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# The Passive Load-Bearing Capacity of the Human Lumbar Spine in the Neutral Standing Posture

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Master of Science

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ABSTRACT

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The human lumbar spine has been shown to support compressive loads of 1000 N in standing and walking, and up to many thousands of Newtons in strenuous activities such as lifting. The literature presents a number of biomechanical models that seek to replicate the load-carrying capacity of the spine while adhering to physiological constraints. While many of these models provide invaluable insights into the mechanisms governing spinal stability, there is a nearly universal disregard for the magnitude of the muscle forces required in the neutral standing posture. In compliance with constraints on metabolic cost and muscle fatigue, muscle activations in excess of 5% maximum voluntary contraction (MVC) in the standing posture are physiologically infeasible. The purpose of this thesis was to investigate the hypothesis that the passive structures of the lumbar spine are sufficient to produce static equilibrium under the body weight load in the neutral standing posture.

A novel method of applying physiologic loads to the lumbar spine *in vitro* to determine its passive stability was developed. Five cadaver specimens were tested and a passive equilibrium posture was discovered for each. Further, the parameters defining the equilibrium posture correlate well with the standing posture as reported in the literature. This is an indication that the lumbar spine is inherently capable of remaining erect in the neutral posture with muscle activations below 5% MVC. It is postulated that the iliolumbar ligament and the thoracolumbar fascia, passive components that are not typically incorporated into stability models of the spine, have the potential to provide added passive stabilization to the system. It is recommended that biomechanical models of the spine incorporate this 5% MVC constraint and place greater emphasis on the contributions of passive structures to overall stability.

Keywords: Shaun Jeffs, spine, biomechanics, stability, lumbar, cadaver

ACKNOWLEDGMENTS

My journey through the academic experience has swept me far beyond the isles of my preconceived notions and has deposited me at the head of a path that was not anticipated. I have discovered a new passion that, unbeknownst to myself, had lain dormant for years, waiting only for a spark to ignite the flames of curiosity. My passion for biomechanics would have remained in obscurity had it not been for the inspired guidance of Dr. Anton Bowden, to whom I am greatly indebted for setting the stage for the rest of my career. This momentous transition in my life would not have been possible without the selfless mentorship of Dr. Robert Todd, who freely allowed me to pursue a new-found direction. To my committee members, Dr. Larry Howell and Dr. Brian Jensen, I not only owe gratitude for guidance in my graduate research, but for making significant and important contributions to my education. Portions of this research were funded by the National Science Foundation, # CMMI-0952758.

I would like to thank my BABEL lab mates who have always helped me keep the proper perspective on life, even in the most difficult of times. Our laughter has made this experience one worth remembering. Special thanks to Cesare Jenkins and Douglas Wright who have provided invaluable technical input and support throughout this project. Finally, I would like to thank my wife who, in my virtual absence over the past two years, has been nothing but understanding and positive. She has provided purpose and direction to my life that has proved an invaluable support throughout my schooling.

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# 1 INTRODUCTION

## 1.1 The Prevalence of Back Pain

The human spine is the core structure of the axial skeleton. It is an important load-bearing component that experiences compressive forces of several thousands of Newtons [1] in activities of daily living while maintaining a great range of mobility that is unparalleled by man-made structures. While impressive in its capabilities, the human spine is nevertheless susceptible to structural degradation and instability that might result from pathological issues such as disc degeneration. Damage to the spinal structure, whether suffered by pathological disorders or by injury, can affect its static and kinematic response to the loading that is imposed on it by gravity and the surrounding musculature. Consequently, patients suffering from the effects of deterioration of the spine often experience severe pain and weakness that can, in many cases, become completely debilitating. Many patients experiencing back pain are unable to function in their normal capacity and will never return to the workforce. The socioeconomic strain of back pain and its effects is enormous. It has been estimated that the total cost of low back pain related issues exceeds $100 billion in the United States alone [2]. Much of these costs are indirect, as back pain is the most expensive cause of work-related disability [3]. These statistics alone are reason enough to engage in a focused investigation into ways back pain can be more accurately characterized, diagnosed, and treated in an effort to not only reduce economic burden but to improve the quality of life of those who are suffering daily from it.

## 1.2 Stability and Equilibrium of the Spine

This work will focus on the stability of the lumbar spine in the neutral (standing) posture and its implications for clinical diagnosis and treatment of spinal disorders. It is important to identify the meaning of the term “stability” as used in this work. There are typically two interpretations of stability of the spine: clinical stability and mechanical stability.

Clinical stability has been defined as the “ability of the spine, under physiologic loads, to limit patterns of displacement so as not to damage or irritate the spinal cord or nerve roots and, in addition, to prevent incapacitating deformity or pain due to structural changes [4].” In short, clinical stability is concerned with the protection of the spinal nerves and pain avoidance. Damage to an intervertebral disc (IVD) or to a spinal ligament can cause clinical instability in varying degrees depending on the amount of damage. Many pathological disorders that are associated with kinematic or kinetic changes in spinal behavior (e.g., spondylolisthesis, disc degeneration, vertebral fracture) cause clinical instability.

Mechanical stability is a more concrete concept than clinical stability. It has been defined as “the ability of a loaded structure to maintain static equilibrium even at (small) fluctuations around the equilibrium position. If stability does not prevail, an arbitrarily small change of the position is sufficient to cause ‘collapse’, i.e. the structure moves further away from equilibrium [5].” This definition allows the spine to be viewed as a structural column that will exhibit characteristics of Eulerian buckling if moved away from the stable position. It also denotes a relationship to equilibrium. Mechanically (and mathematically) there are two types of equilibrium: stable and unstable. An unstable equilibrium can be likened to a ball positioned on the top of a hill (see Figure 1-1a). The ball is in static equilibrium and will remain so until perturbed away from its stable position. A stable equilibrium (Figure 1-1b) can be viewed as a ball at the bottom of a hill. After any finite deviation, the ball will seek to return to its minimum energy state at the bottom of the hill.

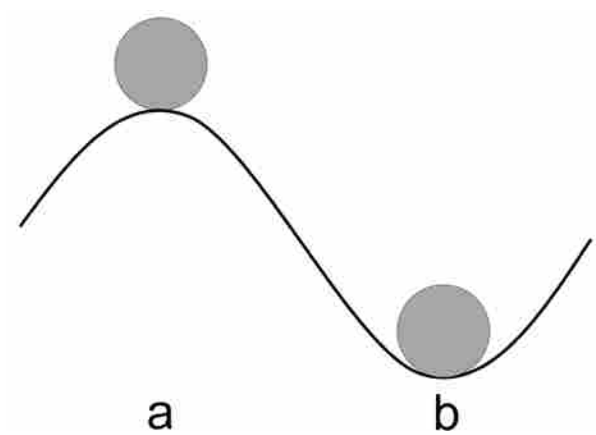


**Figure 1-1: Unstable Equilibrium (a) and Stable Equilibrium (b)**

At an unstable equilibrium, if the slope of the hill is flattened, the ball will be more tolerant to very small perturbations around the equilibrium (see Figure 1-2). Though the ball on the broader hill is still considered an unstable equilibrium, this flattening of the slope can be seen as an increase in relative stability. In this study the usage of the word "stability" will refer to mechanical stability unless otherwise specified. This thesis will seek equilibrium configurations of the lumbar spine and will attempt to quantify their stability.

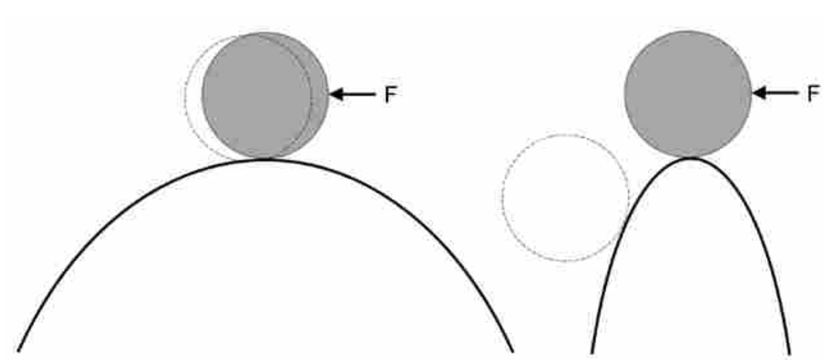


**Figure 1-2: The Relative Stability of a Ball Determined by the Slope of the Hill**

### 1.3 Modeling Techniques and Physiological Constraints

Gaining an understanding of the unique way in which the spine manages the loads imposed on it has the potential to affect the way we approach implant design and implementation, physical therapy procedures, ergonomic design, and diagnosis and treatment of pathological spinal disorders [6-8]. This understanding is dependent upon the monumental task of accurately characterizing the way the physiologic load is distributed among the active components (muscles) and passive components (ligaments, intervertebral discs, vertebrae) of the spine. The unique and complex structure of the spine presents a number of obstacles when approached as a biomechanical system. The majority of the passive structures of the spine exhibit material characteristics that are challenging to characterize and analyze: they are viscoelastic, nonlinear, anisotropic, and inhomogeneous. The difficulty in characterizing the precise response of these materials under load is a huge obstacle that impacts analytical and finite element models. Additionally, due to the structure of the musculature and the high degree of mobility attributed to the intervertebral joints, the spine is a highly indeterminate system (statically and kinematically). As Aspden summarized [9], "This indeterminacy means that for any given posture the spine has many ways in which the forces can be distributed between the spinal column and the attached muscles and ligaments. A corollary of this is that the body has available many different postures to achieve a given end, e.g., picking up a weight from the floor." Though this is currently an area of active research, the relative contributions of the active and passive components of the spine to its stability are still not completely understood [10, 11].

The literature is replete with models that seek to predict the response of the spine under various loading conditions. The objective of these studies is to replicate the physiological load distribution around the spine while accurately simulating the response of the spine as compared to data measured *in vivo*. The majority of these models typically employ one of, or some combination of, three analysis methodologies: finite element analysis (FEA), analytical modeling and optimization, and cadaveric testing. Each has a number of merits and advantages, as well as a number of drawbacks. Validation of these models is challenging. Due to ethical considerations, *in vivo* measurements are difficult to obtain and are limited in their scope. Commonly used as validation for a model's ability to simulate the physiological condition, parameters such as intradiscal pressure, internal shear forces, cortical bone strains, and geometric parameters have been measured either invasively or through imaging techniques such as magnetic resonance imaging (MRI). Chapter 2 will present a representative sample of these models and identify their strengths and their weakness in being able to accurately model the biomechanics of the human lumbar spine. When investigating the biomechanics of the human lumbar spine, it is imperative that physiological constraints be imposed upon the model to maintain physiologic feasibility. The majority of the models presented in the literature make use of a number of constraints on the passive structures and the passive loading of the spine, yet they allow unconstrained action of the musculature. Many require an infeasible amount of muscular support to maintain stability. This is an issue that is largely overlooked in the literature. Though the precise distribution of these loads among the ligamentous spine and the musculature is not well understood, it is reasonable to assume that as least a portion, if not a majority, of the load is carried by the passive structure in the standing posture [12].

## 1.4 Thesis Statement

The primary objective of this research is to quantify the ability of the human ligamentous lumbar spine to support the load of gravity in the absence of active control from the musculature in the neutral standing posture. It is hypothesized that there exists a unique equilibrium configuration of the spine which requires little or no muscular input to remain erect, and that this equilibrium posture coincides with the configuration of the spine in the standing posture. This equilibrium point configuration is the posture of minimal metabolic expenditure and satisfies physiological constraints on muscle activation. This study is an *in vitro* (cadaveric) investigation of the equilibrium point and its stability.

## 1.5 Contributions

Though there have been a number of investigations into the passive stability of the lumbar spine, this is the first study to our knowledge that uses an *in vitro* methodology to demonstrate the load-carrying capacity of the ligamentous spine in the absence of simulated muscle activation. This research has identified a semi-stable equilibrium posture that requires no muscular input. The implications of this finding are far-reaching. We have demonstrated that the lumbar spine is able to stand freely and seek a configuration of minimal energy expenditure while in the standing posture. This new understanding sheds light on how the lumbar spine regulates the contributions of active and passive structures to provide stability. It implies that any spine model that is to be accurate, whether numerical or otherwise, should be able to replicate an equilibrium posture with little or no muscle activation. This finding brings into question the validity of current stability models and indicates a need for significant modification if not a complete revision of the current modeling techniques. Since many of these models are used in the design of spinal implants and devices, a more accurate understanding of the spine's biomechanical response in the neutral posture will allow for better implant design and more accurate simulation of spinal motion. This study is not meant to provide a comprehensive characterization of the stability of the spine. Rather it identifies a critical passive capability of the spine that has been previously overlooked. This thesis will also identify a number of passive components which have not typically been incorporated into stability models that have potential for added stabilization. The stability of the equilibrium posture will be characterized and the need for low-level muscle activations will be assessed.

### 1.6 Thesis Outline

Chapter 2 will begin by giving background information on the anatomy and biomechanics of the spine. It will present the musculature around the spine and will define metabolic constraints on muscle activation in the neutral posture. It will then present a number of the spine stability models found in the literature and will identify the strengths and shortfalls of each. Chapter 3 is a technical journal article describing this research that was submitted to the *The Spine Journal* and is currently under review. It will present the testing methods and the major results of the work. Chapter 4 will provide detail on additional insights that were gained during accomplishment of the work. Chapter 5 will make some concluding remarks and provide recommendations for future research.

# 2 BACKGROUND AND LITERATURE REVIEW

This chapter presents a brief overview of spinal anatomy including the primary active and passive components of the lumbar spine that contribute to stability. It then presents the physiological rationale for a previously-overlooked constraint on muscle activation while in the standing posture. It will then discuss a number of spinal stability models and will identify their strengths as well as their inadequacies with respect to the constraint on muscular activation.

## 2.1 The Structures of the Lumbar Spine

In order to understand the control of stability of the spine, a basic understanding of the active and passive structures that might be involved in stabilization must be gained. This section will present basic spinal anatomy and a number of structures that have been identified as contributing to stability.

### 2.1.1 Gross Anatomy

The spinal column is divided into three segments: the cervical, the thoracic, and the lumbar. It is composed of seven vertebrae in the cervical region (C1-C7), 12 in the thoracic (T1-T12), and five in the lumbar (L1-L5). Caudally, the lumbar spine attaches to the sacrum which in turn is joined to the pelvis by the fibrous sacro-iliac joint. There are two principal curvatures of the spine: kyphosis and lordosis. The thoracic spine is concave anteriorly (kyphosis) and the lumbar and cervical spines are concave posteriorly (lordosis), shown in Figure 2-1. When viewed as a structural column, these curvatures are the cause of instability when the spine is subjected to a purely vertical load. When loaded in this manner, internal moments are generated within the intervertebral discs and the free-standing lumbar spine is only able to withstand loads of up to 100 N [13-15], an order of magnitude less than what is experienced during activities of daily living. The bony structures of the vertebrae are roughly divided into the anterior body (i.e., the vertebral body) and the posterior elements (pedicles, lamina, transverse processes, spinous process, and the facets) (shown in Figure 2-2).

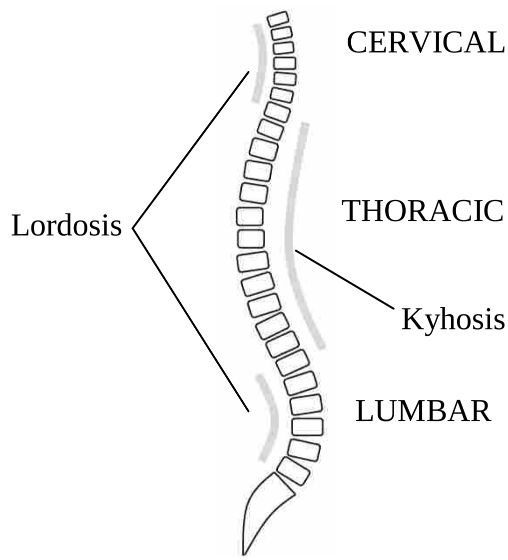


**Figure 2-1: The Columns and Curvatures of the Spine**

### 2.1.2 Spinal Ligaments and Intervertebral Discs

The functional spinal unit (FSU) is composed of 2 adjacent vertebrae and the disc in between. There are 7 spinal ligaments in the lumbar region: the anterior longitudinal, posterior longitudinal, supraspinous, interspinous, ligamentum flavum, intertransverse, and capsular ligaments. These are shown in Figure 2-2. The primary function of the spinal ligaments is to restrict motion. The physical make-up of ligaments provides a very unique set of mechanical behaviors. Ligaments are composed of bundles of collagen fibers surrounded by a ground substance matrix that is primarily composed of water and proteoglycans. The mechanical response of ligaments is non-linear, anisotropic, viscoelastic and inhomogeneous. These properties pose challenges in characterizing their constitutive response. An important property of ligaments is that they exhibit a low modulus of elasticity at low strains and a high modulus of elasticity at high strains. This is primarily due to a crimp pattern of the collagen fibers that "flattens out" as strain increases [16]. In general, the spinal ligaments have some amount of pre-strain in the neutral posture to provide stabilization. As the spine is articulated, these ligaments prevent excessive motion and protect the intervertebral discs.

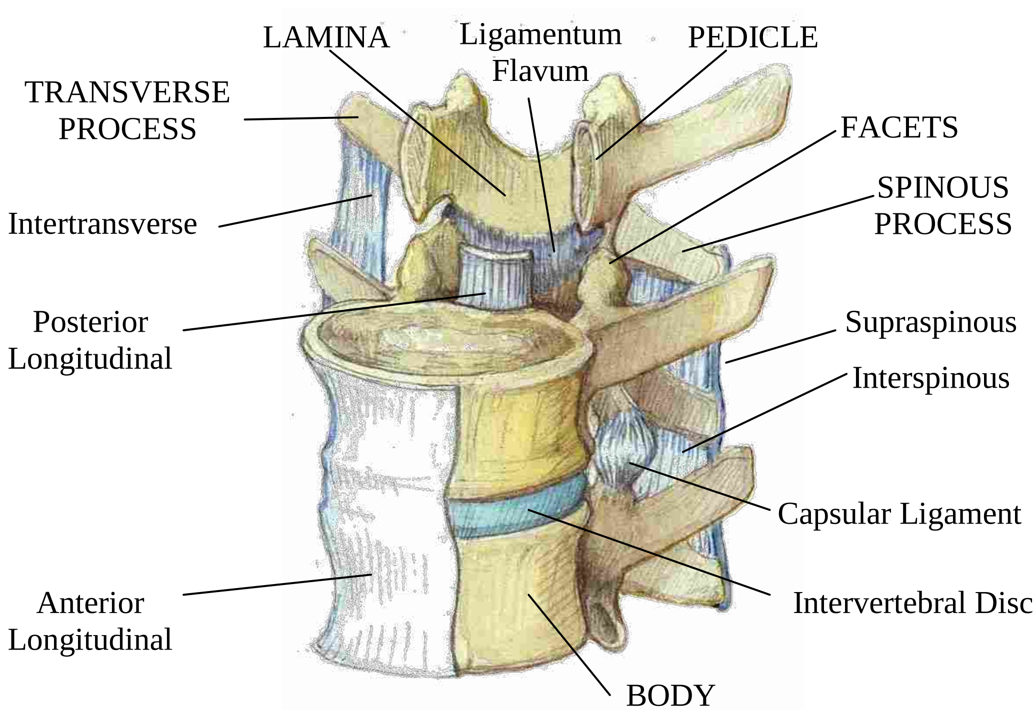


**Figure 2-2: The Functional Spinal Unit (FSU)**

The intervertebral discs join adjacent vertebrae. They are cartilaginous joints and are composed primarily of collagen fibers, proteoglycans, and water. Structurally, discs are composed of an outer ring called the annulus fibrosis and an inner core called the nucleus pulposus, shown in Figure 2-3. The annulus fibrosis is made up of concentric rings of fibrocartilage, termed lamellae, and collagen fibers that attach the disc to the bodies of adjacent vertebrae. The nucleus pulposus is a gelatinous core that functions as a hydrostatic shock absorber [17]. It provides resilience and contributes to the disc's resistance to compression.

