to perform exertions that simulate muscle activation patterns that would occur during an asymmetrical loading task. Each effort will last for 3-5 seconds. Subjects will perform three trials within approximately 2 minutes for each posture. They will be given adequate rest between postures.

Subjects will be tested in 48 conditions in one session.

POTENTIAL RISKS AND DISCOMFORTS

Subjects may experience some muscular fatigue in the back, shoulder and or upper arm. Subjects will be given adequate rest periods however, and this should help to minimize muscle fatigue.

Should a subject experience abnormal amounts of pain in the back, shoulder, or upper arm, the experiment will be terminated immediately and may complete the testing on another day.

POTENTIAL BENEFITS TO SUBJECTS AND/OR TO SOCIETY

Benefits of participating in the study would be to experience first hand some of the methods and procedures used in conducting biomechanical research. It is likely that a number of the younger subjects will be Kinesiology students, currently learning about biomechanical methodologies in there undergraduate and graduate courses.

The results of this study can be used by industrial ergonomists to determine spianl compression under different loading conditions. This information is invaluable when designing new jobs or rebalancing existing jobs. The results can be directly applied to the design stages of job task design where lifting is a risk factor, or low back pain is an issue.

PAYMENT FOR PARTICIPATION

Participation in this study is voluntary. Subjects will receive no monetary compensation for participation.

CONFIDENTIALITY

Any information that is obtained in connection with this study and that can be identified with you will remain confidential and will be disclosed only with your permission.

Subjects will be identified by alphanumerical code, not by name. Any digital records (photos, videos) of the subjects will have their face blanked out to ensure that they are not identifyable. All digital files will be stored securly by the researcher. No others will have access to them.

PARTICIPATION AND WITHDRAWAL

You can choose whether to be in this study or not. If you volunteer to be in this study, you may withdraw at any time without consequences of any kind. You may also refuse to answer any questions you don't want to answer and still remain in the study. The investigator may withdraw you from this research if circumstances arise which warrant doing so.

FEEDBACK OF THE RESULTS OF THIS STUDY TO THE SUBJECTS

Upon completion of the study a summary page will be produced summarizing the findings and industry implications. This summary page will be mailed to each participant.

RIGHTS OF RESEARCH SUBJECTS

You may withdraw your consent at any time and discontinue participation without penalty. This study has been reviewed and received ethics clearance through the University of Windsor Research Ethics Board. If you have questions regarding your rights as a research subject, contact:

Research Ethics Coordinator
University of Windsor
Windsor, Ontario
N9B 3P4

Telephone: 519-253-3000, ext. 3916
E-mail: lbunn@uwindsor.ca

SIGNATURE OF RESEARCH SUBJECT/LEGAL REPRESENTATIVE

I understand the information provided for the study *A Biomechanical Examination of Calculating Spinal Compression: Towards a New Polynomial Approach* as described herein. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

____
Name of Subject

______________________________ _______________
Signature of Subject Date

SIGNATURE OF INVESTIGATOR

These are the terms under which I will conduct research.

Signature of Investigator ____________________ Date ____________

APPENDIX B

CONSENT FOR AUDIO/VIDEO TAPING

Title of the Project: **A Biomechanical Examination of Calculating Spinal Compression: Towards a New Polynomial Approach**

Research Subject's Name: ________________________________

ID# ________________

Birth date: ________________

I consent to the audio/video-taping of interviews, procedures, or treatment included in this study.

I understand these are voluntary procedures and that I am free to withdraw at any time by requesting that either the taping be stopped or the viewing be discontinued. I also understand that my name will not be revealed to anyone and that taping and viewing will be kept confidential. Tapes are filed by number only and store in a locked cabinet.

I understand that confidentiality will be respected and the viewing of materials will be for professional use only.

____________________________ ____________________
(Signature of Research Subject) (Date)

APPENDIX C

Table 5. Standard Errors (N) for individual muscle forces for the 3-way interaction Height*Shoulder Angle*Moment presented as a percentage of MVE (n= 15)

Shoulder Angle	Moment Condition	Height	Muscle				
			RLES	LLES	RTES	LTES	RMULT
0°	Proximal/Light	Low	0.630%	1.028%	1.599%	1.288%	0.896%
		Mid	0.586%	1.158%	2.707%	1.595%	0.855%
		High	1.861%	2.204%	5.563%	4.334%	1.844%
	Distal/Light	Low	0.960%	0.990%	2.197%	1.617%	1.315%
		Mid	0.922%	1.075%	2.855%	1.674%	1.182%
		High	1.674%	2.067%	3.908%	3.778%	2.142%
	Proximal/Heavy	Low	0.507%	1.409%	2.586%	2.036%	0.843%
		Mid	0.676%	1.499%	3.511%	2.570%	0.882%
		High	0.939%	1.072%	2.472%	1.963%	1.207%
	Distal/Heavy	Low	1.410%	1.123%	2.494%	2.183%	1.881%
		Mid	1.042%	1.042%	3.390%	2.751%	1.617%
		High	0.231%	0.231%	4.018%	2.135%	0.514%
45°	Proximal/Light	Low	0.153%	1.097%	1.353%	1.429%	0.383%
		Mid	0.187%	1.072%	2.368%	1.664%	0.463%
		High	0.231%	1.042%	4.018%	2.135%	0.514%
	Distal/Light	Low	0.280%	1.065%	1.928%	1.511%	0.443%
		Mid	0.326%	1.178%	2.655%	2.099%	0.472%
		High	0.268%	1.329%	2.930%	2.540%	0.528%
	Proximal/Heavy	Low	0.318%	1.263%	1.892%	1.939%	0.414%
		Mid	0.590%	1.817%	3.625%	2.831%	0.696%
		High	1.224%	2.412%	4.972%	4.115%	0.889%
	Distal/Heavy	Low	0.318%	1.587%	2.331%	1.889%	0.480%
		Mid	0.342%	0.342%	3.231%	2.686%	0.440%
		High	0.666%	0.666%	4.408%	3.768%	1.086%
90°	Proximal/Light	Low	0.132%	0.555%	1.267%	0.752%	0.241%
		Mid	0.125%	0.539%	2.425%	1.097%	0.205%
		High	0.201%	0.598%	3.629%	1.630%	0.329%
	Distal/Light	Low	0.115%	0.401%	1.769%	0.843%	0.279%
		Mid	0.119%	0.457%	2.624%	1.005%	0.252%
		High	0.168%	0.707%	3.429%	1.655%	0.411%
	Proximal/Heavy	Low	0.148%	0.694%	1.795%	1.238%	0.196%
		Mid	0.193%	0.914%	3.253%	2.036%	0.368%
		High	0.439%	1.360%	5.048%	2.864%	0.418%
	Distal/Heavy	Low	0.185%	1.008%	2.281%	1.356%	0.427%
		Mid	0.179%	0.179%	3.079%	1.605%	0.313%
		High	0.254%	0.254%	3.967%	2.327%	0.424%

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