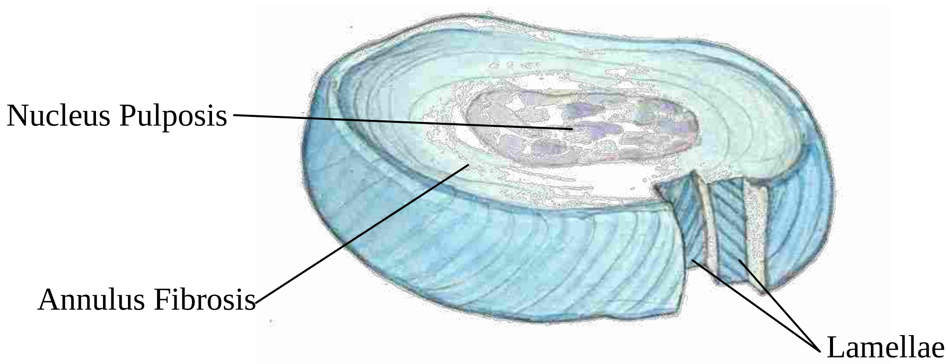


**Figure 2-3: The Intervertebral Disc**

### 2.1.3 Spinal Musculature

The musculature surrounding the spine can broadly be categorized into two systems: the global musculature and the local musculature [5] (see Figure 2-4). The global musculature controls the bulk motion of the upper torso and (generally) causes relative motion between the pelvis and the rib cage. Global muscles typically have a larger physiological cross-sectional area (PCSA) and are responsible for large motions. Co-contraction of the posterior flexor muscles and the anterior extensor muscles can provide supplemental stabilization to the spinal column [18]. Examples of global muscles are the erector spinae and the rectus abdominis. Local muscles generally have a much smaller PCSA and are shorter in length. They may attach lumbar vertebrae to the pelvis or they may articulate adjacent vertebral levels. They are well suited for small adjustments and fine motor control of individual vertebrae [19]. In general they control the minute alterations to the intersegmental angles that are required to create the optimal spinal geometry. Examples of local muscles are the multifidus and rotatores.

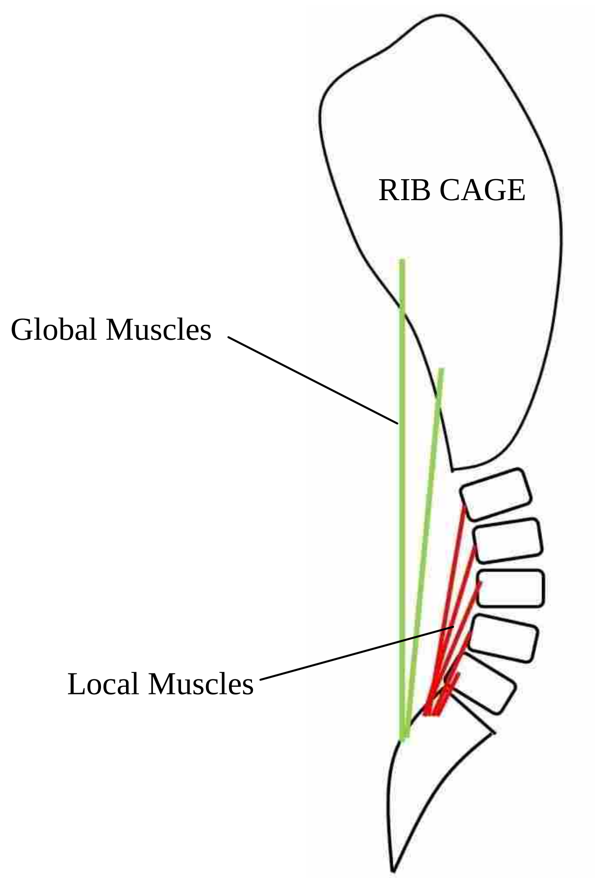


**Figure 2-4: Local and Global Musculature of the Lumbar Spine**

When determining the action of muscles and their effect as they contract, it is useful to quantify muscular input as a percentage of maximum voluntary contraction (MVC). Because the actual force within a muscle is difficult to measure *in vivo*, the activation of a muscle is measured with electromyography (EMG) and is normalized to the maximum possible activation for that muscle. In general, the maximum amount of force a skeletal muscle can exert is proportional to its PCSA. The force in a muscle under sub-maximal contraction then becomes a function of the MVC and the PCSA. The equation for the muscular force then becomes,

$$F_{\mathrm{m}} = \frac{\%MVC * PCSA * k}{100\%} \qquad (2-1)$$

where $k$ is the maximum contractile force constant. In the literature, the value of this constant varies greatly [20-23]. A mid-range value of $k$ = 25 N/cm$^2$ was proposed by Nigg and Herzog [16] based on measurements of *in vitro* muscle force in skeletal muscle under full tetanic contraction. Their experimental procedures give a more reasonable estimate of $k$ than some others, so it will be used in this thesis. Alternatively, if the force required is given (in a numerical model of the spine for example), Equation 2-1 can be rearranged to calculate the %MVC:

$$\%MVC = \frac{F_{\mathrm{m}}}{k * PCSA} * 100\% \qquad (2-2)$$

Naturally, the PCSA of the muscle in question must be known. There are many sources of empirical data on these cross-sectional areas [20, 24]. This calculation of %MVC will become important later on as a constraint on muscular activation in the standing posture is introduced.

### 2.1.4 Intra-Abdominal Pressure

There is a dynamic pressure within the abdominal cavity that rises and falls with different activities and postures. This intra-abdominal pressure (IAP) has been shown to increase from 16.7 and 20 mm Hg in sitting and standing to 107.6 and 171 mm Hg when coughing and jumping, respectively [25]. Increased IAP has been shown to reduce the compressive load on the spine and provide stabilization during physical exertions such as lifting [26]. However, there is conflicting data in the literature concerning the role of IAP in the standing posture. Some researchers have attempted to simulate the effects of IAP in the standing posture with numerical models and have suggested an increase in stability [9, 27]. A more recent study has shown these effects to be negligible [28]. Since IAP is nominally low in standing, its direct local action on the vertebrae is not likely to cause any motion or stabilization. However, it is possible that it may be a necessary restraining force to the muscles that insert into the thoracolumbar fascia. Because of its likely minor contribution, and because of the difficulty in simulating it *in vitro*, IAP is neglected in this study.

### 2.1.5 Thoracolumbar Fascia

The thoracolumbar fascia is a thin sheet of fibrous connective tissue that runs nearly the entire length of the lumbar spine. It is the insertion site for a number of torso muscles, including the latissimus dorsi, external obliques, internal obliques, and the transversus abdominis. These muscles wrap anteriorly around the lower torso and join at the linea alba on the ventral side. Tension in these muscles pulls laterally and anteriorly on the spinous processes of the vertebrae and can affect the spine's postural configuration. The action of these muscles on the thoracolumbar fascia is thought to stiffen the intervertebral joints and increase resistance to flexion [29, 30]. Barker et al. [29] simulated fascial tension in individual FSUs to investigate its contribution to segmental stability. They demonstrated that with a 20 N tension in the thoracolumbar fascia, the FSU significantly (~44%) increased its initial stiffness. They propose that fascial tension is better suited for fine-tuning of segmental motion rather that for bulk motion of the entire spine. Tension in the transversus abdominis and internal obliques has also been suggested to contribute to segmental stiffness via the generation of IAP [31]. Though this may be the case in postures other than standing, a constraint on muscle activation that will be presented later will restrict the amount muscle force, and therefore the amount of IAP, that can be generated. Due to complexities in simulating the thoracolumbar fascia, it was not included in the present study. Rather, the relative contribution of the ligamentous lumbar spine alone will be characterized and suggestions for further investigation into the contributions of other passive structures will be made.

### 2.1.6 The Iliolumbar Ligament

The iliolumbar ligament (ILL) is a passive structure that has been largely overlooked in full-spine stability models. There is, however, a body of literature that investigates the ILL in isolation to describe its contribution to stiffening the lumbosacral joint. It spans the gap between the posterior superior aspect of the ilium (pelvis) and the transverse process of the L5 vertebra. A number of sources report that an additional band that attaches to the L4 transverse process [32, 33], but one study was unable to identify the L4 band in 100 cadaver specimens [34]. The L5 portion of the ILL is typically divided into a posterior, middle, and anterior portion. The posterior band is normally the thickest and has been shown to restrict flexion motion [32, 34, 35]. Pool-Goudzwaard et al. [36] concluded that with L5 being stabilized by surrounding tissue and by the trunk weight, the L5-S1 joint cannot give way with tension in the ILL. Despite these very convincing findings, the ILL has not been incorporated into any full-spine biomechanical models that attempt to characterize stability. This is perhaps due to difficulty in obtaining cadaveric specimens with intact ILLs or due to the lack of constitutive properties for the ligament that could be incorporated into finite element models. The role of the ILL will be explored very briefly in the present study and will be discussed hereafter.

## 2.2 Biomechanics of the Spine

The mechanical response of the intervertebral joint is typically characterized by sectioning the spine into functional spinal units. The motion of the FSU is tracked (usually optically) as it is subjected to loading conditions (usually combinations of torques and compressive loads). A load-displacement graph of a typical spinal segment in flexion/extension is shown in Figure 2-5. There is a narrow section in the middle of the graph that can be approximated as linear. At larger deflections, the response becomes highly non-linear. Hysteresis causes the curve to follow a different path for the loading and unloading phases.

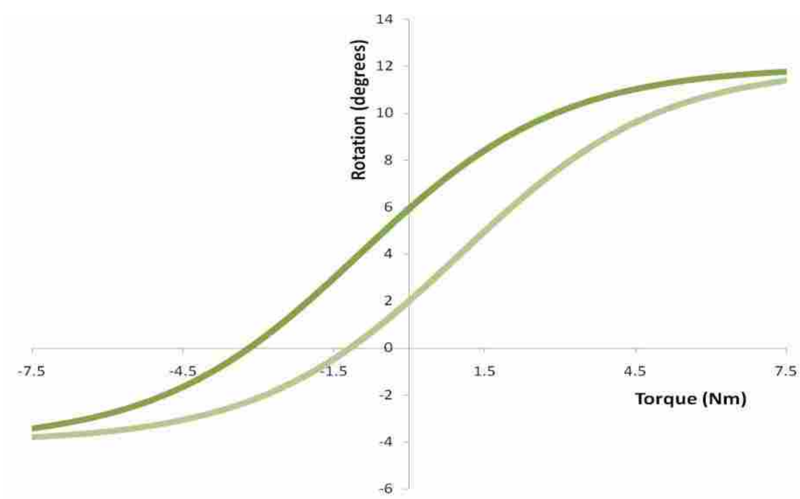


**Figure 2-5: Flexion-Extension Response of a Typical Lumbar FSU**

The instantaneous axis of rotation (IAR) of the FSU is the axis in space about which the vertebrae rotate and defines the kinematics of spinal motion. A typical location for the IAR of the lumbar FSU in flexion and extension is shown in Figure 2-6. Around the neutral posture, the IAR has been shown to be located approximately 4 mm posteriorly from the center of the vertebral endplate [37, 38]. As the spine experiences physiologic motion, the location of the IAR translates in the A-P direction and in the superior-inferior direction [39] due to the non-linear nature of the disc. In the healthy spine the IAR resides within a small region around the posterior portion of the vertebra. Pathologically degenerate discs will affect the location of the IAR [8, 21] and will therefore affect the biomechanical response of the entire spine.

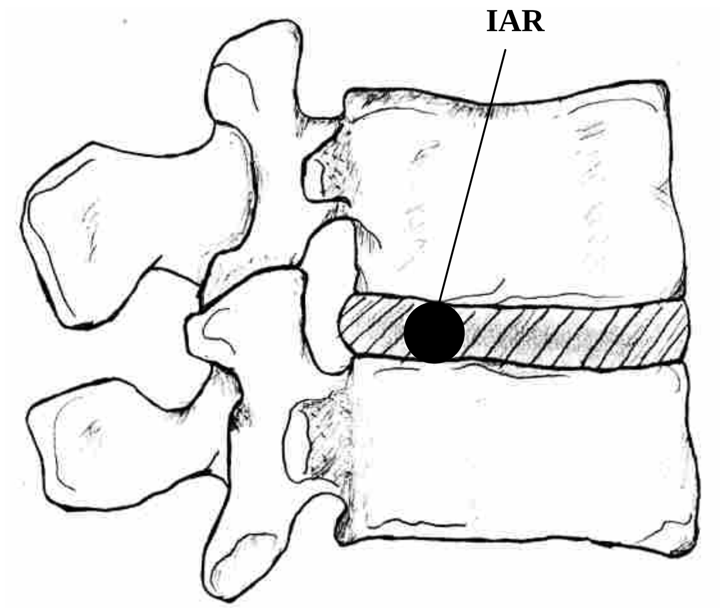


**Figure 2-6: The Instantaneous Axis of Rotation (IAR) of a Lumbar Segment**

The health of the intervertebral disc has been shown to have a significant effect on its biomechanical properties. Thompson et al. [40] proposed a standardized characterization scheme to classify the health of intervertebral discs into one of five categories (I – IV), known as the Thompson grades. The healthy disc (Grade I) has been described as well-hydrated with a bulging gelatinous nucleus, distinct fibrous lamellae in the annulus, and thick, uniform endplates. Pathological discs will exhibit varying degrees of dehydration, fibrous bundles in the nucleus, and osteophytes throughout. As the disc becomes dehydrated, its resistance to hydrostatic compression is reduced and more mobility is observed. Figure 2-7 shows the general trend of the amount of segmental motion that occurs for increasingly degenerate discs [37, 40]. The drop in mobility (increase in stiffness) between Grade IV and Grade V can be attributed to a nearly complete degeneration of the disc resulting in interference of the adjacent vertebral bodies, and therefore increased stiffness. In a cadaver-based experiment such as the present study, the overall health of the spine and specifically of its intervertebral discs must be considered to have an effect on the overall biomechanical response.

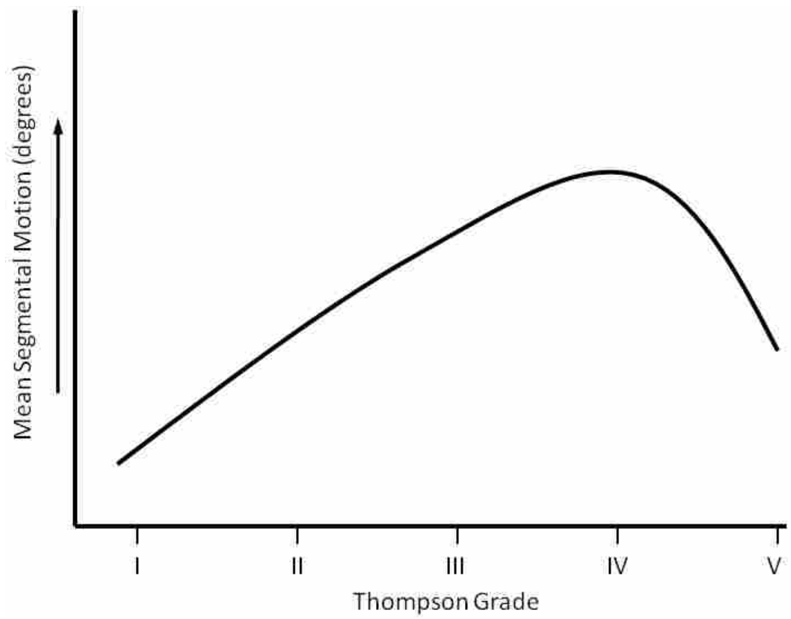


**Figure 2-7: Mean Segmental Motion as Correlated with the Thompson Scale for Intervertebral Disc Degeneration, Adapted from Thompson et al. [40]**

## 2.3 The 5% MVC Constraint

The passive structures of the spine contribute to its overall stiffness, but spinal motion control cannot be achieved without the musculature. It is therefore important do define the role of the muscles in stabilizing the spine. Though a relationship can be drawn between the EMG pattern of a muscle and its force, it is extremely difficult to isolate individual muscles in the back to measure their level of activation. Therefore, there is little if any *in vivo* data on the contributions of each muscle component to the overall spinal motion [20]. Consequently, the primary method of determining muscle force as it relates to stability has been through numerical simulation of the back musculature. In general, researchers use some type of universal model of the musculature such as in [20] to determine the role of the morphological features that are incorporated into the biomechanical model.

There is an almost universal consensus that the human spine requires muscle force to remain stable [1, 18, 19, 41, 42]. While this may be true in the very strictest sense, any amount of muscular activation in the neutral standing posture must be constrained to a reasonable level. This is based on the qualitative observation that a human is able to sustain this posture for an extended period of time before experiencing fatigue. Jonsson [43] described this quantitatively by observing that sustained muscle activations beyond 5% MVC will cause pain and eventual fatigue. Cholewicki et al. [18] recognized that the spine should be able to remain stable in the standing posture while being subjected to this constraint. One of the most commonly-cited muscles of the lumbar region to contribute to stability is the multifidus. Cholewicki and McGill [44] measured the activity of the lumbar multifidus in the standing posture and found it to not exceed 3% MVC.

The constraint on muscle activation is not to suggest that muscles are entirely uninvolved in the stabilization of the spine around the neutral posture. Weyand et al. [45] have shown an increase in metabolic expenditure from the supine resting position to the standing position (1.08 $W \cdot kg^{-1}$ and 1.25 $W \cdot kg^{-1}$). However when contrasted with the metabolic rate of brisk walking (6.90 $W \cdot kg^{-1}$), this difference between expenditures in resting versus standing is minimal. O'Sullivan et al. [31] observed a significant increase in activation of the internal obliques and erector spinae muscles from resting to erect standing. However, in that study the level of muscular activation was normalized to a sub-maximal contraction, not to the MVC as is the more common practice. It is therefore impossible to draw any definitive inference from their results as compared to the data reported by others. They did, however, show very little increase in erector spinae activation. Recognizing that some small amount of muscle contraction may be necessary in the neutral posture, any biomechanical model of the spine must adhere to the 5% MVC requirement in order to be physiologically feasible. This constraint is largely overlooked in the literature but is a critical requirement and must be taken into consideration. A summary of the PCSA of a number of spinal muscles and their 5% MVC capacities are reported in Table 2-1. Any muscular force in the neutral standing posture beyond these values is considered unreasonable. As will be shown hereafter, the vast majority of the biomechanical models of the spine in the literature violate this constraint.

**Table 2-1: 5% MVC Values for Select Lumbar Muscles**

| Muscle | PCSA ($mm^2$) | MVC (N) | 5% MVC (N) |
|---|---|---|---|
| Iliopsoas | 334 | 83.5 | 4.175 |
| Multifidus | 211 | 52.75 | 2.638 |
| Quadratus Lumborum | 88 | 22 | 1.10 |
| Longissimus (local) | 116 | 29 | 1.45 |
| Iliocostalis (local) | 189 | 47.25 | 2.363 |
| Longissimus (global) | 1100 | 275 | 13.75 |
| Iliocostalis (global) | 600 | 150 | 7.50 |
| External Obliques | 1576 | 394 | 19.70 |
| Internal Obliques | 1345 | 336.25 | 16.813 |
| Erector Spinae | 1700 | 425 | 21.25 |

## 2.4 Stability Models of the Spine

Spinal stability has been an area of intense research for many years. The present study was inspired by preliminary work done by Peter Halverson at Brigham Young University on the "balance spine" concept, which will be briefly discussed. Early models of the spine focused on simplifying the system to be statically determinate and characterizing the bulk loading conditions. More recently, models have become complex three-dimensional finite element analysis (FEA) optimizations and have attempted to approximate physiological conditions as closely as possible. In the literature there are three principal research groups that have made the most significant contributions to understanding the mechanisms behind spinal stability: Patwardhan et al., Rohlmann et al., and Shirazi-Adl et al. Each of these groups has contributed a number of important findings which have helped define the basic principles governing spinal stability. However, despite their invaluable insights, all of these groups fail to adhere to constraints on muscle activation in the neutral posture, much less the 5% MVC requirement. The positive contributions of each of these groups as well as their shortcomings will be discussed in succession. Additionally, two other models that have proved important to understanding spinal stability will be briefly discussed.

### 2.4.1 The Balanced Spine Concept

The lack of attention to the passive structures of the spine and their contributions to stability in standing was recognized by Peter Halverson at BYU. He proposed a model called the "balanced spine" that attempted to determine the ability of the thoracolumbar spine to remain upright under the gravity load [38]. This simple model used an estimation of the distribution of the body weight as discrete masses and reduced the spinal system to a series of stacked levers, each balanced around a fulcrum (see Figure 2-8). To accomplish this, a method introduced by Pearsall [46] which partitions a full-body computed tomography (CT) scan into a series of cross-sectional slices was used. The thickness of each slice is the combined height of its vertebra and the inferior disc. Using a calibration factor, the image intensity of each pixel is correlated to a material density, enabling the total mass and center of mass (CM) of each segment to be calculated. The masses and CG locations from Pearsall [46] were used in the balanced spine model. For a comprehensive description of the CT segmentation method, and for the calculations of the CMs that were used in the present study, see Appendix A: Calculation of Mass Segments from CT Images.

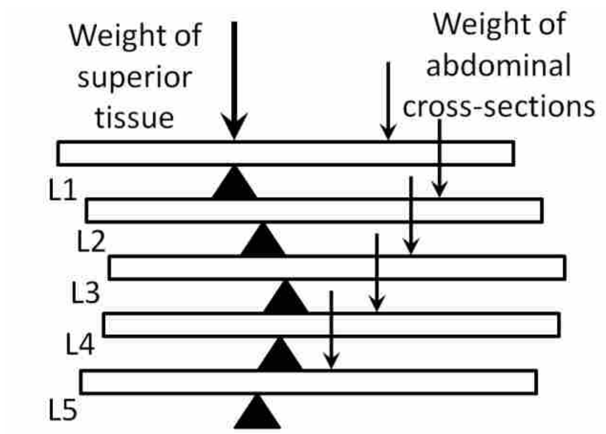


**Figure 2-8: The Balanced Spine Model**

In the balanced spine model, the freely-rotating fulcrum of a given level rests atop the massless lever of the inferior level. The CM weights are applied on the lever and create flexion moments at each level. The cumulative mass of all levels superior to the current level acts through the fulcrum of the superior level. A free-body diagram of a single level is shown in Figure 2-9. With the CM location and vertebral center locations known, the equations of moment equilibrium are applied to calculate the location of the instantaneous axis of rotation of the level, namely:

$$l_{IAR} = \frac{m_{ST} l_{ST} + m_i l_i}{m_{ST} + m_i} \qquad (2-3)$$

where $l_{IAR}$ is the distance to the IAR, $m_{ST}$ and $l_{ST}$ are the mass and distance to the weight of the superior tissue (the superior level IAR), and $m_i$ and $l_i$ are the mass and distance to the CM of segment *i*. The distances used in this equation are measured from an arbitrary reference point; only the relative differences in the distances are of import. The balanced spine theory postulated that the more massive superior tissue weight is balanced by the smaller segmental weight which is applied at a much longer lever arm. The static equilibrium equations of the system initially indicated that this loading scenario forced the IAR at each level to be near the vertebral center (-4 mm), which corresponds to the physiological location of the IAR *in vivo*. However, after closer investigation, the equations used in the calculation were found to be in error and did not, in fact, produce fulcrum locations that corresponded with the physiologic IAR. In the geometric configuration of the lumbar spine, the vertebral centers (and therefore the IARs) of L2 and L3 are located anteriorly of the L4 and L5 vertebral centers. The equilibrium equations therefore produced L4 and L5 IAR locations that fell much further anterior than their physiologic locations. It is postulated that the *in vivo* loading scenario is much more complex than can be represented by the simple balanced spine model. Though the balanced spine model was found to be invalid, it did identify a method to more accurately simulate the weight of the torso *in vitro.* It also sparked interest in the passive capabilities of the spine to "balance" in a passive equilibrium posture.

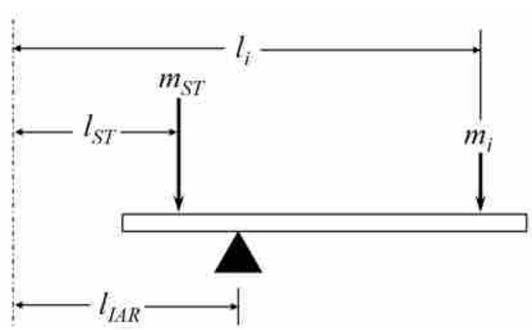


**Figure 2-9: Free-Body Diagram of a Single Spinal Level in the Balanced Spine Model**

### 2.4.2 Patwardhan et al.

*In vitro* testing of the human spine presents a challenge in simulating the complete loading condition of the *in vivo* spine. With a complex distribution of the body weight and interactions with hundreds of muscle fascicles, a comprehensive *in vitro* model is not possible. The work of many researchers has therefore centered on making simplifications to the physiological loading condition that would allow for more simple application to cadaveric specimens. Until recently, the body weight has been simplified as a single lumped mass applied at the most cranial vertebra, which, as has been previously noted, can only reach a fraction of the total body weight before the spine buckles.

Patwardhan et al. discovered a loading condition termed the "follower load" that allows for a compressive load of physiologic magnitude to be applied to the spine while maintaining stability [15, 47]. The follower load is applied in such a manner that the line of action of the force follows the tangent to the curvature of the spine. In order to induce stability, it must pass through the IAR of the spine at all levels. It is implemented *in vitro* by guiding a tensioned cable bilaterally along the curvature of the spine with eyelets embedded into the vertebral bodies. The follower load creates a state of nearly pure compression in the intervertebral discs with very small internal moments and shear. Patwardhan submits that this is a necessary physiological requirement for the standing posture, yet there has been little evidence to suggest that the intervertebral disc does not experience shear and moment loads in standing. The follower load does, however, accurately replicate intradiscal pressures. A study by Rohlmann et al. has shown that a follower load of 500 N replicates the intradiscal pressures of the *in vivo* spine.

The literature has demonstrated that a compressive load applied to an FSU increases its resistance to bending [40, 48]. While the follower load applies this condition to all discs simultaneously and thereby increases the overall stiffness of the spinal column, there has yet to be presented a reasonable explanation for the origin of such a load based on the morphological features composing the spine. Patwardhan suggested that the follower load is created in the lumbar spine by the action of the muscles surrounding it [15]. Studies by Patwardhan et al. [1] and Han et al. [49] attempted to replicate the follower load in the lumbar spine with muscle forces using numerical simulations. Patwardhan presented a frontal plane model that made a number of idealizations to simplify the musculature, namely that a single muscle force was applied bilaterally to each vertebra in the frontal plane. He allowed the spine to undergo small deviations in lateral bending and recorded the muscular force that would be necessary to restore the column to the upright position. Though the resultant force of the muscular components followed the curvature of the spine (the follower load condition), the average muscle force required was approximately 44 N with the maximum force reaching 219 N. These are well outside the range of feasible values for muscle activations based on the 5% MVC requirement. Han presented a more sophisticated model of the sagittal-plane stability of the lumbar spine in the standing posture. He incorporated 232 muscle fascicles and a rigid mass above the L1 vertebra representing the weight of all tissue superior to L1. He used a nonlinear optimization routine to determine the muscle activations needed to create the follower load. The model was successful in creating a follower load condition, but the forces in individual muscles ranged from 4 N to 117 N (1% - 100% MVC). Metabolic energy expenditure and muscle fatigue arguments dictate that forces of these magnitudes cannot reasonably be sustained in the standing posture.

In summary, the follower load model can be very useful when conducting *in vitro* testing of the human spine. It is simple to implement and allows compressive loads of physiologic magnitude to be applied to the spine. This is particularly useful when conducting mobility testing, as the hardware is minimal and unrestricting. Despite its ability to simulate physiological intradiscal pressures, researchers have been unable to describe it from a physiologically realistic standpoint.

### 2.4.3 Rohlmann, Bergman, Wilke et al.

Many of the studies conducted by this group revolve around a specialized spinal fixator device that is instrumented to record axial forces and bending moments in the lumbar spine. The spinal fixator is a rigid orthopaedic device that is used to immobilize two adjacent FSUs. It spans an entire vertebra and is screwed into the pedicles of superior and inferior vertebrae. The authors created a custom fixator with load cells and strain gauges to measure the loads and deflections in the fixated segments. The spinal fixator serves as the basis for comparison between *in vivo* and *in vitro* spinal loading, as it has been implanted in a number of live patients and is also used in cadaveric testing. Though this is a very clever method of measuring the loads in an *in vivo* spine, it is important to point out that the spinal fixator bridges an entire FSU and will therefore restrict motion almost entirely at that level. The result is a drastic change in the biomechanical response of the spine as a whole, a fact that must be taken into consideration as the results of a number of studies are discussed. When comparing analytical or cadaveric data to the *in vivo* data recorded by the spinal fixator, the quantitative results are restricted to spines instrumented with a spinal fixator and cannot be extrapolated to include un-instrumented spines. However, some general trends that they identify can be applied to spinal stability as a whole. In an effort to define the mechanisms behind stability, they explore a number of different loading conditions in search of the one that provides results most similar to *in vivo* data. There are number of invaluable contributions in this body of research, but similar to studies by Patwardhan, excessive muscle activations go unnoticed.

Two experimental investigations in 1998 [50, 51] attempted to determine the effect of antagonistic muscle activation on overall spinal motion. They instrumented cadaver lumbar spines with a cables attached to the L4 vertebra to simulate the action of the psoas and multifidus muscles. With coactivation of 90 N in the agonist muscle (muscle causing the desired motion) and 30 N in the antagonist muscle (muscle restricting the desired motion), they observed that the spine was stiffer in lateral bending and axial rotation, but less resistant to mobility in flexion/extension. The decrease in stability in flexion/extension is contradictory to observations made by Patwardhan [15] and Cholewicki [18]. These studies did not focus on the neutral posture, so it is difficult to determine the physiological feasibility of the simulated muscle forces they used. A significant contribution of these two studies is that they pointed out the possibility of complex muscle interactions that might be required for stability.

One of the most commonly used loading conditions for *in vitro* testing is the application of pure bending moments to the most cranial vertebra. This method has primarily been used for a lack of a more complete understanding of the true loading condition of the spine, and because it is easy to implement. Wilke et al. [41] used the instrumented spinal fixator to determine if the pure moment condition was an accurate simulation of the physiologic loading. For lateral bending and axial rotation, they found qualitative similarities between *in vivo* and *in vitro* fixator loads. However, the differences recorded in flexion did not allow them to make a definitive correlation between *in vitro* and *in vivo* data. They attribute this difference to muscle forces but do not supply any further explanation as to the potential function of the muscles. Though they conclude that pure moments provide a reasonable simulation for physiological loading, their data is based solely on the loads measured in the spinal fixator and overlooks other important measures for comparison such as intersegmental angles. Further, no body weight load is applied to the system so many potentially important interactions could not be evaluated.

In a much more advanced cadaveric study conducted in 2003, Wilke et al. instrumented a number of cadaver spines to determine the role of trunk muscles in flexion and extension [19]. This study is perhaps the most comprehensive *in vitro* simulation of the physiological loading condition in the literature. The experiment included the ability to simulate body weight, apply pure moments, apply a follower load, and simulate a number of muscle groups with pneumatic cylinders. The objective was to explore 12 different combinations of body weight, follower load, and muscular activation and determine which scenario most closely approximated the *in vivo* condition. Data from *in vivo* spinal fixator studies were used as the baseline for comparison. The study identified a number of key principles. As the vertical force, follower load, and muscle force were increased, an increase in intersegmental rotation, intradiscal pressure, and internal fixator load were observed. Specifically, the global muscle force (erector spinae) increased as the spine was moved into flexion. They also made a very interesting observation that the follower load had very little effect on the magnitude of the global muscle force required to maintain stability. Others (including members of the Rohlmann research group) have attempted to describe the follower load as being supplied by local muscles. That being the case, the results of this study would indicate that the local muscles are uninvolved in flexion-extension motion of the spine, which does not agree with the literature [18, 52]. Under the optimum loading scenario with a follower load of 200 N and an upper body weight of 260 N, they reported a force of 100 N in the erector spinae for standing. This value is approximately five times the acceptable 5% MVC threshold calculated in Table 2-1. This model did, however, show that hip flexion angle had a significant influence on the total force required in the erector spinae. The effect of hip (sacrum) flexion will be studied in the present work.

A finite element model of the lumbar spine was developed by Zander et al. to determine the muscle forces involved in upper-body inclination [52]. This study comes closer than most others presented in this thesis to obeying the 5% MVC constraint. The advantages of an FE model lie in its ability to simulate the complete musculature and loading conditions around the spine and thereby gain a more complete understanding of their role in stabilization. This study simulated a global dorsal muscle force and the local muscle forces of 70 muscle fascicles. To eliminate the problem of redundancy, the force of all local muscles was assumed to be the same. The body weight was positioned in such a way that dorsal muscle forces were required for stabilization (weight applied anterior of the spinal column). The study showed a curvilinear relationship between global muscle force and flexion angle, with a linear relationship between $0^o$ and $10^o$ of flexion, followed by a parabolic section of decreasing slope between $10^o$ and $30^o$. When in the standing posture, and with a 5 N activation in all local muscles, a 50 N force in the global muscles was required for equilibrium. Referring to Table 2-1, these muscle forces still exceed the 5% MVC requirements, but they are on the same order of magnitude, so it is conceivable that they could still be physiologically feasible. In the absence of local muscle activation, a global muscle force of 300 N is required, which is unreasonable. Though they produce some promising results, they do recognize the shortcomings of their model. Requiring that the force in every local muscle fascicle be the same is not realistic. In reality the force in each muscle will be governed by the amount of segmental rotation that is required at that level and by the complex neuromuscular recruitment pattern that is unique to every person. In short, this study demonstrated the possibility of modeling the lumbar spine using FE tools and obtaining fairly realistic results that compare favorably with *in vivo* data. It is one of the few studies that does not use or attempt to simulate a follower load, yet the reported muscle forces are much lower than those studies that do use the follower load.

In a follow-up study to [19], Rohlmann et al. attempted to rectify the disagreement between *in vitro* and *in vivo* spinal fixator loads in flexion/extension. They had previously achieved agreement in the cadaveric model only with hip flexion. In [28] they made use of the finite element model mentioned above and explored the contributions of intra-abdominal pressure, preload in the fixators, and combined hip and lumbar flexion to achieve better agreement with *in vivo* measurements. Once the model was deemed more reliable, they removed the spinal fixators to determine the muscle forces needed for stability in the free-standing lumbar spine. They reported that IAP has a negligible effect when applied directly to the vertebral bodies in standing. An erector spinae force of 170 N was estimated in the standing posture (without a spinal fixator), which is 70 N more than they reported in [19]. They concluded that while the follower load is a convenient tool for adjusting the *in vitro* intradiscal pressure to physiological level, it may not provide the most accurate simulation of the *in vivo* loading condition. Nevertheless, without a better paradigm, the authors support the postulation that the follower load is created within the lumbar spine with local muscle forces.

In [11] the same group further studies six other loading conditions and compares the response to intersegmental rotations found in other studies. They conclude that the follower load most accurately simulates the *in vivo* condition. However, the new loading conditions studied were not very diverse; the only external load applied was a vertical body weight at the L1 vertebral center. They do not attempt to simulate the continuous distribution of the body weight throughout the lumbar region. Interestingly, they also find that the intradiscal pressure is generally unaffected by the mode of loading, i.e. all six loading conditions produced similar pressures. As was previously mentioned, the group attempted to determine the muscular activations that would be required to produce the follower load and reported activations of up to 100% MVC [49].

In summary, this group has developed some innovative ways of measuring forces within the *in vivo* and *in vitro* spine. They have used a number of tools (cadaveric studies, FE simulation, and spinal fixator device) to study various loading conditions and have compared results to *in vivo* data. They have identified a number of important contributing factors to spinal stability such as:

1. The stiffening of the spine under compressive load
2. Increased global muscular force required as the spine is rotated into flexion
3. The insignificant effect of the follower load on global muscle force
4. The importance of hip flexion on lumbar stability
5. Minimal contribution of IAP to stability

A number of these observations will be shown to be significant in the present work. In these studies, some of the muscular forces reported are among the lowest in the literature, though they are still significantly greater than the 5% MVC criterion. They also report some of the largest activations.

### 2.4.4 Shirazi-Adl et al.

Whereas the Patwardhan and Rohlmann groups have used a combination of *in vitro* and numerical testing methods, Shirazi-Adl et al. almost exclusively employ numerical techniques. Their focus is on the combinatorial effects of postural parameters and the distribution and positioning of body weight on overall stability. Their work does not presume the follower load model. There are a number of key principles identified by this group, yet as the others presented here, the muscle activations they report are generally in violation of the 5% MVC criterion.

An early study by Shirazi-Adl and Parianpour [53] explored mechanisms to stabilize the lumbar spine in the neutral posture. Though the methods presented in this paper do not represent a particularly realistic loading scenario, they do provide valuable insights into stability. They use a simple non-linear FE model of the lumbar spine that incorporates all ligaments and the non-linear inhomogeneous nature of the discs. They identified two simple boundary conditions which, when applied at the L1 vertebra, greatly enhanced the load-carrying capacity of the lumbar spine. With an applied flexion moment coupled with a constraint on the translation of L1, they were able to apply an axial load of 400 N at the L1 vertebral center before instability ensued. This is approximately a fourfold increase from the capacity of the unconstrained lumbar spine. Though the authors do not provide an explanation for the application of the translational constraint, they do pose a reasonable source for the flexion moment. In the study they apply the compressive load at the L1 vertebral center, yet they recognize that in the *in vivo* spine the body weight generally falls anterior of the L1 vertebral center. The applied flexion moment compensates for the anatomically off-centered location of the body weight. They therefore postulate that by carefully positioning the CG of the upper body, the optimal flexion moment is created to stiffen the spine and increase its load-bearing capacity. This is an extremely important observation that has identified a passive mechanism that has the potential to provide added stabilization to the free-standing spine. The load on the musculature is thereby reduced and metabolic expenditure is minimized. This observation will be valuable as the current work is presented. Additionally, this study proposes that the thoracolumbar fascia can have a stabilizing effect on stability.

A subsequent study extended the investigation to include the entire thoracolumbar spine [12]. Additionally, the body weight was uniformly distributed on all vertebrae at their centers. Working on the observation that the *in vivo* body weight induces moments in the spine, flexion moments were applied at each vertebra. With optimized moments at each level, a total load of 1000 N was applied to the full spine without buckling. However, the segmental moments required at each level for this scenario reach as high as 22 N-m, which seems quite unrealistic. Even if the largest muscle in the back, the erector spinae, were activated at 5% MVC, it would need to work at a moment arm of approximately 1 m to achieve this torque. Nonetheless, the qualitative observation that segmental torques stabilize the spine is noteworthy. Another conclusion made by the authors is that an optimal pelvic rotation will provide additional stabilization, though pelvic rotation alone is insufficient to keep the spine erect.

In a similar study [54] Kiefer et al. applied the body weight segmentally at the CM locations of body slices corresponding to each vertebral level. This was found to be a more accurate simulation of the real physiologic gravity load and to create favorable flexion moments at each level. They also explored sacral rotations and applied global muscle forces at the T1 and L1 vertebrae to determine the muscular requirement to maintain stability. They found that the simulation of the distributed body weight as segmental masses applied at their respective mass centers greatly improved the stiffness of the spine. The authors confirm that sacral rotation has a notable effect on stability. In one of their two FEA models they report very low muscle forces (~5 N each), but in the second model they report activations between 1 N and 407 N. Though the authors attribute the motivation for their study to the observation that muscle activation in the trunk is very low in standing, they fail to address the 407 N muscle force as being potentially unrealistic. Despite this oversight, they have shown that the geometry of the spine and the method in which the body weight is simulated are important factors for stability.

In a study utilizing a more comprehensive model of the musculature, Keifer et al. [55] sought to define a synergistic relationship between the role of muscle force and the passive stiffness of the spine. 60 local and global muscles were simulated as linear springs and were combined with the effects of the passive spine. Global muscles that attach to the ribcage were found to be important for control of the overall spinal posture and maintenance of equilibrium. Their contribution is manifest in the careful positioning and modulation of the CG of the torso. One of the few models seeking to minimize muscle force, they apply a minimizing cost function on muscle activation and pursue stability of the spine. The cumulative force in all local and global muscles at equilibrium ranged from 128.5 N to 216.0 N. When distributed across all 60 muscles, it is conceivable that this model maintains stability while adhering to the 5% MVC criterion. However, individual muscle forces are not reported, so it is not possible to draw a definitive conclusion. Additionally, in this model an arbitrary horizontal stiffness is applied at the T12 vertebra that remains unexplained from a physiological standpoint. Though this horizontal support does provide additional stiffness to the spine and relieves the load on the musculature, the authors' failure to describe the physiological origins of this force brings into question the validity of the reported muscle forces. In relation to the role of the global and local muscles, they found that the action of the global muscles alone, in conjunction with the passive resistance of the spine, are sufficient to maintain equilibrium for small sagittal translations of the spinal column, but that the addition of local muscle activation provides additional stiffness and relieves some of the load on the global muscles.

El-Rich et al. [6] extended this muscle-driven model to include *in vivo* validation of the results. They measured the kinematic motion of the spines of 15 individuals using skin marker tracking. EMG muscle activation was also measured as a guide for muscle force used in the computational model. In addition to the muscular activation required for stability, they predicted the response of the spinal system as weights were held at the sides and out in front of the body. In general they predicted reasonable muscle activations (2 N – 34 N) in all muscles of the back, yet many still exceed the 5% MVC criterion as calculated in Table 2-1. All of the EMG activations were normalized to the largest contraction observed during the series of exercises that were performed. The activations in neutral standing were on the order of 15%, but this number is not directly comparable to the activations computed in Table 2-1 because it was not normalized to the maximum voluntary contraction. It is likely that these activations would be closer to 5% if normalized to the 100% MVC. A significant principle was observed in this study in both the numerical model and in the *in vivo* study: weights held in the hands at the sides did not increase the amount of back muscle activation. This observation points towards an inherent ability of the spine to adapt to vertical loads without muscle action. Interestingly, a corresponding increase in the margin of stability was observed under the additional load held in the hands. Coactivation of the abdominal muscles were shown to increase overall stability but also resulted in increased back muscle activity. There is an apparent tradeoff between coactivation providing added protection to the soft structures of the spine and the additional metabolic cost of higher muscle activation. In short, this model endeavored to minimize muscle stress while exploring the degree of stability induced by different levels of muscular input. They have created very sophisticated FE model of the spine and have achieved relatively low activations in the standing posture.

In summary, the studies conducted by this group began with very simple simulations of the spine and summarily increased in complexity and in the impact of their contributions. They provided a number of invaluable insights into the role of the passive structures of the spine, particularly to the importance of the body weight and the loads induced by it. Other key contributions are:

1. Flexion moments applied at each vertebra stabilize the spine
2. These flexion moments can in part be created by the segmental masses of the torso
3. Careful positioning of the CG of the upper torso with the global muscles can add to the load-bearing capacity of the spine
4. Optimal pelvic rotations can stabilize the spine
5. The global and local muscles can work synergistically with the passive structures of the spine to resist moments
6. Weight added to the hands held at the sides does not increase back muscle activation.

Most importantly, this group has been able to simulate stability in their later studies with relatively low muscle activations without the use of a follower load. The local muscles, without being constrained to adhere to the follower load, are able to make the fine postural adjustments that are necessary to add stability without the large forces required by the follower load. Though a number of the muscular activations might be large compared to the 5% MVC requirement, their later studies produced activations that are much closer and could be very near the physiologically feasible limit.

### 2.4.5 Others

Early models of the spine were dependent on the force-couple model, a method of determining the forces and moments at each vertebral level using equations of static equilibrium.

One of the hallmarks of spinal stability research, and one of the first models to use this method, is a paper produced by Bergmark [5]. The basis of this model is what is often termed the "inverted pendulum" system (shown in Figure 2-10). This model treats the spine as a series of stacked pendulums that are weighted with a vertical load and are restrained by a linear spring representing a muscle force. Often, the bulk response of the intervertebral discs and the spinal ligaments are lumped into a torsional spring applied at the axis of rotation. This method can be very powerful in understanding bulk motion of the spine while being very simple computationally. However, the simplifications that are made prohibit any comparison to *in vivo* values. The main contributions of Bergmark lie in his thorough analysis of the passive structures and their potential role in stability. Also, he proposed a method of simplifying the complex muscular architecture of the back into local and global systems. In general he provided an efficient method of simplifying the spinal system into a set of linear equations that could be solved with some degree of accuracy. Current biomechanical models have advanced well beyond the simple inverted pendulum model, yet aspects of Bergmark's work are still used today.

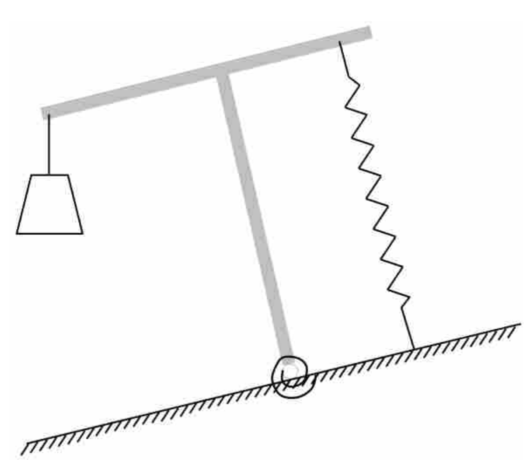

**Figure 2-10: The Inverted Pendulum Model (a Single Spinal Segment)**

The precursor to the follower load model has often been cited as a paper written by Aspden [9]. It recognizes the insufficiency of many models that either treat the spine as a cantilever or that perform an elastic analysis of the system. A number of models report reactions caused by muscular force that are large enough to be damaging to the soft structures of the spine. As has been shown with the studies already discussed, Aspden reveals that most of the models in the literature contain a sufficient number of parameters that may be adjusted within physiological limits such that the model can be made to agree with virtually any set of measurements. Therefore, they provide only limited insight into the actual response of the spine. This study treated the spine as a structural arch and explored the bounds for the applied loads that would allow it to remain stable. The key contribution made is that the resultant force vector of all forces (internal and external) imposed on the spine must remain within the column of the spine to induce stability. This "thrust line" is affected by changes in posture and by the addition of weight to the system. In activities such as flexion to lift a weight off the ground, the muscle forces, body weight, and IAP all work together to cause the thrust line to be enclosed by the spinal column. Not only are the forces important, but the natural flexibility of the spine allows its shape to be adapted such that the thrust line is enclosed and the magnitude of this force is minimized. In the standing posture, this means that the lordotic curvature of the spine is crucial in being able to adapt to the gravity loads that are imposed. Rather than being the cause for instability, the curvature is inherently vital to achieving stability. The present work will show agreement with these observations.

## 2.5 Summary

This chapter has identified a number of key principles of spinal stability that will be important as the current work is presented. First, that postural parameters such as sacral rotation and linear positioning of the vertebrae have the ability to add stability. Second, that the body weight of each vertebral segment applied at its respective CM location has the ability to provide flexion moments at each level that help stabilize the spine. Third, that despite the ability of the follower load to replicate some physiological conditions *in vitro,* there has not been a reasonable explanation as to the source of this load, and that the models that have not imposed the follower load constraint on the musculature have recorded results that are much closer to adhering to the 5% MVC requirement. Fourth, that other passive structures such as the thoracolumbar fascia and the iliolumbar ligament have been ignored in current models and could potentially add to passive stability.

In the studies discussed in this chapter, a disproportionate amount of emphasis has been placed on active (muscular) stabilization of the spine while the passive structures have not received sufficient attention. Consequently, the muscle activations reported in many of the studies are unreasonably high. The present work will identify equilibrium postures of the lumbar spine and their relative stability, features which allow the spine to remain erect in the neutral standing posture without violating the 5% MVC criterion. The following chapter is a technical journal article that was submitted to *The Spine Journal* and is currently under review.

# 3 EVIDENCE FOR A PASSIVE PARADIGM IN LUMBAR SPINAL EQUILIBRIUM IN THE STANDING POSTURE

## 3.1 Introduction

The compressive load in the *in vivo* lumbar spine has been estimated to approach 1000 N in standing and walking and up to several thousand Newtons in strenuous activities such as lifting [1]. Conversely, researchers have found that when the *in vitro* lumbar spine is subjected to a vertical compressive load it will buckle under loads less than 100 N [2-4]. This disparity between the loading capacities of the *in vivo* spine and the *in vitro* free-standing cadaveric spine has driven many researchers to seek after a set of loading conditions that will replicate spinal stability while satisfying physiological constraints (please note that in all instances in the present work, the term *stability* refers to mechanical stability rather than clinical stability.) However, the nature of *in vivo* testing prohibits the use of many invasive methods that could most accurately characterize the relationship between the active and passive support systems of the spine. Further, the static and kinematic redundancy of the spinal system makes numeric analysis extremely difficult. Consequently, despite the numerous biomechanical models of the spine that have been presented, the precise loading conditions that govern it are not completely understood [5, 6].

*In vivo* metrics such as intradiscal pressure, intersegmental rotation, internal moments and shear stress are typically used as the basis for determining the ability of a biomechanical model of the spine to simulate physiological conditions. While these are important requirements that must be satisfied, there are other essential physiological constraints that must not be overlooked. One such constraint is a restriction that should be placed on the level of muscle activation in the neutral standing posture.

The spine is stable in the standing posture, yet the difference in energy consumption between standing and supine resting is minimal [7, 8]. Maintaining such a low amount of metabolic expenditure requires that skeletal muscle activation remain at a minimal level. In an EMG-based *in vivo* study, Cholewicki and McGill [9] confirmed the ability of the lumbar spine to remain stable with less than 3% maximum voluntary contraction (MVC) activation in the lower multifidus muscles. Additionally, if an activity such as standing is to be sustained for an extended period of time, muscle activation must remain below the threshold for inducing muscle fatigue. Jonsson [10] reported that prolonged muscle activation at or above 5% MVC would result in muscular pain and fatigue. It is therefore reasonable to assume that any muscular activity associated with stabilization of the lumbar spine in the neutral posture should not exceed 5% [8]. This is an important criterion that must be satisfied if a spine model is to be considered physiologically feasible.

The literature contains a number of spine stability models that provide invaluable insights into the types and magnitudes of the loads that might be involved in the stabilization of the spine during standing posture. The follower load, a nearly purely compressive loading condition proposed by Patwardhan et al. [4], offers a loading condition that creates stability in the *in vitro* lumbar spine. Working on the assumption that the follower load is created within the *in vivo* spine by the action of the local muscles [11] surrounding it, Patwardhan et al. [1] and Han et al. [12] attempted to quantify the magnitudes of the muscle forces needed to generate the follower load in the frontal and sagittal planes respectively. They were successful in simulating the follower load using muscle forces and showed increased stiffness of the intervertebral joints under this loading condition. Studies conducted by Rohlmann et al. [6, 13], Wilke et al. [14, 15], and Zander et al. [16] explored the utility of the follower load in conjunction with global muscle support and simulated body weight. They demonstrated that with the presence of global muscle activation the magnitude of the follower load needed for stability in the standing posture was greatly reduced. Shirazi-Adl and others [17-22] have studied the sensitivity of spine stability to changing postural parameters such as T1 translation in the sagittal plane and sacrum rotation. They identified the importance of accurate simulation of the distributed weight of the torso and careful modulation of postural parameters.

Though the objective function of the majority of these studies is to minimize muscle stress, they all report muscle activations that violate the 5% MVC criterion in the standing posture. Table 3-1 shows a representative sample of these studies and the muscle activations required by them. It reports the maximum muscle contraction found in each model, as well as any muscles that each model indicates are in violation of the 5% MVC requirement. If not explicitly stated in the model, the %MVC for each muscle was calculated using Equation 2-2 where $F_m$ is the muscle force reported in the study, $k$ is the maximum contractile force constant from [23] and *PCSA* is the physiological cross sectional area from [24] and [19]. In light of the consideration of muscle expenditure as an essential criterion for spinal stability in the neutral posture, the present work seeks to identify stable postures of the lumbar spine that result in an equilibrium configuration that requires zero external force to maintain the upright posture when loaded under body weight.

**Table 3-1: Maximum Contractions Found in Select Studies**

| Author | Han et al. | Zander et al. | Shirazi-Adl et al. | Patwardhan |
|---|---|---|---|---|
| Max % MVC | 100% | 17.2%* | 96.6%* | 75.3%*† |
| Other muscles violating the 5% MVC criterion | Rotatores, Latissimus Dorsi, Longissimus, Intertransversarii, Interspinales | Multifidus, Iliocostalis, Longissimus, Erector Spinae | Multifidus, Iliocostalis, Longissimus, Iliopsoas, Quadratus Lumborum | Multiple |
| [reference] | [49] | [52] | [6] | [1] |

* The %MVC was calculated for all muscles used in the study and the largest reported here.
† Specific data for the individual muscles was unavailable. It was assumed that all muscles in [20] were acting simultaneously on each vertebral level.

A number of other components have been suggested to contribute to the stability of the spine. Some of the most prevalent include the iliolumbar ligament and the thoracolumbar fascia. Some studies [25, 26] have shown that intra-abdominal pressure (IAP) contributes significantly to the overall stability of the spine. More recently IAP has been shown to have a negligible effect in standing [13]. While the iliolumbar ligament and thoracolumbar fascia will be briefly examined in this study, IAP will be neglected due to the disagreement in the literature and based on findings that it is nominally low during standing (20 mm Hg [27]).

While the contributions of these and other passive structures are significant, it is unlikely that the lumbar spine is stabilized entirely by its passive structures alone during standing [1, 8, 14, 15, 28]. However, *we hypothesize that the neutral standing posture corresponds with a position of passively unstable equilibrium.* This hypothesis implies that any sustained muscular activation in excess of 5% MVC in the neutral standing posture is unreasonable and unnecessary. Additionally, we propose that postural parameters such as sacral tilt, modulation of the center of gravity (CG) of the upper torso, and the distribution of mass in the torso are all important factors which can provide transient passive stability to the spine. The present work utilizes *in vitro* testing of five human cadaver spines to investigate this hypothesis and to characterize the passive stability of the compressively loaded lumbar spine in the neutral posture in the absence of muscular support.

### 3.2 Materials and Methods

Each spine was loaded with a simulated body weight and postural parameters were subsequently altered within the physiological range of neutral standing posture to explore potential positions of passive equilibrium and their relative mechanical stability. Equilibrium positions can broadly be categorized as either stable or unstable. To illustrate the difference between a stable and an unstable equilibrium position, consider a ball on the top of a hill. With precise initial positioning at the top of the hill, and in an ideal world, the ball would be able to remain on the top as long it was not perturbed from its initial position. At this position the ball is in unstable equilibrium. On the other hand, a stable equilibrium position corresponds to the ball being at rest at the bottom of a valley. Any perturbation would result in the ball returning to its initial position. This study aimed to not only identify the existence of passive equilibrium configurations of the lumbar spine, but to explore their stability.

#### 3.2.1 Specimen Preparation

Five cadaver lumbar spines (T12 – sacrum), obtained through accredited tissue banks under an IRB-approved protocol, were used in the study (see Table 3-2 for age, height and weight). The surrounding soft tissue was dissected, leaving all ligaments and discs intact. A saline spray was applied to the spine every five to 10 minutes throughout the duration of dissection and testing to prevent dehydration. The T12 vertebra and sacrum were fixated using a polyester resin (Bondo®, 3M Corp) and anchor screws. Special precaution was taken to minimize freeze-thaw cycles and each spine was tested immediately after dissection to minimize autolysis of the soft tissue.

**Table 3-2: Age, Height, and Weight Statistics for all Test Specimens**

| Specimen | Sex | Age | Weight (kg) | Height (m) |
|---|---|---|---|---|
| A | M | 67 | 52.16 | 1.72 |
| B | M | 50 | 36.28 | 1.78 |
| C | F | 79 | 36.74 | 1.60 |
| D | F | 47 | 113.40 | 1.50 |
| E | F | 56 | 58.97 | 1.55 |
| Mean | | 59.8(13.2) | 59.5(31.7) | 1.63(0.12) |

### 3.2.2 Test Protocol

In the experimental setup the distributed weight of the *in vivo* lumbar torso was simulated with discretized segmental masses in order to provide a more accurate representation of the body weight than can be obtained with a single lumped mass. This was done by dividing calibrated CT images into transverse-plane slices corresponding to each vertebral level (L1-L5). The location and magnitude of the center of mass (CM) for each of these segments was then calculated by correlating pixel intensity with a calibrated material density value. The segmental mass properties (location and magnitude) from three CT images were averaged with values published by Pearsall, Reid, and Livingston [29] and Keifer, Shirazi-Adl, and Parianpour [18]. The upper body weight (UBW) of all tissue superior to the L1 vertebra was simulated with a free-hanging weight applied at the T12 vertebra. Table 3-3 shows the average values of the segmental masses and the UBW as used in this study, as well as their respective distances of application from their vertebral center.

**Table 3-3: Segmental Mass Parameters Applied to each Spine**

| Level | Weight (N) | Location (cm)* |
|---|---|---|
| UBW | 311.4 | Variable |
| L1 | 17.27 | 3.18 |
| L2 | 18.01 | 2.45 |
| L3 | 18.33 | 1.56 |
| L4 | 18.98 | .99 |
| L5 | 19.58 | .84 |

* The location of the segmental weight is measured in cm anterior of the vertebral center

A custom testing apparatus was constructed to allow for the adjustment of the sacrum angle, location and magnitude the UBW, and the application of the segmental weights from Table 3-3. The UBW was applied bilaterally to the superior endplate of the T12 vertebra via cables and deadweight. The relative location of this weight (*d* in Figure 3-1) was adjustable in the anterior-posterior (A-P) direction to explore the effect of torso CG location on the equilibrium position. *d* was measured as the distance from the T12 vertebral center and is defined as positive in the anterior direction. A graphical representation of the setup is shown in Figure 3-1 and an image of a fully-loaded lumbar spine is shown in Figure 3-2.

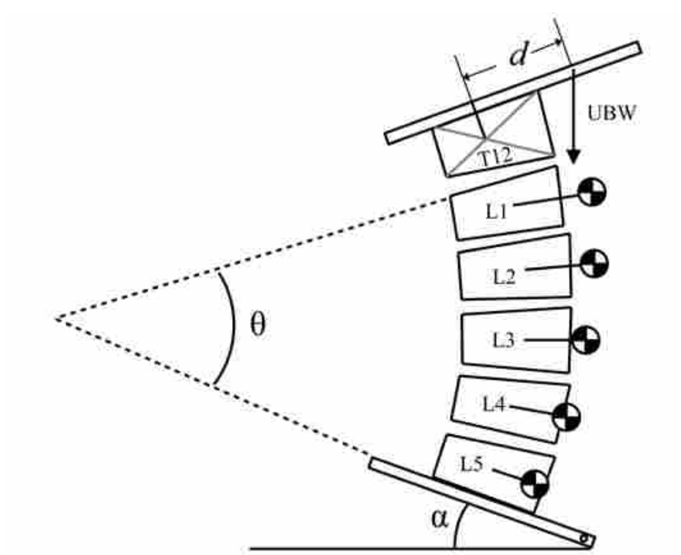


**Figure 3-1: The Experimental Setup**

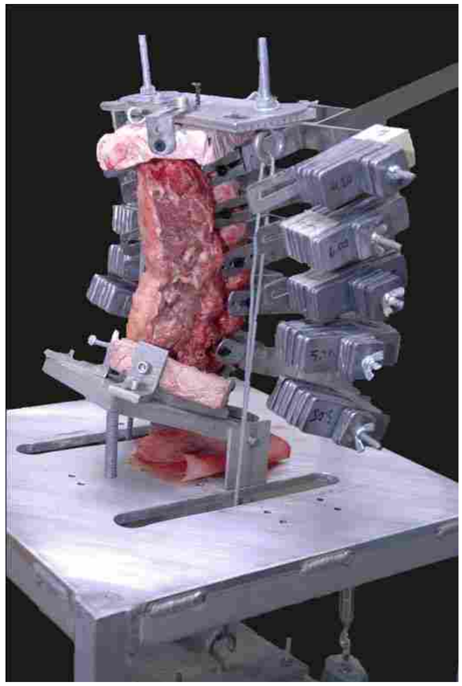

**Figure 3-2: A Fully Loaded Lumbar Spine**

Average values taken from the literature for the sacrum angle (α) and UBW location (*d*) were used as a starting point for the equilibrium point search. From this starting position, α and *d* were manually adjusted until an equilibrium point was discovered (a postural configuration that required no external stabilization.) The Cobb angle, θ, measured between superior endplates of L1 and S1 [30, 31], was recorded and used as a general measure of overall lordosis and as a metric for comparison to published values.

A stability analysis was performed on all of the spines but one (spine E) to quantify the amount of external (muscular) stabilization that is required as the spine is perturbed away from the initial equilibrium point. As each spine was allowed to rotate into flexion and extension in incremental steps, the magnitude of the A-P restraining force required to prevent hypermobility was measured with a linear spring scale attached horizontally to the T12 vertebra. During the procedure sagittal plane digital images were taken of the spine for post-processing. These were evaluated in Analyze 8.1 (Analyze Direct, Inc., Overland Park, Kansas, USA) to measure intersegmental angles between adjacent levels at each increment of flexion/extension.

With the equilibrium position characterized, α and *d* were then systematically altered within a feasible physiological range to identify additional equilibrium points for each spine. This was done to characterize a predictable range over which the spine should be expected to manifest an equilibrium position.

## 3.3 Results

### 3.3.1 Equilibrium

An equilibrium position in which the spine remained free-standing without any external support was discovered for each specimen. After a reasonable time delay (2-7 seconds) the spine would begin to “drift” into flexion or extension due to viscoelastic creep. Table 3-4 reports the values of the location of the upper body weight, sacrum angle, and Cobb angle that produced the equilibrium point.

**Table 3-4: Postural Parameters at the Equilibrium Point**

| Specimen | UBW Location *d (cm)* | Sacrum Angle α (°) | Cobb Angle θ (°) |
|---|---|---|---|
| A | -0.5 | -28.9 | 37.5 |
| B | 1 | -31.4 | 40.9 |
| C | 0 | -44.3 | 44.3 |
| D | 1 | -53.3 | 68.5 |
| E | 2.4 | -51 | 52.6 |

### 3.3.2 Stability Analysis

The stability analysis was performed to investigate the resistance of the lumbar spine specimens to perturbations away from equilibrium. As the spine was moved away from the equilibrium point, an increasing amount of force was required to maintain the upright posture, indicating that the equilibrium points investigated in this study were in unstable equilibrium. The results of the stability analysis are shown in Figure 3-3. The reported force represents the horizontal component of the muscular force that would be required at T12 to keep the spine erect. This is a simplified representation of the muscle components involved and is not meant to be interpreted as the force of any single muscle or any particular group of muscles. In Figure 3-3, extension data was unavailable for spine A because the equilibrium point was close to the full extension limit of the spine.

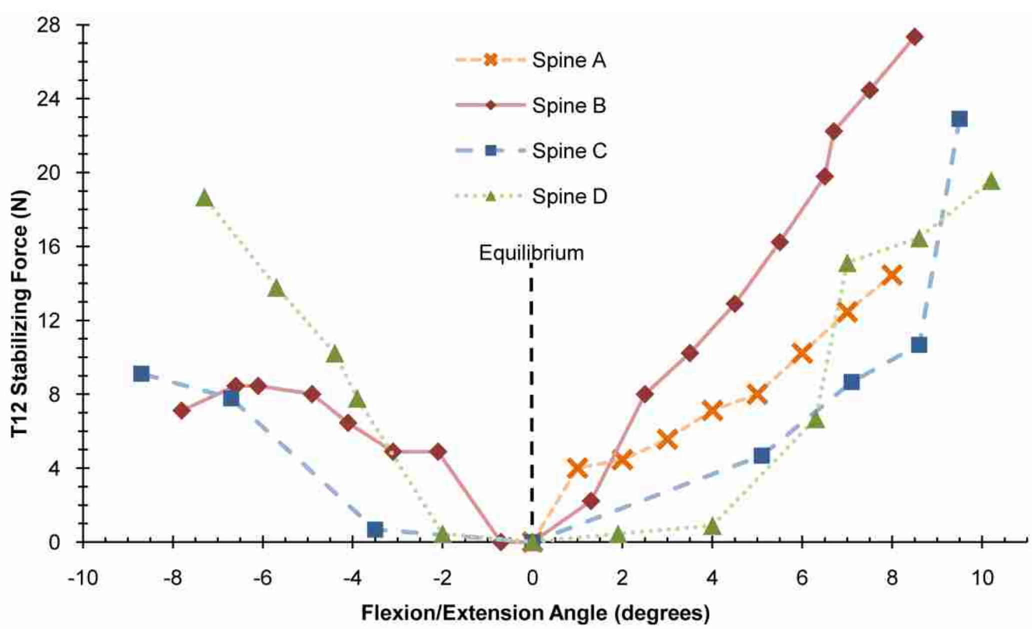


**Figure 3-3: The Results of the Stability Analysis**

The curves for spines C and D in Figure 3-3 exhibited a distinct “neutral region” around the equilibrium point where very little force was required within a several degrees of flexion or extension. Once a certain threshold of UBW translation was reached, the curves became much steeper, indicating that much more external (i.e., muscular) force was required for stabilization.

Though spines A and B did not show as distinct of a neutral region, they followed the same general trend of requiring more active stabilization with increased flexion or extension angle.

### 3.3.3 Other Equilibrium Points

As the design space for α and *d* was explored, each spine tested was found to have several equilibrium points, i.e. there were at least two combinations of α and *d* that would enable the spine to remain upright without any additional support. All equilibrium points were found to exist within a narrow range of α and *d*. Figure 3-4 shows the combinations of these variables that produced an equilibrium point. The shaded region encircling the equilibrium points represents a two standard deviation envelope for α and *d* at equilibrium. The average values of UBW location and sacrum angles as well as resulting Cobb angles for all equilibrium points identified in the study are reported for each spine in Table 3-5. Also reported are the average values for these parameters as published in the literature. Though multiple equilibrium points were identified for every spine, it was not possible to perform an exhaustive search of the design space. Therefore, the shaded regions in Figure 3-4 do not necessarily represent the absolute bounds wherein equilibrium may be expected to exist. Due to the natural variability between spines, the range of α and *d* that produced equilibrium was unique to each spine. Since only two equilibrium points were identified for Spine A, the standard deviation envelope for that spine was not included in Figure 3-4.

**Table 3-5: Postural Parameters Averaged Between all Equilibrium Points for each Specimen**

| Specimen | UBW Location *d (cm)* | Sacrum Angle α (°) | Cobb Angle θ (°) |
|---|---|---|---|
| A | -1 (0.71) | -30.2 (1.8) | 40.1 (3.6) |
| B | 0.58 (0.49) | -32.2 (1.3) | 41.8 (2.5) |
| C | 0.13 (0.25) | -43.8 (1.9) | 43.1 (3.1) |
| D | 1.50 (0.32) | -54.3 (2.1) | 73.0 (5.2) |
| E | 1.78 (0.44) | -51.5 (1.7) | 52.0 (4.0) |
| Mean | 0.93 (0.94) | -44.9 (12.8) | 52.2 (13.0) |
| Literature (ref) | 1.77 [28, 46, 56] | -51.6 (7.8) [57, 58] | 59.4 (12.5) [7, 57] |

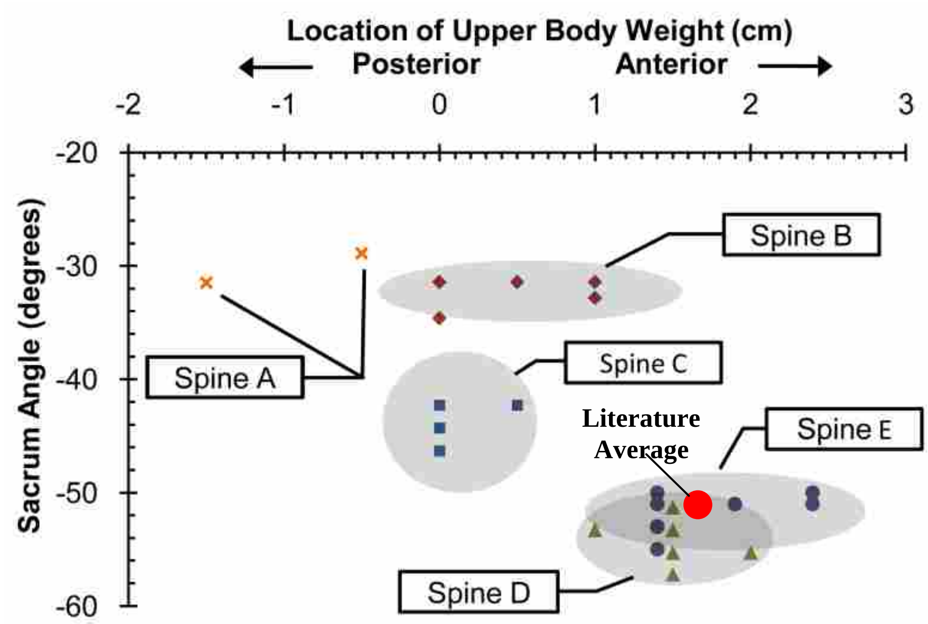


**Figure 3-4: Sacrum Angle and UBW Location for all Equilibrium Postures**

## 3.4 Discussion

That the spine was able to stand freely within several degrees of the equilibrium point is an important finding. It is an indication that muscles need not be involved in stabilization around the standing posture until a threshold value of flexion or extension is reached. The Cobb angles and sacrum angles measured in this study define the postural configuration of the spine at equilibrium. In general these measurements fell within the ranges for the standing posture as reported in the literature (see Table 3-5). Thus, the neutral standing posture is an energetically favorable equilibrium configuration of the spine. It is therefore reasonable to conclude that with the fine motor control of the local and global muscles required to adjust the sacrum angle and the location of the CG of the upper body, the lumbar spine is able to stabilize with less than 5% MVC in the surrounding musculature. Muscles such as the multifidus and rotatores are particularly well suited for the fine postural adjustments that are required to maintain the equilibrium posture [14].

### 3.4.1 Characteristics of Spinal Equilibrium

Variations in $\alpha$ and $d$ affected the overall angle of lordosis (the Cobb angle) at which equilibrium occurred. Though multiple equilibrium points were discovered for each spine, not all exhibited the same degree of stability. Some equilibrium points were quicker to succumb to viscoelastic creep than others. This suggests that even though there is a range of postures over which the lumbar spine is passively stable, there exists an optimal posture that is the most energetically favorable.

In this study the weight of the upper torso was applied at each lumbar segment and by a lumped mass at T12, which causes internal moments at each vertebral level. Without some initial positioning of the CG of the trunk weight, these moments will cause buckling [1-4]. Shirazi-Adl and Parianpour [17] suggested that by regulating the location of the CG of the torso, the body creates the optimal moments that are needed to relieve the load on the musculature surrounding the spine. Additionally, they observed that simulating the lower torso weight with anteriorly off-centered segmental weights creates flexion moments that stabilize each level and, coupled with positioning of the upper body CG and an optimal sacrum angle, greatly enhances the load carrying capacity of the spine. These observations were confirmed in the present study.

### 3.4.2 Intersegmental Angles

As the spines were moved through flexion and extension in the stability analysis, it was of interest to observe the individual contributions of each lumbar segment to the overall quasi-static motion of the spine. The average changes in intersegmental angle for each spinal segment are reported in Figure 3-5. The heights of the columns in Figure 3-5 represent the angular change in the individual segments as the spine is perturbed away from the equilibrium position. On average the spines were moved through a flexion angle of ±8$^o$ (i.e. $\Delta\theta = \pm 8^o$).

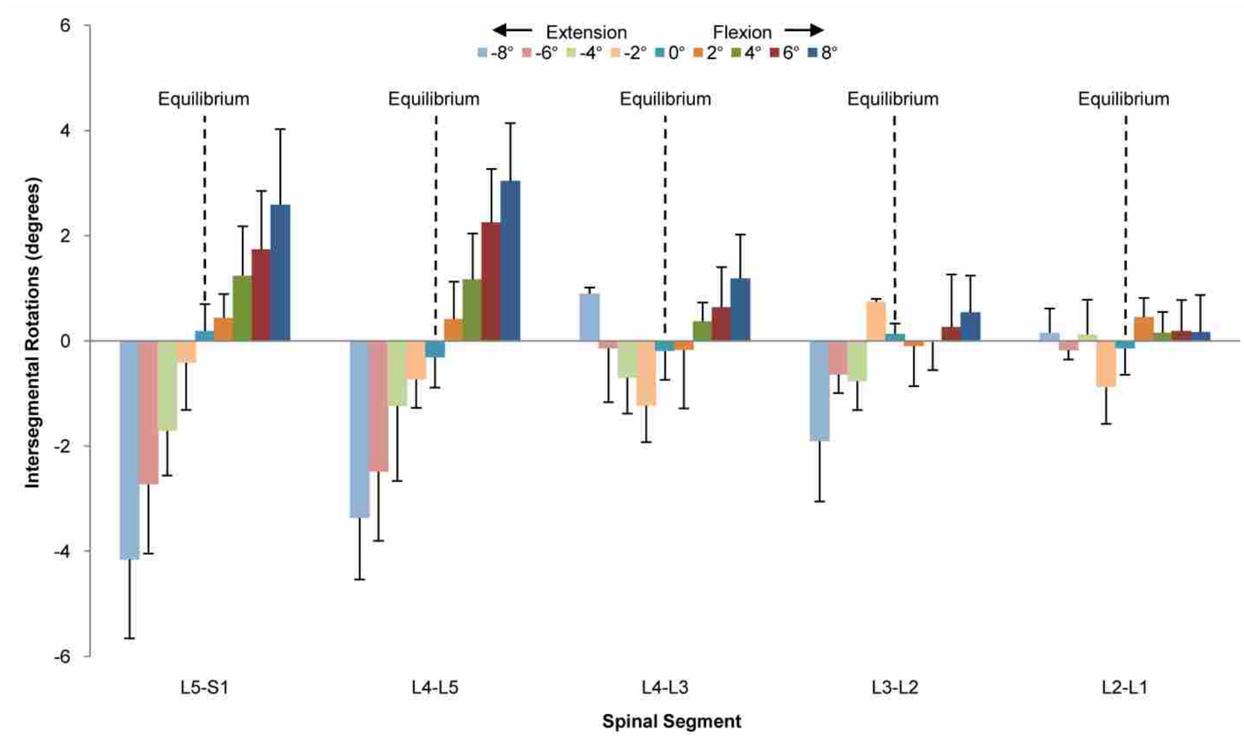


**Figure 3-5: Change in Intersegmental Angle During Stability Analysis, Averaged for all Spines**

Figure 3-5 identifies a very interesting trend in the overall motion of the spine. The L5-S1 and L4-L5 segments experience much more mobility than all other segments. A study by Renner et al. [33] reported a similar trend with more segmental motion in the L5-S1 and L4-L5 segments than in other segments. The intersegmental rotations for the L4-L5 segment are in close agreement with those reported by Rohlmann et al. [13].

### 3.4.3 The LOA of the Gravity Load

The line of action (LOA) of the upper body weight is defined by a plumb line dropped from the CG of the UBW. As the spine is allowed to rotate into flexion, the LOA translates anteriorly by the horizontal distance δ (see Figure 3-6). The location of the LOA with respect to the sacrum was shown to have an important effect on the stability of the spine and on the definition of the equilibrium point. In general, at the equilibrium point the LOA intersected the S1 superior endplate near its geometric center. The standard deviation of the LOA for all equilibrium points was δ = 0.78 cm. This is an indication that the lumbar spine is able to adapt to postural changes in such a way that an energetically neutral position can always be found as long as the CG of the torso can be positioned above the sacrum. Significant deviation of the LOA in flexion or extension requires external input (muscle activation) to prevent the spine from collapsing. Rohlmann, Zander, and Bergmann made the observation that the upper body weight has its center of gravity positioned in front of the spine [6]. However, with the CG of the torso more centrally positioned above the sacrum as found in the present study, the need for large extension moments is decreased and therefore the need for muscle activation is reduced.

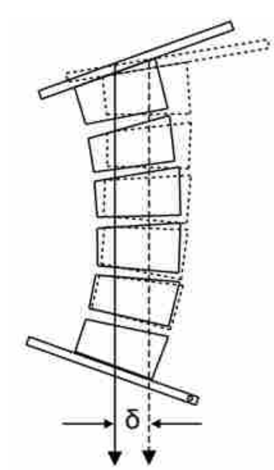


**Figure 3-6: The Translation of the LOA of the UBW During Flexion**

This finding is complementary to a study performed by Aspden [26]. He treated the spine as a structural arch and characterized the path that the internal load of the spine must follow in order to maintain stability. The path of the internal load, or thrust line, was shown to induce stability as long as it remained inside the physical column of the spine. He concluded that the flexibility of the spine allows it to adapt is geometric configuration in order to enclose the thrust line within the anterior column. This theory has been corroborated by the present study. The LOA at the equilibrium posture falls in such a way that it is encompassed completely by the lumbar column. Without any muscular input the passive structures alone are capable of self-adjusting their geometric configuration to ensure that the thrust line is enclosed by the column.

### 3.4.4 Limitations

The lumbar spine possesses an inherent ability to stand freely when positioned into the neutral standing posture. Though we have identified postures of passive equilibrium, this is not to suggest that all muscle activation is unnecessary for standing. Initial positioning of the CG of the UBW requires muscular input, and muscles would also be required to reposition the spine subsequent to the viscoelastic creep that was observed in this study. Additionally, as has been shown in the literature, co-activation of the lumbar muscles can provide additional stabilization [8, 21] which may be necessary to protect the spine from unexpected events or balance deviations. Though this study has identified a passive equilibrium position of the lumbar spine, it is not intended to be interpreted as a comprehensive model of spinal stability. A full understanding of the overall stability of the spine cannot be gained without incorporating all active and passive components involved.

This study simulates the body weight of the lower torso with discretized segmental masses applied at each vertebral level. Though we present that this method provides a more accurate approximation of the load distribution imposed on the spine [18] as compared to a single point mass or a cabled follower load, it is unlikely that the weight of the body behaves as a series of rigid segments. It is likely that the visceral organs and their fibrous attachments to the abdominal cavity create a more complex loading condition, which should be investigated more closely.

The values that were used for the UBW and segmental masses were averaged from three different sources. Differences in methodology among these studies are a potential source of error. Further, these average values likely do not provide the same *in vivo* body weight distribution of any of the individuals whose donated spines were used in this study. However, considering the range of values over which equilibrium points were found, it would not appear that the existence of an equilibrium point is particularly sensitive to the load distribution.

While the individual donors for the cadaver spines were not ideal representations of the population mean for height and weight, the good agreement of resultant postural parameters with published values indicates that the specimens yielded reasonably representative results. Two of the spines (A and B) had abnormally shallow curvatures when unloaded, i.e. their Cobb angles were small compared to anthropometric data. These spines reside in the upper-left region of Figure 3-4 indicating that a spine's inherent shape could affect its equilibrium point posture. Specifically, the UBW must be placed more posterior and the sacrum angle reduced in order to achieve equilibrium. Spine C showed signs of osteoporosis and had suffered a complete compression fracture of the entire L1 vertebral body and complete crushing of the T12-L1 disc. The health of these spines may have affected their biomechanical response as reported in Figures Figure 3-3 and Figure 3-4.

### 3.4.5 Other Potential Contributors to Passive Stability

As previously mentioned, the iliolumbar ligament and the thoracolumbar fascia are two passive structures that may be involved in stabilization but have historically not been incorporated into full-spine biomechanical models. While these structures could not be fully integrated into this study, the results offer some insight into their importance.

The iliolumbar ligament (ILL) attaches the lateral aspects of the L5 transverse processes to the postero-superior aspect of the ilium [34, 35]. The ILL has been shown to restrict flexion and extension motion of the lumbosacral joint [36, 37] and to stabilize the lower lumbar spine. In this study the ILL was not present on any of the specimens, and the majority of the motion of the spine away from the equilibrium point was observed to originate in the L4-L5 and L5-S1 intervertebral joints. To determine the capacity of the ILL to supply additional passive stabilization to the lumbar spine, two specimens were instrumented with simulated iliolumbar ligaments using digital load cells (MLP-25, Transducer Techniques, Temecula, CA). As the spines were allowed to move into flexion, it was observed that with 30 – 40 N of force in the ILL, the spine was able to remain in stable equilibrium for flexion angles of up to $2.6^{o}$. Whether a force of this magnitude can be expected to be sustained by the ILL without injury has yet to be determined, however it is very reasonable based on reported cross-sectional areas and reported properties for other spinal ligament [34, 35, 37, 38]. Most importantly, we have demonstrated that the ILL has the potential to provide additional passive stabilization to the lumbar spine. To our knowledge, there are no studies that take into consideration the iliolumbar ligament while attempting to characterize the stability of the entire lumbar column.

Research has shown that tension in the thoracolumbar fascia causes increased resistance to flexion [39, 40]. More specifically, Barker et al. [39] have shown that a tension of 20 N in the lumbar fasciae stiffens the lumbar spine particularly near the neutral zone. This tension would primarily be supplied by the transversus abdominis and oblique muscles. These muscles have a relatively large PCSA compared to the posterior local muscles surrounding the spine (1,576 $mm^2$ for external obliques compared to 211 $mm^2$ for multifidus [19]). It is therefore likely that a tension of 20 N is well within the 5% MVC requirement. As has already been stated, intra-abdominal pressure is not likely to have a direct effect on the stability of the lumbar spine. However, in conjunction with the action of the transversus abdominis and obliques, IAP may have an important effect as a resistive force to the posteriorly-directed action of these muscles at the linea alba. Since the muscles must be constrained to less than 5%MVC, the change in IAP will be small, which has been observed in the literature [27].

### 3.5 Conclusion

This study has identified a passive equilibrium position of the lumbar spine which is able to support the body weight load with no external stabilization. This spinal configuration corresponds to the neutral posture of standing and can be sustained with little muscular input. It is our recommendation that a 5% MVC constraint on muscle activation in the standing posture be incorporated into spinal stability models. Further, we recommend that a more careful examination into the relative contributions of the passive elements of the spine to overall stability be conducted. Particularly, the postural configuration of the lumbar spine in the standing posture, the distribution and positioning of the body weight, the iliolumbar ligament, and the thoracolumbar fascia should be more carefully considered. Further investigation into the spinal control mechanisms necessary to maintain the passively stable posture could offer great insight into overall spinal stability.

# 4 ADDITIONAL INSIGHTS

The technical journal paper comprising the previous chapter was limited in its ability to comprehensively address some of the methods and results of this research due to length constraints on the manuscript. This chapter will cover in greater detail some of the procedures and implications of the results that were not included in the previous chapter.

## 4.1 The Significance of the Equilibrium Posture

The definitive results of this study beg the question as to why the passive equilibrium capabilities of the spine have not yet been described in the literature. This study has clearly identified equilibrium postures for five cadaver spines that correlate very well with the neutral standing posture. Particularly, for spines D and E the average postural parameters at equilibrium were very close to the literature average values. Though a stability analysis was not performed for spine E, the curve for spine D shows a much more distinct neutral region in Figure 3-3 than the other spines, so we can assume that the stability curve for spine E would be similar to that of spine D. It is non-coincidental that these two spines were observed to have greatest amount of overall curvature in the unloaded posture. Aspden [9] made the observation that not only is the spine able to remain stable in spite of its curvature, but that the curvature is in fact *necessary* for stability. The geometric configuration of the spine is therefore a critical parameter in determining its stability. This observation may account for the inability of current numerical and finite element models to produce the equilibrium posture. In this study, equilibrium was found to be highly sensitive to the A-P location of the T12 vertebra, the UBW, and the sacrum angle. In the FE models discussed in this thesis, the initial geometric configuration of the spinal column is typically extracted from either a standing radiograph or from a supine MRI or CT image. The angles between adjacent vertebrae are then kept constant in the analysis, and the loads required to maintain that posture are solved for. Errors in the method of measurement of the intersegmental angles can potentially produce a spinal geometry that is slightly different than the configuration that is required for equilibrium. It has been shown that it is not possible to consistently measure geometry parameters from a medical image without a fair degree of statistical variation [7]. For MRI and CT images, though the curvature of the spine in the supine posture has been shown to differ only slightly from the standing posture, overall lordosis can vary by $5.5^{o}$ [11], which would be sufficient to cause instability as defined by the "neutral region" discovered in this study. Additionally, the use of values averaged over large populations for either spinal geometries or for the loading parameters further dilutes the geometric specificity that is required to produce equilibrium. It is very probable then that the spinal geometries used in these numerical studies are not conducive to the discovery of the neutral region and passive equilibrium. As previously mentioned, the rotational freedom of the individual vertebrae is critical to the spinal column's ability to passively adjust to the loads imposed on it. This is supported by El-Rich [6] who showed that an increased vertical load on the spine did not cause any additional muscular activation, suggesting that the passive structures alone are compensating. Many numerical studies constrain this movement in an effort to replicate the geometry from the medical image. It is therefore not surprising that previous numerical studies have not identified the passive equilibrium capabilities of the lumbar spine. Though average values were used for the weights applied to the spines in the present study, the intervertebral joints were free to adjust their postural configuration and find an equilibrium posture.

In *in vitro* studies, the failure to identify the equilibrium posture may partially be attributed to the bias towards the follower load found in the literature. Since the introduction of the follower load, virtually all cadaveric studies have made use of it because of its simplicity and its apparent ability to replicate the physiologic loading condition. Many studies have been occupied with attempting to rationalize the follower load from a physiological standpoint. Though the lines of action of the posterior local muscles have been shown to act in such a way that the follower load could be produced [49], the forces required by these muscles when acting in isolation are much too large. The results of this study suggest that the follower load alone is not an accurate representation of the *in vivo* loading condition and is irrelevant in the neutral posture. However, it is possible that in conjunction with the equilibrium posture defined in this study, a partial follower load generated within the local muscles could provide additional compressive stabilization without altering the geometric configuration of the spine. Naturally, the muscle activations would need to be below 5% MVC for this contribution to be physiologically feasible. This potentially synergistic interaction merits further attention and research.

The individual FSU has been shown to exhibit a distinct "balance point", or neutral zone around which little force required to maintain stability [48]. If each segment in the lumbar spine can exhibit this balance behavior on an individual basis, there is no reason to believe that all the segments of the bulk spine cannot simultaneously be in their balance point configuration and thereby require zero external stabilization. This study has shown that under the proper loading conditions, the spine has a natural disposition to produce this bulk balance point. In the clinical setting, full disc replacement devices should therefore not only be able to replicate the biomechanics of the individual segment, but reproduce the motion of the bulk system as well. As has been previously mentioned, additional low-level muscle contraction may be needed for additional stabilization and protection of the spinal column.

The 5% MVC activations calculated in this study are only approximations since it is extremely difficult to measure muscle force directly. They are based on a single maximum force constant, $k$ = 25 N/cm$^2$, which may not necessarily reflect the *in vivo* capabilities of the muscle. The inconsistency in this valve as reported in the literature is a source of potential error. The effect of using a different value for $k$ in this thesis will only alter the acceptable limits for muscle contraction. For example, Bogduk [20] reports a $k$ value of 46 N/cm$^2$, which would increase the 5% MVC limit for the multifidus muscle from 2.638 N to 4.853 N. Though the percent increase is significant, these two values are on the same order of magnitude, meaning that the majority of the studies discussed here are still in violation of the constraint. It may be difficult to assign a specific value to the acceptable force limit for any single back muscle, yet the general principle holds true that any activation in standing must be only a small fraction of the maximum.

Though equilibrium behavior has been only been demonstrated in the lumbar spine in this study, it is reasonable to assume that the other regions of the spine (thoracic and cervical) will exhibit similar behavior. In this study the thoracic region was treated as a single rigid mass. This technique has been used in the majority of the other studies discussed in this thesis. The thoracic spine is indeed much stiffer than the lumbar or cervical spines because of the ribcage, which supplies a large amount of passive stiffness to the intervertebral joints. The FSUs of the thoracic spine are very similar to those of the lumbar, with the main difference being in the orientation of the mating surfaces of the facet joints. This similarity, along with the additional stiffness of the ribcage, indicates that the thoracic spine will be even more predisposed toward equilibrium postures. The anatomy of the cervical spine is significantly different from the thoracic or lumbar, so no inference can be made with respect to its passive equilibrium. However since this region is supporting a much lighter load than the lumbar spine it can be assumed that any required muscle activation should be on a low order of magnitude and thereby not contribute significantly to the overall metabolic expenditure. The cervical spine has shown similar behavior to the lumbar under a follower load [47], so it is possible that an equilibrium configuration does exist. The most interesting question then becomes how the three spinal columns work in harmony to remain in a state of minimal metabolic expenditure in standing. This question merits further research.

### 4.2 Experimental Procedures

Though the testing procedure for this study may seem simple and straightforward, countless hours were spent in conceptualizing and devising a robust protocol that would provide consistent results. Additionally, all hardware used in the testing had to be designed and custom manufactured. For a detailed description of all devices and custom hardware used in the testing, see Appendix B: Testing Fixture.

One of the most time-consuming aspects of the testing procedure was the dissection and the preparation of the cadaver specimens. Prior to testing, each specimen was stored in a deep-cycle freezer at -25$^{o}$ C. Before dissection, each spine was thawed at room temperature overnight for approximately 14 hours. Special care was taken to reduce freeze-thaw cycles for the spines, an excess of which can have destructive effects on the soft tissues. Spine E was the first spine to be investigated and was tested a number of times in order to solidify the testing protocol. It was therefore subjected to five freeze-thaw cycles. All other spines underwent only a single cycle. Dissection was performed to remove all muscle and fatty tissue from the anterior portion of the spine. Since un-contracted muscle has very little if any stiffness, the removal of this tissue had little effect on the biomechanics of the spine. The main purpose of the removal of this tissue was to allow clear access to the vertebral bodies so the segmental weight brackets could be attached. A few of the spines came with a portion of the T11 vertebra still attached. This was removed leaving the superior endplate of the T12 vertebra exposed. The majority of the sacrum was removed at approximately the S2 or S3 level with a vibratory cutting tool. The inferior two-thirds of the remaining sacrum and the superior two-thirds of the T12 vertebra were cleaned down to the bone to allow for proper adhesion of the potting material. For each the sacrum and T12, Bondo® adhesive was poured into a 4”X4”X1” square plastic mold and the bone was imbedded as far as possible without contacting the adjacent disc. Because the heat of the reaction of the adhesive had the potential to dehydrate the specimen, it was continuously sprayed with saline until the adhesive cooled.

With the caudal and cranial ends of the spines fixated in the potting material, the caudal end was mounted to the adjustable-angle base plate of the spine tester. Because of the difficulty of continuously measuring the sacrum angle throughout the duration of the testing, the constant relationship between the sacrum angle and the base plate angle was measured with a laser level before the spine was loaded. The base plate angle was then measured throughout the testing so that the sacrum angle could be calculated. With the spine securely mounted, the segmental weight brackets were attached one at a time starting with the L5 vertebra moving up to the L1 vertebra. A metal plate was then attached to the superior endplate of T12 and cables were connected between it and the deadweight platform hanging below the testing platform. The attachment points for these cables were secured inside a slot in the top plate that allowed for adjustment of approximately 5” in the A-P direction and 1” in the medial-lateral direction. From this point on, it became important to manually support the spine at all times to keep it from collapsing into flexion or extension. Deadweight was then applied in 10 lb increments to the hanging platform until a final weight of 70 lbs was reached. Based on anthropometric data, it was calculated that this weight should be closer to 90 lbs for the average population. However the total body weight for two of the donors (B and C) was approximately 80 lbs at time of death, so the UBW of 70 lbs was used in the experiments to avoid excessive loading of these potentially more fragile specimens.

With the full body weight applied, the sacrum angle and UBW position were placed in their starting locations (the average values from Table 3-3). It became immediately apparent whether these parameters would need to be adjusted in order to find the equilibrium point. For example, since spine B had very little initial lordosis, the initial sacrum angle was much too steep, i.e. the facet joints made contact and prevented any further extension. The sacrum angle therefore had to be significantly reduced. The sacrum angle and UBW position were altered until equilibrium was achieved. For each combination, the sacrum angle and the L1 vertebra angle were measured (and the Cobb angle calculated) with a digital level. Because of the unstable nature of the equilibrium point, the L1 angle could only be measured with an accuracy of approximately $\pm 0.5^{o}$. Every effort was made to measure this angle as consistently as possible.

As multiple equilibrium postures were identified for each spine, it became apparent that there is likely an “optimal” posture that is more stable than the others. The qualitative observation was made that although each equilibrium posture was indeed a configuration of unstable equilibrium, some were quicker to succumb to viscoelastic creep than others. No attempt was made to quantitatively define the "optimal" equilibrium posture, recognizing that since the precise *in vivo* loading conditions for each specimen was not accurately replicated in this study, the *in vitro* optimal posture would have little meaning. The qualitative observation is, however, a relevant indicator that a most efficient posture should exist *in vivo* and the body likely seeks after this posture whenever possible.

During the stability analysis, a digital image was taken at each step away from the equilibrium posture. The camera was placed at a distance of approximately five feet away from the test apparatus, and was manually aligned to be perpendicular to the sagittal plane of the spine. The images were analyzed in an open-source image processing and analysis software package, ImageJ (National Institute of Health). This software was able to determine the angle between any two lines in the image. Since the camera may not have always been aligned parallel to the testing platform in each test, the known angle of the base plate (sacrum angle) was used as the reference angle. The angels of each vertebral level were then measured at each perturbation so that the angular change of each segment could be calculated.

## 4.3 Iliolumbar Ligament Study

As has been previously mentioned, a simulated iliolumbar ligament was attached to two of the specimens (spines B and E) in this study to explore its ability to constrain motion. This study would have been performed on all specimens, but the L5 transverse processes for spines A, C, and D were either broken or completely removed by the tissue bank. In future testing, specific instructions should be given to the tissue bank to take special care to leave the L5 transverse processes intact.

The procedure for the iliolumbar ligament study was very similar to the stability study, but the spines were only moved into a single angle of flexion. Holes were drilled near the ends of the transverse processes and 1/8” braided steel cables were looped through the holes. In order to obtain realistic results, care was taken to simulate the spatial geometry of the ILL. The ILL can be divided into a number of segments which typically lie at some positive or negative angle to the transverse plane [34, 59]. In this study, the simulated ILL was placed parallel to the transverse plane to act at the average line of action of the *in vivo* ligament. The ILL has also been observed to be at an angle of about $45^o$ to the sagittal plane [35, 36, 60]. The ILL cables were placed approximately at this angle using a protractor. Tension was placed in the cables and then the spine was allowed to rotate into a few degrees of flexion, with the load cells measuring the force. Spine B was able to rotate into $2.6^o$ of flexion with a force of 7.5 lbs in the left ILL and 7.8 lbs in the right ILL. Unfortunately, the flexion angle was not measured for Spine E, but it is estimated that it rotated approximately $3\text{-}4^o$ with a force of 6 and 7 lbs in the left and right ligaments respectively. These postures with the attached ILL were considered to be “stable”, i.e. they had no propensity to continue into further flexion. However, if allowed to rotate much further, the force in the ILL would rise steeply, quickly exceeding the breaking strength of the ligament. It is therefore plausible that the ILL is a stabilizer around the neutral posture, but any significant deviation away from the equilibrium posture would likely be accompanied by sacral rotation and muscle activation to prevent damage to this ligament.

# 5 CONCLUSION

The human lumbar spine is a complex biomechanical system. The high degree of static indeterminacy of the system and the difficulties of measuring spinal loads *in vivo* have made it nearly impossible to characterize the precise load-sharing relationship between the active and passive systems. Despite the extensive research that has attempted to characterize this relationship, it would seem that there is not a single universal solution that is applicable to every spine. Morphological and biological variation in the structures and the tissues of the spine from one subject to another would suggest that there is a specific neuromuscular recruitment pattern that is unique to each individual. Though it may not be possible to determine this precise neuromuscular strategy for any single individual, a number of generalized principles can, and have been made that shed light on the load-bearing capacities of the spine. This thesis has shown the weight of the upper body to be supported by the lumbar spine in the absence of muscular input.

This new understanding about the load-bearing capacities of the lumbar spine has the potential to influence the way we approach the diagnosis and treatment of spinal disorders. Particularly, this understanding sheds light on the compressive load that is imposed on the spine in the standing posture. Others have postulated that the compressive load in the spine reaches 1000 N in standing [15], but this work has shown static equilibrium under approximately 400 N. This study has also identified the postural parameters that are critical to obtaining passive equilibrium, namely the Cobb angle, the angle of rotation of the sacrum, and the positioning of the upper body weight. Surgical procedures must therefore avoid altering the spine's ability to freely adjust these parameters. Due to the lack of a more viable solution, one of the most common methods to repair damaged discs is to perform inter-body fusion of adjacent vertebrae. This procedure eliminates intersegmental rotation at the level of fusion and places additional strain on adjacent levels and, in accordance with the results of this thesis, will therefore require additional strain on the musculature to maintain stability. It is therefore imperative that a more dedicated effort be focused on total disc replacement devices and restoring the natural mobility of the segment that will allow it to attain passive equilibrium in the standing posture.

## 5.1 Summary of Contributions

The primary contributions of this research are as follows:

- The discovery that the neutral standing posture is a state of passively unstable equilibrium.
- Compliance with a constraint on muscle activation (i.e. 5% MVC) is necessary and attainable in standing.
- The identification of postural parameters critical to passive equilibrium of the spine, namely lordosis angle, sacrum angle, and upper body weight positioning.
- The identification of passive structures that have the potential to provide additional stabilization: the iliolumbar ligament and the thoracolumbar fascia.
- A call for revision of biomechanical models to comply with the 5% MVC requirement in standing.

## 5.2 Recommendations for Future Work

The results of this study are a promising indication that the lumbar spine is stable in the neutral standing posture with little muscle activation. Since this phenomenon has not been discussed in the literature to date, it is important to verify these results and to conduct further research to determine the sources of the passive equilibrium posture.

This study was limited in its scope and magnitude. Only five cadaver spines were available for testing, and the majority of these either had some degree of degenerative disc disease or the physiological curvature was atypically flat. Further, the average age of the specimens was nearly 60 years. Future work should focus on obtaining a greater number of specimens that are more representative of the average healthy population.

The most promising continuation of this work is to experimentally determine the effects of the iliolumbar ligament and the thoracolumbar fascia on stability in the standing posture. Lumbar specimens with an intact pelvis and iliolumbar ligament could easily be obtained from accredited tissue banks. To isolate the effect of the ILL, the intact specimen should be tested, and then re-tested after the ligaments are transected bilaterally. It is postulated that the ILL will provide additional stiffness, particularly to the L4 and L5 segments. Since it may not be possible to obtain an entire human torso for testing, the simulation of tension in the thoracolumbar fascia may be a more complex challenge. The testing protocol should closely follow a study conducted by Barker [29] where tension was applied transversely to the fascia with spring scales. The effects of the ILL and the thoracolumbar fascia should be investigated independently to avoid combinatorial effects. In reality, these two components may have complex interactions that may only be able to be fully understood with computer simulation and optimization. The incorporation of these components will only add to the static indeterminacy of the system, yet their effects can be studied in isolation without active muscle control.

The most likely source of error for this study is the body weight that was applied to each spine. Though the UBW and the segmental masses were averaged from a number of sources, the specimens used in the study were far from average in their heights and body mass index (BMI), so the weights that were applied are not accurate representations of the *in vivo* body mass. The *in vivo* loading of each spine could be more accurately simulated by customizing the applied weights and their CGs. This can only be accomplished by imaging a much larger sample of volunteers and, rather than taking the average segmental mass properties and applying them to all the cadaver specimens, creating a database of masses and CGs based on height and BMI. Each cadaver specimen would then be matched with the imaged subject that most closely matches its height and BMI. The sensitivity of the equilibrium posture should be explored with respect to these mass parameters. This would give a sense the effect of weight gain or weight loss on the shape of the spine at equilibrium.

Another potential factor in the accuracy of the applied masses is the fact that the images used to calculate segmental masses were taken in the supine position. The body weight distribution in this position is potentially different than in standing, and therefore the CM calculations could be inaccurate. Upright MRI is a relatively new technology that is able to take a full-body image in the standing position. Ideally, the subjects imaged for future studies should be imaged in an upright MRI to get the best representation of the distribution of the body weight in standing. There is, however, a challenge with using this technique: pixel intensity in MRI cannot be directly correlated with material density as it can with CT. Other techniques do exist to calculate the density of biological materials from MRI images, but they are more involved.

The behaviors discovered in this study should be incorporated into a numerical model of the spine for validation. A finite element model is currently being developed at the BYU Applied Biomechanical Engineering Laboratory. This model should be compared against the results of this study and it should be altered, if necessary, to comply with the 5% MVC criterion. Once this validation is completed, other research labs would be encouraged to produce similar results.

Finally, a more comprehensive *in vitro* test protocol should be created. As previously mentioned, this would include passive structures that have not typically been incorporated into biomechanical studies of stability. It should also include a more complete representation of the musculature. Muscle insertions at each vertebral level have the potential to alter the intersegmental rotations as the spine is moved into flexion or extension. A more accurate method of determining intersegmental angles should also be devised. This more comprehensive experimental study will give additional insight into the synergistic roles of the muscles and the passive structures as they relate to stability. Once the behavior of the lumbar spine is well understood, this technique should be expanded to explore the passive equilibrium behavior of the thoracic and lumbar spines, and then incorporate the entire spinal column into a single comprehensive model.

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[60] Rucco V, Basadonna PT, Gasparini D. Anatomy of the iliolumbar ligament: a review of its anatomy and a magnetic resonance study. Am J Phys Med Rehabil. 1996;75:451-5.

## APPENDIX A: CALCULATION OF MASS SEGMENTS FROM CT IMAGES

The CT segmentation method used by Pearsall [46] was employed in the present work to determine segmental mass parameters for three full-body CT images which were obtained from the Utah Valley Regional Medical Center in Provo, UT. This method relies on assessing the image intensity of each pixel of a CT image and assigning a material density value to each based on a linear calibration curve. Image intensity is measured in units of Hounsfield number (HU), where the HU of air is -1000 and the HU of water is 0. Using a linear fit between these two values, and knowing the HU number of any given pixel, the density can be calculated with the relationship:

$$\rho_{pixel} = HU * \frac{\rho_{water} - \rho_{air}}{1000} + \rho_{water} \qquad (A-1).$$

This is a fairly good approximation to calculate the density of any material that is being imaged. However, with high density materials such as bone the approximation becomes slightly less accurate. One way to obtain a more precise relationship is to use a calibration phantom with known material densities that are closer to those that are anticipated to be imaged. For the purposes of this study, the above equation provides sufficient accuracy for the calculation of the mass of body segments.

To divide the full-body CT images into segments at each vertebral level, the raw CT data was imported into AMIDE (by user Andy Loening, SourceForge.net), an open-source CT analysis software package. A region of interest (ROI) box was drawn around each vertebra and all the tissue extending out radially from it. A text file containing the HU number and the XYZ coordinates of each pixel was created for each ROI. These values were imported into a C++ code to calculate the center of mass of each segment (the source code is included at the end of this appendix). The CM locations were then plotted in the AMIDE image on the mid-sagittal plane. The parameter of interest in this study was the distance of the segment CM from the vertebral center of that level. To calculate the VC, two lines were drawn connecting opposite corners of the vertebral body, the intersection of which corresponds to the vertebral center. The distance between the CM and the VC could then be calculated. The CM locations from the CT images were averaged with [46] and [56]. A representative view of a segmented image with CM locations is shown in Figure A-1.

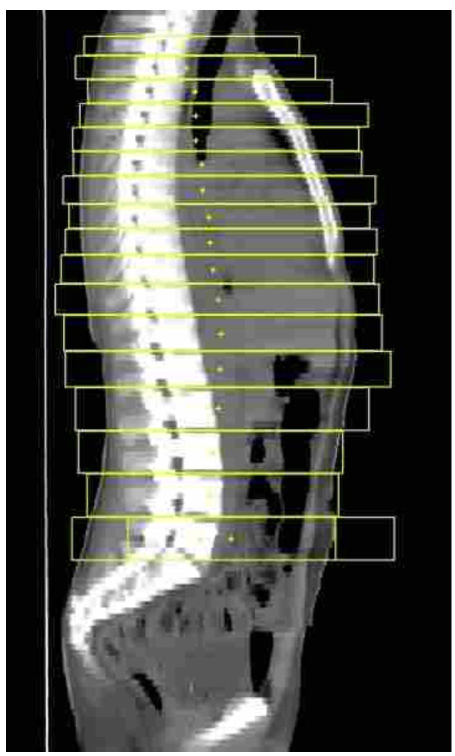


**Figure A-1: Segmentation of a CT Image and CM Locations (Yellow Dots) for each Segment**

A convenient feature of AMIDE is its ability to calculate the average Hounsfeld number and the overall volume of the region of interest. Using this information and Equation A-1, the mass of each segment was calculated. These mass values for the three images were averaged with values from [54] and [46] and are reported in Table 3-3.

There are a number of potential sources of error that were introduced in the execution of this methodology. The resolution of the CT images in the sagittal plane was quite low. Therefore, the calculation of the vertebral centers may not be entirely accurate, though they are likely within ±2 mm of the actual location. Also, the selection of a region of interest to define a torso segment was somewhat arbitrary and subject to the preferences of the person performing the analysis. However, the purpose of the method was not necessarily to determine precisely how much weight is acting on any given vertebral level, but to provide a more reasonable representation of the body weight load than with a single load applied at the most cranial vertebra. As has already been recognized, this method may not be the most accurate representation of the distributed weight of the body, though it is the most accurate that has been presented in the literature. It is therefore less important how an ROI is defined, just as long as the global effect is to simulate the distributed weight of the body by applying masses at each vertebral level.

### Source Code for CM Calculation

```
/***************************************************************************
Project:    CM
File:       main.cpp
Date:       6/8/2010
Notes:            This program is designed to read in raw data point values
                  exported from AMIDE and calculate the center of mass for
                  each ROI (correlated to a vertebral body and its
surrounding tissue.

****************************************************************************/
```

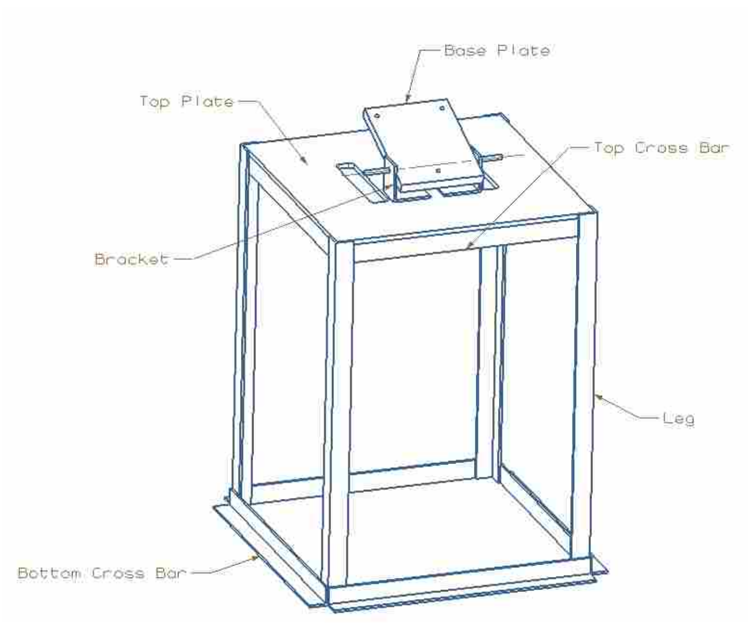
```
#include <iostream>
#include <fstream>
#include <string>
#include<vector>
#include<sstream>
using namespace std;

void COM(ifstream& infile);

int main()
{
	ifstream input;
	vector<string> filename;
	char buffer[100];
	stringstream ss;
	filename.resize(17);
	string name;
	string number;
	name="C:\\Users\\Cesare Jenkins\\Documents\\Research\\Balanced
Spine\\PA1 Data\\T";
	for (int i=1; i<13;i++)//write filenames for thoracic vertebrae
	{
		sprintf(buffer,"%d%s",i,".raw");
		ss<<buffer;
		ss>>number;
		filename[i-1]=name+number;
		ss.clear();
	}
	name="C:\\Users\\Cesare Jenkins\\Documents\\Research\\Balanced
Spine\\PA1 Data\\L";
	for (int i=1; i<6; i++)//write filenames for lumbar vertebrae
	{
		sprintf(buffer,"%d%s",i,".raw");
		ss<<buffer;
		ss>>number;
		filename[i+11]=name+number;
		ss.clear();
	}
	for (int i=0; i<17; i++)//evaluate each file
	{
		input.open(filename[i].c_str());
		if(input.is_open())
		{
			COM(input);
			input.close();
			input.clear();
		}
		else
		{
			cout<<"problem opening file "<<filename[i]<<endl;
			input.close();
			input.clear();
		}

	}

	/* Code for foing the cadaver spine
```

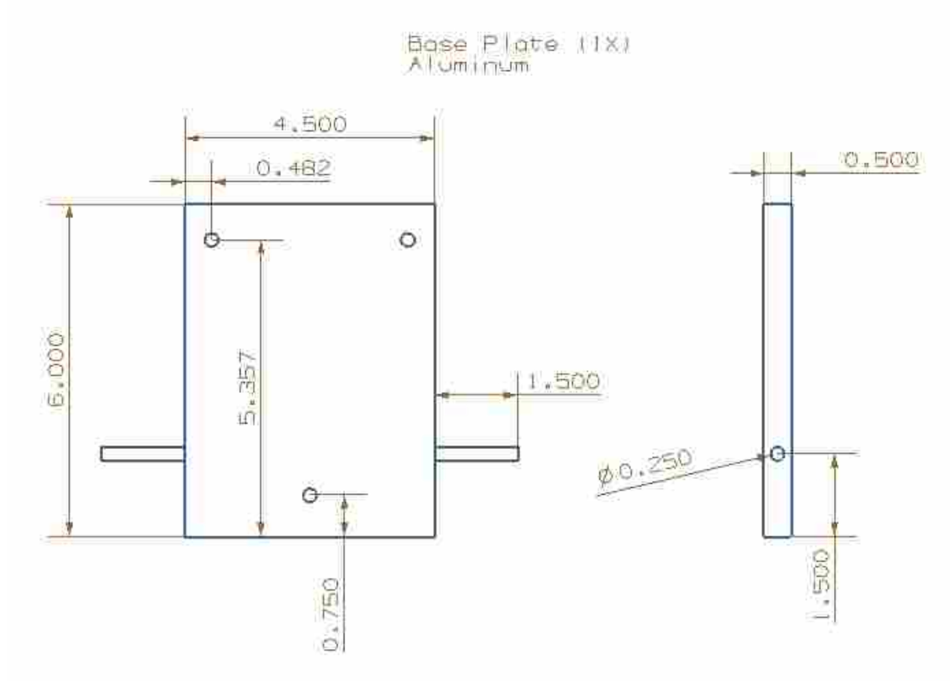
```
	input.open("C:\\Users\\Cesare Jenkins\\Desktop\\RawData_T11_L3");
	COM(input);
	input.close();
	input.clear();
	input.open("C:\\Users\\Cesare Jenkins\\Desktop\\RawData_L4_S");
	COM(input);
	input.close();
	input.clear();*/
	cout<<"Finished"<<endl;
	system("Pause");
	return 0;
}
void COM(ifstream& infile)
{


	int raw_val;
	double x_val;
	double y_val;
	double z_val;
	double wght;
	string stuff;
	double pix_mass;
	double mass(0);
	double xmass(0);
	double ymass(0);
	double zmass(0);
	double x_com;
	double y_com;
	double z_com;
	string ROI;
	//get the first thing from the file, it should probably be a #
	infile>>stuff;
	while(!infile.eof())
	{
		if (stuff=="#")//if you have header info, just read over it until
you get to the end
		{
			infile>>stuff;
			while (stuff!="#" && stuff!="Z")//keep reading until you
get to another # or to the Z
			{
				if (stuff=="ROI:")
					infile>>ROI;
				infile>>stuff;
			}

		}
		if (stuff=="Z")//if you have a Z, there is data to follow
		{
			infile>>stuff;//read the (mm) to get it out of the way
			infile>>raw_val;
			do //read only pixels that are greater than -200
			{
				infile>>wght;
				infile>>x_val;
				infile>>y_val;
```

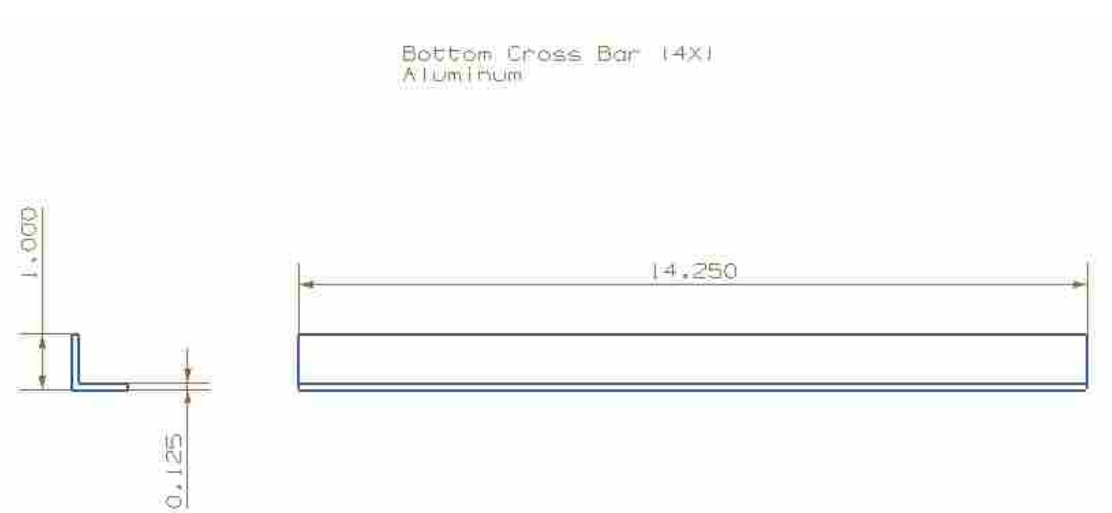
```
                        infile>>z_val;
                        pix_mass=raw_val*0.998303+999.6;//adjust pixel value
to always be positive
                        mass+=pix_mass*wght;//add mass to aggregate mass, do
not multiply by volume since this just cancels out
                        xmass+=pix_mass*x_val*wght;
                        ymass+=pix_mass*y_val*wght;
                        zmass+=pix_mass*z_val*wght;
                        infile>>raw_val;
                  }while(raw_val >= -850);
                  x_com=xmass/mass;
                  y_com=ymass/mass;
                  z_com=zmass/mass;
                  cout<<"The center of mass for "<<ROI<<" is located
at:"<<endl;
                  cout<<"("<<x_com<<", "<<y_com<<", "<<z_com<<")"<<endl;
                  while(stuff!="#"&&!infile.eof())//keep reading the rest of
the values until you reach the next header
                        infile>>stuff;
                  mass=0;
                  xmass=0;
                  ymass=0;
                  zmass=0;

            }

      }

}
```

## APPENDIX B: TESTING FIXTURE

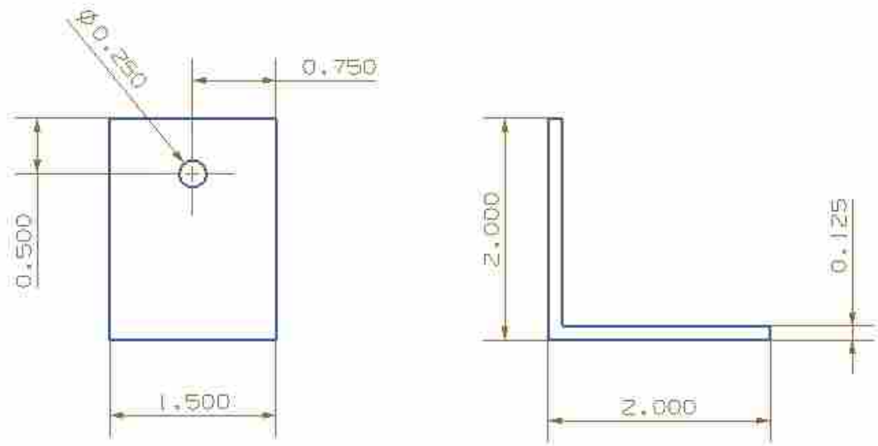
The testing of human cadaver spines under a simulated body load necessitated the design and manufacture of a custom testing apparatus. It was important that the design be simple yet able to support up to 100 lbs. The primary design criteria were that the sacrum angle could easily be adjusted and that a vertical load of up to 70 lbs could be applied to the superior endplate of the T12 vertebra. It was important that the test fixture be resistant to rusting and be easily cleanable; Aluminum was used for all components of the base structure. The testing platform with adjustable-angle mounting plate is shown below. Also found below are the dimensions of all parts of the platform. The slots in the top plate allow for passage of the cables that connect from the T12 plate to deadweight hanging below.



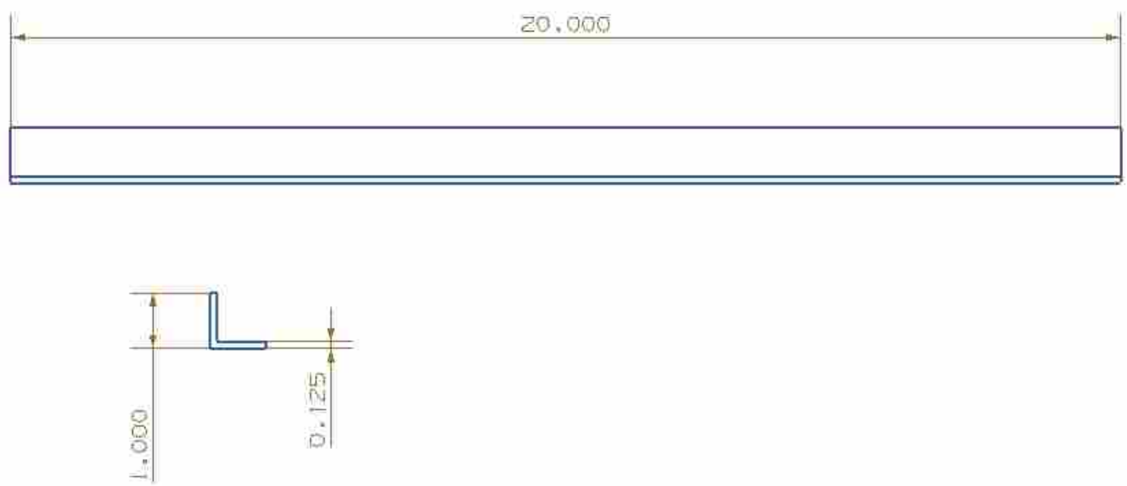

Base Plate (1X)
Aluminum
4.500
0.482
0.500
6.000
5.357
1.500
ϕ0.250
0.750
1.500

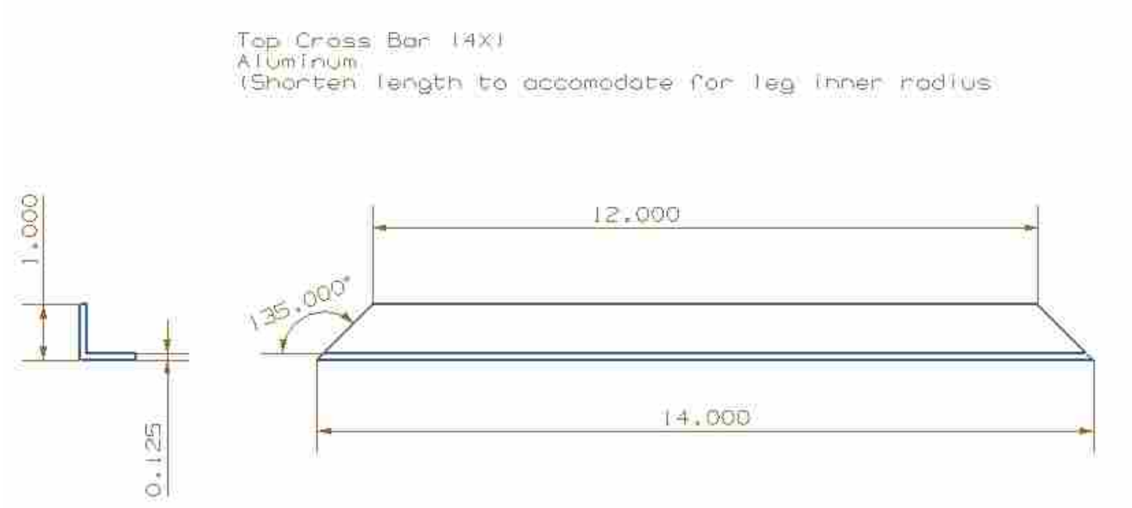

Bottom Cross Bar (4X)
Aluminum
1.000
14.250
0.125

Base Plate Mounting Bracket (2X)
Aluminum




Leg (4X)
Aluminum




Top Cross Bar (4X)
Aluminum
(Shorten length to accomodate for leg inner radius
1.000
0.125
12.000
135.000°
14.000

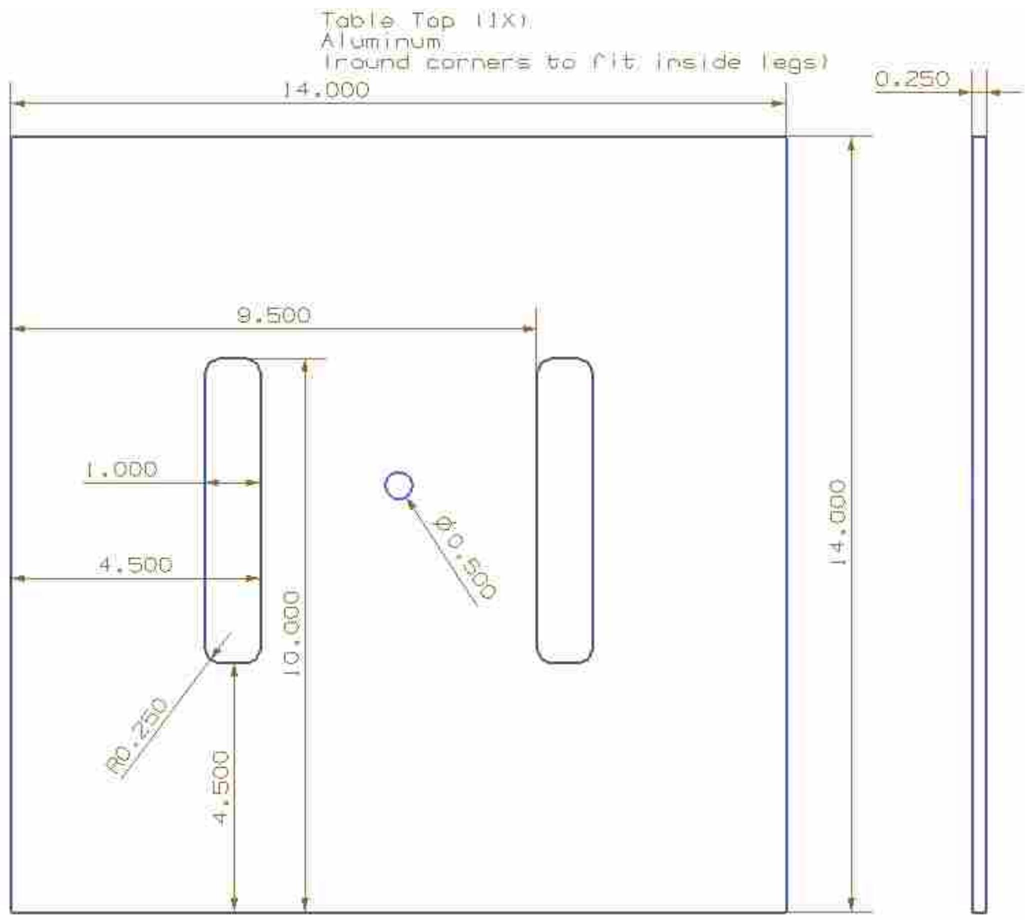
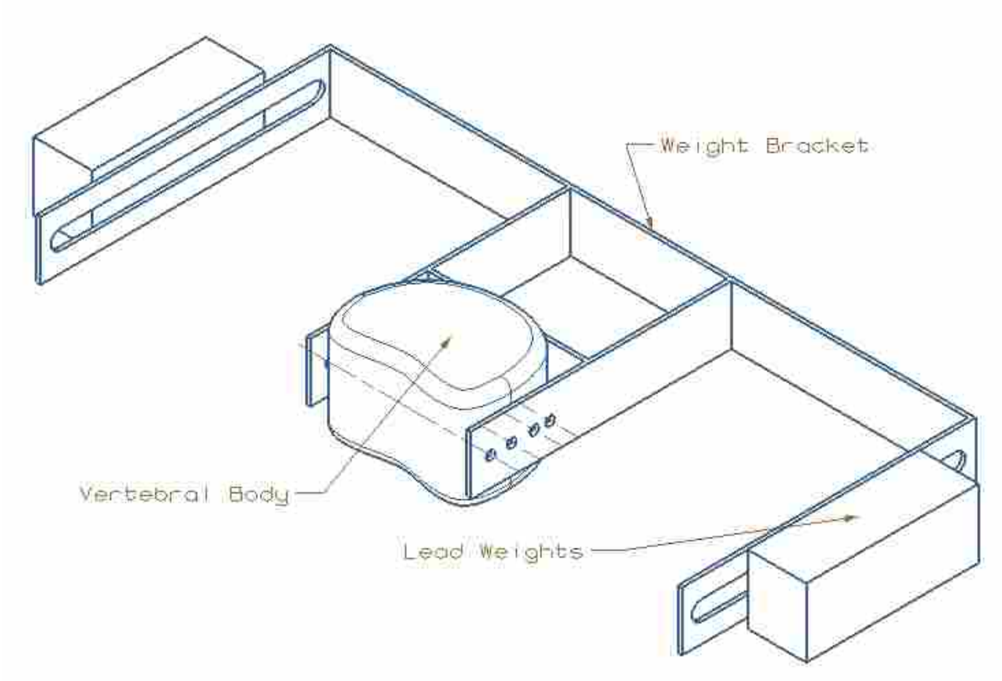


Table Top (1X)
Aluminum
(round corners to fit inside legs)
14.000
0.250
9.500
1.000
4.500
Φ0.500
10.000
14.000
R0.250
4.500

One of the greater challenges of this research was determining the best method to apply an individual weight to each vertebral body. An important consideration was to be able to apply weight at a single point in space corresponding to the segmental masses calculated from the CT images. Additionally, because of the compact structure of the lumbar spine, the weights needed to be as compact as possible. After exploring a number of options, the design shown below was decided upon. The bracket is made up of a number of stainless steel sheet metal strips that are spot welded together. The stainless steel provided a sufficient amount of ductility to be formed into the desired shape while maintaining resistance to oxidation. The lead weights shown in the image below are comprised of a series of thin lead plates. These plates ranged in weight from 2 grams to 5 grams and were formed by melting lead shot and forming it in a rectangular mold. Plates were added to each side until the total weight of the bracket and weights reached the desired segmental mass weight. The weights were then adjusted within the slots in the bracket to be at the CM location from Table 3-3. The weighted bracket was affixed to the vertebral body with three 1-1/4” wood screws: two on the sides and one through the front. Care was taken to align the bracket parallel to the superior endplate of the vertebra. The weights are extended out laterally from the vertebra so that the deadweight cable could pass bilaterally down along the spine and through the slots in the top plate. .




The dimensions for the L3 bracket are shown below. In general the dimensions for all other brackets are similar, with the exception of the arms that attach to the vertebral body: smaller for the cranial vertebrae and larger for the caudal vertebrae